Wearable Conformal Fiber Sensor for High Fidelity Physiological Measurements

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Abstract—Wearable devices are becoming increasingly important, addressing needs in both the fitness and the medical markets. In this article, we describe a novel sensing platform based on a hollow-core polyurethane optical fiber, operating through capillary guidance, that acts as a conformal sensor of pressure or deformation. The novelty is achieved by combining a simple structure (hollow capillary) and a simple detection technique (intensity-based measurement) with unconventional material properties (extreme deformability and high optical absorbance). Used on the wrist and ankle, the sensor allows detailed features of the cardiac pulse wave to be identified with high fidelity, while on the chest it allows the simultaneous measurement of breathing rate and walking cadence. Used together, an array of such sensors (with others) could be incorporated into clothing and provide physiologically rich real-time data for health monitoring.

Index Terms—Capillary fibers, black fibers, optical fiber sensors, polymers, wearable sensors, wearable health monitoring.

I. INTRODUCTION

Wearable medical devices use non-invasive sensors to measure physiological characteristics, including heart rate, oxygen saturation, body temperature, and motion analysis. These devices range from consumer devices such as smart watches, to devices for optimizing sporting performance, to medical devices such as continuous glucose monitoring, fall detection devices, and sleep trackers. Wearables for health monitoring have developed two distinct, though complementary uses: in fitness and wellbeing, and for the management of chronic diseases.

Elite athletes now routinely use wearable sensors to monitor their physiological performance (e.g., devices from Catapult or Zephyr). Millions of consumