A Wireless, Battery-Free Photoplethysmography-Based Implantable Optical Force Sensor

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***Abstract* — Tactile sensations play a critical role in guiding motor actions. However, several neurological diseases and injuries prevent this somatosensory information from reaching the brain, making coordinated movement difficult for many patients. To restore this crucial sense of touch in patients with nerve damage, we present a novel implantable photoplethysmography (PPG) -based sensor that measures changes in subcutaneous blood flow to infer forces applied to the skin. In this design, a photodiode amplifier circuit is used to detect changes in subdermal absorption of red and infrared light, and a monotonic relationship between the absorbed light and applied force was observed. The sensor receives power and transmits data wirelessly over a near-field communication link to a nearby base unit. The fabricated sensor occupies a footprint of 20.67 mm^2.**

***Index Terms* — tactile sensors, photoplethysmography, sensor interface, body channel communication (BCC), wireless power transfer**

I. Introduction

Tactile sensations are crucial for informing movements in everyday life and allow us to dexterously manipulate objects in our environment. Without this somatosensory information to provide feedback for motor control, precise and reliable motion is very difficult. Strokes, spinal cord injuries, and other medical conditions can cause irreparable damage to somatosensory pathways. To restore this communication pathway between the brain and body, a brain-machine interface could be used to detect tactile sensations in the limbs and wirelessly relay them to a neural stimulator that encodes touch perceptions directly into the brain. To acquire these tactile sensations, existing technologies have integrated sensors into prosthetic limbs [1] or gloves that can be donned over the natural hand [2], but these are limited in their versatility. Full prosthetic limbs are often costly and not medically advisable for paralyzed patients with intact extremities. Tactile sensors integrated into a glove that can be worn over natural skin are prone to slippage and donning these devices may require the help of an able-bodied assistant. External sensors may also be affected by environmental conditions such as moisture, temperature, and abrasion. To overcome the drawbacks of these devices, an implantable sensor can be used to detect forces applied to the skin.

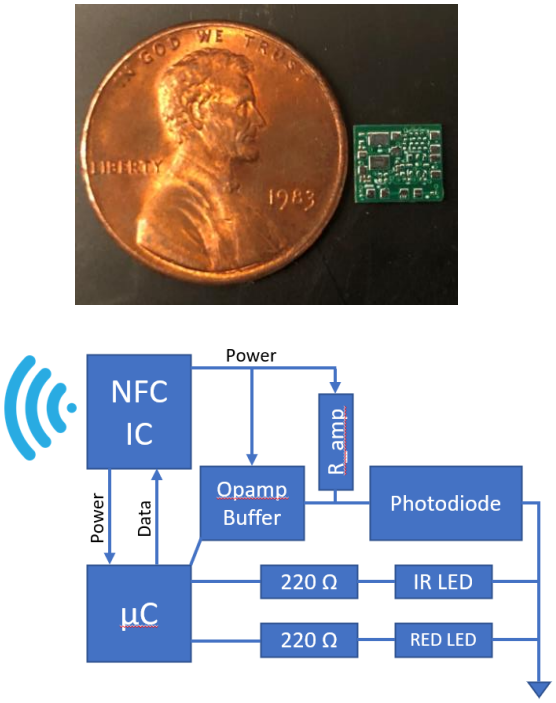

Fig 1. (A) The manufactured prototype with a US penny for scale. (B) A block diagram of the implantable PPG sensor proposed in this paper. An ATtiny 20 microcontroller (µC) switches the LEDs and processes photodiode signals. An NFC chip harvests power and transmits data over a wireless link to an external base station.

To collect tactile information, implantable capacitive sensors [3] have often been used to transduce pressure signals in the human body. However, these sensors are often physically large [3], require additional circuitry to amplify signals due to small active capacitances [4], and are not stable across varying environmental conditions [5]. To mitigate these issues, we propose an implantable optical photoplethysmography (PPG) sensor to measure subdermal changes in blood volume as a metric of tactile pressure. Although PPG sensors are typically

used in the medical setting to measure oxygen saturation, blood pressure, and heart rate, their use as force sensors is motivated by the changes in blood flow through the fingertip in response to pressure, as evidenced by blanching of the skin when grasping objects. Additionally, these sensors are resilient to environmental changes and do not require a large footprint for accurate measurements [6].

While implantable designs have been proposed for cardiovascular applications [7,8], these sensors have not been used for recording tactile signals. Previous work by Kim et al. has also shown that wireless, battery-free PPG sensors can be fabricated using discrete components [9]. In this paper, we present an implantable PPG force sensor that wirelessly relays data and receives power over a near-field communication (NFC) link to a nearby base station. We show that the signal output by this sensor changes monotonically with force applied to the skin. The sensor occupies a footprint of 20.67 mm^2; schematics and sensor layout can be seen in Fig. 1. The tactile data collected by the sensor can be relayed to a neural prosthetic to generate sensations of touch.

II. Background

A. Somatosensation and Proprioception

Somatosensation consists of the sensations produced by our skin and joints and is commonly thought of as our sense of touch. To process this sensory information, the brain integrates signals carried by nerves that connect mechanoreceptors and other sensory organs embedded in the skin to the somatosensory cortex (S1). Activation of this region of the brain is responsible for the generation of conscious touch percepts. The integration of somatosensory signals also provides critical information about the positional alignment of the body, a sense known as proprioception.

Although severe spinal cord injuries can sever the nerves that carry sensory information to the brain, conscious touch perceptions can be elicited by artificially stimulating S1 [10]. By implanting microelectrode arrays in the somatosensory cortex of a patient with long-term spinal cord damage, Flesher et al. were able to successfully evoke conscious sensations of pressure originating in the hand and found that the intensity of the perceived pressure sensation scales with the magnitude of electrical stimulation. In addition, the patient was able to discriminate between sensations elicited by various portions of the electrode array, allowing the researchers to map stimulation location to the region of conscious perception. Further studies [11] showed that the sensations evoked by stimulation of S1 can also be modulated by the frequency and duration of electrical pulses delivered to the somatosensory cortex. By varying the properties of electrical stimulation, a wide variety of conscious sensations can be artificially generated, suggesting that somatosensory information collected by implantable sensors can be faithfully encoded into the brain using implanted neural stimulators.

B. Limb Reanimation

In addition to recreating somatosensory sensations, implantable devices can also be used to reanimate muscles in paralyzed patients. In the human brain, the motor cortex is primarily responsible for the execution of movements; motor neurons in the spinal cord link this region of the brain to muscle groups throughout the body. Although severe spinal cord injuries can sever these connections, the motor cortex still activates when paralyzed patients attempt to initiate movements. Like the somatosensory cortex, the motor cortex contains a topographic map of body elements. To restore voluntary movement in patients with spinal cord injuries, electrode recording arrays can be implanted in the motor cortex to infer the patient's desire to move certain muscle groups; this information can be used to stimulate peripheral nerves in the body that control limb movement. This technology, known as brain-controlled functional electric stimulation (BC-FES), has allowed patients to regain basic motor functions, such as grasping and moving objects [12,13]. However, without somatosensory signals to inform these motor movements, motion may be unreliable and uncoordinated. By combining a reanimation system with implantable mechanoreceptors, patients may be afforded much more precise control over their natural limbs.

C. Photoplethysmography (PPG) Sensors

PPG sensors primarily consist of a light source pointed towards the skin and a photodetector to capture reflected light. These sensors use the amount of light reflected by the skin to estimate blood flow and supply a reading that is proportional to subcutaneous blood volume [14]. The DC component of these signals is indicative of the total blood volume under the sensor and the AC component can be used to analyze changes in blood flow (e.g., heart rate) [15]. By comparing the absorbance of two different wavelengths of light, the amount of dissolved oxygen in the blood can be calculated; this is known as pulse oximetry. Because of their simple and noninvasive design, PPG sensors are commonly found in the medical setting to measure oxygen saturation, blood pressure, and heart rate. Because external forces can change subdermal blood flow, an implanted PPG sensor may be able to infer the magnitude of applied forces from the measured subcutaneous blood volume.

III. Methods

A. Prototype Construction

To determine if a PPG-based optical sensor would be capable of inferring forces from changes in subdermal blood flow, a large-scale prototype device was first constructed on a protoboard; a schematic and image of the device can be found in Fig. 2. In this prototype, a red LED (SML-P12VTT86R, Rohm Semiconductor) and infrared LED (SFH 4043, OSRAM Opto Semiconductors) with current limiting resistors were mounted 2.5 mm from each side of a reverse-biased photodiode (TEMD7000X01, Vishay Semiconductors). To amplify the optical signal, a 2 MΩ resistor was placed in series with the photodiode. To reduce the output impedance of the circuit, the photodiode output was fed through a unity-gain op amp buffer. An Arduino Mega was used to measure the signal and power the circuit. After verifying the connections, the

circuit was coated with clear epoxy to prevent interference from the environment.

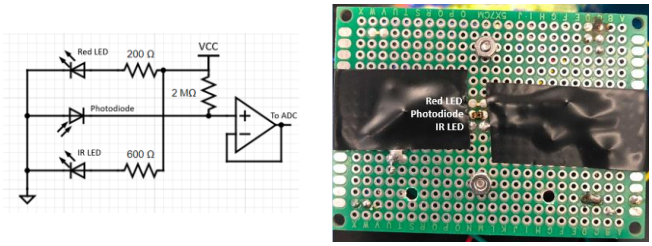

Fig 2. Schematic and layout of the large scale prototype. For testing, the photodiode was centered on the thumb and a normal force was applied to the back of the protoboard using a load cell.

B. Supracutaneous Testing

As a proof-of-concept, the large-scale prototype was mounted on the hand and the photodiode was pressed firmly against the center of the thumb. A load cell was attached to the protoboard to measure the forces applied to the skin. The Arduino Mega was used to switch the LEDs at a rate of 200 Hz while recording the photodiode and load cell outputs. Measurements were also taken with the sensor covered with a thick, uniform layer of epoxy to determine whether the changes in photodiode output are due to altered blood flow or changes in the mechanical structure of the finger.

C. Implantable Sensor Design

Upon confirming the functionality of the optical force sensor, a miniaturized version was fabricated to determine the viability of an implanted sensor. In addition to implementing the core LED-photodiode circuitry, the implantable sensor contains an NFC chip (NT3H1101W0FHKH, NXP USA, 1.60 x 1.60 mm) that enables wireless data transmission and energy harvesting capabilities. Similar to the macroscale prototype, the implantable device probes the skin using light from a red LED (XZMDR155W, SunLED, 0.65 x 0.35mm) and IR LED (SFH 4043, OSRAM Opto Semiconductors, 1.00 x 0.50mm). Current limiting and photodiode amplification circuitry were implemented using five resistors in 01005 packaging (size). The photodiode (TEMD7000X01, Vishay Semiconductors, 2.00 x 1.25mm) signal is fed through a unity-gain op amp buffer circuit (MAX44281OANS+T, Maxim Integrated, 0.84 x 0.84mm) and is read by an ATtiny20 microcontroller (Atmel,1.55 x 1.40 mm) with an inbuilt ADC. The ATtiny controller is also used to switch the two LEDs and send photodiode data to the NFC chip using I^2C. Components were only placed on one side of the PCB to simplify assembly; circuit layout can be seen in Figure 3. The manufactured device occupies a footprint of 5.6 x 6.2 mm including programming pads that can be removed after setup; the necessary electronics occupy a footprint of 3.9 x 5.3 mm.

The ATtiny20 microcontroller was programmed using the AVR assembly language. Upon startup, the on-chip ADC, I^2C communication, and timer interrupts are initialized. The microcontroller continually transmits 10-bit photodiode readings to the NFC chip; these values are updated approximately 200 times per second. LED switching is driven by timer interrupts that fire at a rate of 100Hz.

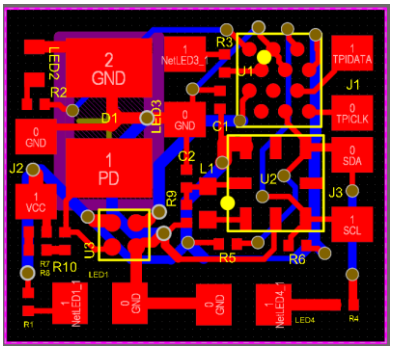

Figure 3. The physical layout of the implantable sensor. The entire board measures 5.6 x 6.2mm, and the necessary electronics occupy an area of 3.9 x 5.3mm.

D. In Vivo Testing

To determine if the miniaturized sensor could successfully infer forces when placed under the skin, an assembled prototype was subcutaneously implanted in the thigh of an anesthetized rat (Fig. 4). Due to manufacturing issues and time constraints, the implanted circuit solely consisted of the photodiode and two LEDs; four wires (red and IR LED anodes, photodiode cathode, and ground) connected the implanted PCB to amplification and control circuitry implemented on a breadboard with an Arduino Mega. A clear plastic case designed to house the PCB was added to prevent external fluids from shorting electrical connections. After successful implantation, a load cell was used to measure forces applied to the thigh while recording the photodiode output for each LED.

Figure 4. In vivo testing setup. The sensor was coated in clear epoxy and inserted into the hindlimb of an anesthetized rat.

IV. RESULTS

As shown in Figure 5, a clear monotonic relationship exists between the photodiode output and applied force for both red and IR light. However, the sensor response for the IR light is more scattered than the response to red light. When a normal force is applied to the fingertip, the measured photodiode reading decreases proportionally for both red and infrared LED light. Figure 6 illustrates the photodiode response to a changing force over time; the sensor reading tracks the applied force quite well.

The *in vivo* test revealed that the signal-to-noise ratio is much lower in the implanted sensor. As shown in Figure 7, the photodiode reading for the implanted sensor is much noisier than the signal produced by the large scale prototype for constant IR light. The peaks in photodiode output seem to correlate with local force maxima, but there does not appear to be a relation between the amplitude of the signal and magnitude of the force. Further inspection of the implanted sensor readout reveals that implanted sensor readout appears to represent the superposition of multiple signals, as shown in Figure 8. After removing outliers and filtering the data using a moving average of width 20 revealed a cyclical pattern in the photodiode data, as shown in Figure 9.

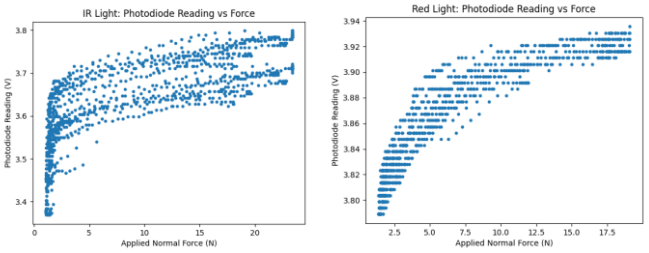

Figure 5. Photodiode output for constant IR (top) and red (bottom) light as a function of applied force for the large-scale prototype. The monotonic relationship between force and sensor output suggests that the sensor can reliably track forces applied to the skin.

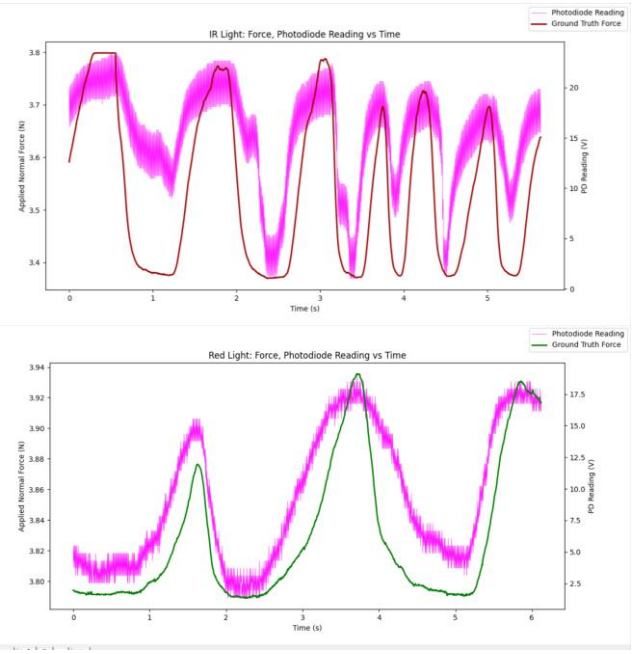

Figure 6. Photodiode output and applied force as a function of time for constant IR (top) and red (bottom) light. The photodiode output tracks the applied force quite well in both cases, although the response to IR light is slightly more distributed.

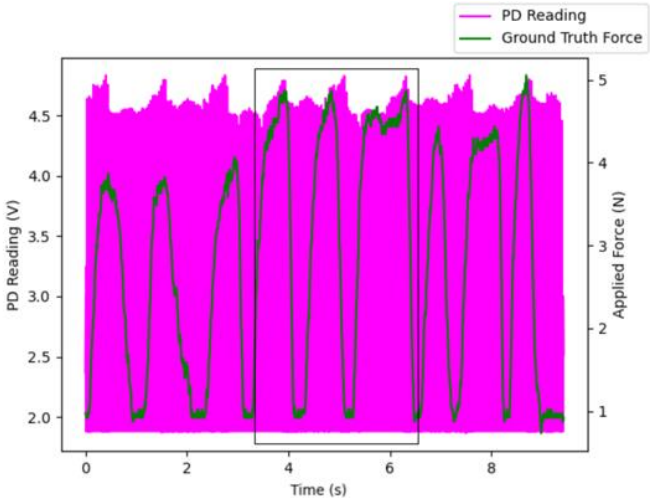

Figure 7. Photodiode reading and applied force as a function of time for the implanted sensor with constant IR light. The peaks of the photodiode output seem to correlate with force maxima. However, the signal is much more noisy than in the large-scale prototype

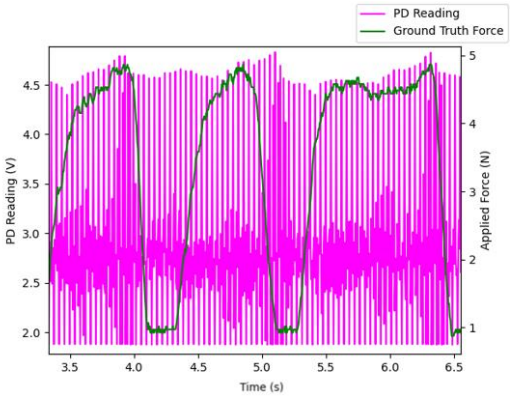

Figure 8. An expanded view of the box drawn in Figure 7 to better illustrate the noise in sensor readings. It appears that at least two signals are superimposed on one another.

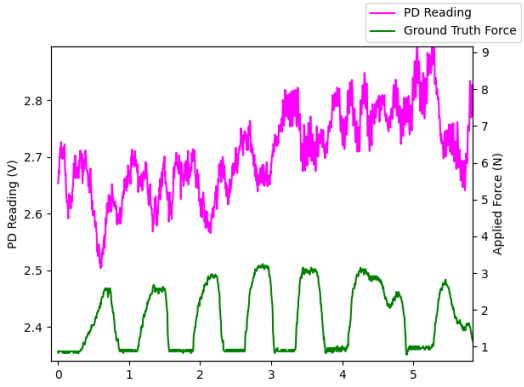

Figure 9. Filtered photodiode signal (pink) overlaid with ground truth force measurements (green); each datapoint in the plotted photodiode signal represents the average of the past 20 original datapoints. Note the change in scale from the previous figure. Cyclic variations in the photodiode signal can be observed in the filtered data.

V. DISCUSSION

These results demonstrate the feasibility of a novel PPG-based optical force sensor. The existence of a monotonic

relationship between the photodiode reading and applied force suggests that forces can be inferred from the subdermal sensor output. The cyclical patterns observed in the filtered implantable sensor data suggest that respiratory and cardiac signals may also be extracted from the sensor readout. Due to time and manufacturing constraints, wireless data transmission and energy harvesting could not be verified with the device, but these capabilities remain feasible with the current system. More advanced signal processing of the implanted sensor output must be performed to determine if a meaningful force measurement can be extracted from the photodiode reading.

Because photoplethysmography sensors measure changes in hemoglobin concentration [9], the observed changes in photodiode output presumably arise from changes in blood flow in the fingertip. When forces are applied to the skin, blood is forced out of the subdermal microvasculature; this can be seen in the blanching of fingertips when grasping objects. When red blood drains from the finger, less red light is reflected by the underlying tissue back into the photodiode. When fewer photons are absorbed by the photodiode, less reverse current is produced and the photodiode reading drops; this corresponds to the measured readings shown in Figure 4. Since the monotonic relationship between applied force and photodiode output was observed when glass or a thick epoxy layer was placed between the sensor and skin, the changes in sensor output can be attributed to altered blood flow in the fingertip rather than changes in the mechanical alignment of the finger.

Although the photodiode signal appears to correlate with applied force in both the implantable and large-scale prototype, the signal is much more noisy in the implantable sensor. Discrepancies in sensor data between the two prototypes may arise from differences between human and rat skin. Human and rat skin blood flow rates differ significantly [16], suggesting that a rat may be a suboptimal animal model for a subcutaneously implanted sensor that infers forces based on blood flow. Furthermore, because the distance between the light sources and photodiode is much smaller in the implanted sensor, the amount of light entering the photodiode directly from the LEDs may be larger in the implantable sensor; this may degrade the signal-to-noise ratio in the implantable device.

The sensor discussed in this paper serves as a good proof-of-concept to show that photoplethysmography can be used to infer forces applied to the skin. Because large implants can irritate patients, the circuit board could be made even smaller by placing on components on both sides to improve viability for implantation in human patients. By placing the small passive components on the reverse side of the PCB, the surface area of the sensor can be significantly reduced with a negligible change in height. However, this would most likely require advanced manufacturing capabilities that were not procured in this experiment due to time constraints. To further minimize surface area, pads required for programming and other excess material can be removed using a laser. Implantation viability may also be increased by using a flexible PCB to improve the form factor of the device; this may reduce the thickness of the sensor and increase patient comfort.

Power consumption is often another crucial metric for implantable devices. Because implanted batteries are often difficult to recharge or replace, many devices have stringent power requirements. Although our implantable sensor relies on wireless energy harvesting rather than batteries for power, minimizing the device's power consumption is necessary to ensure the longevity of batteries in the wearable base unit. Due to the low efficiency of RF power transfer, small changes in the power consumption of the implanted device can have large impacts on the required amount of power radiated from the base unit. Because the power consumption of the optical sensor is dominated by the two LEDs due to their large current consumption, the duty cycle can be significantly reduced to save power; the LED only needs to be turned on when the photodiode output is sampled by the onboard ADC. Interpolation can also be used to decrease the sampling rate

In addition to reducing power consumption and functional surface area, future optical sensors should take advantage of the differing absorption spectra of red and IR light to reduce environmental effects. Because several external factors (exercise, temperature variations, etc.) can affect blood flow to the extremities, force measurements may be more accurate if they rely on the differences between the photodiode response to red and IR light. Many conventional pulse oximeters use the variations in red and IR absorption to accurately calculate oxygen saturation levels. Although the accuracy of these sensors can decrease with temperature [17], the dermis should offer some protection from the environment for our implanted sensor. Further methods to reduce the effects of environmental changes should also be considered.

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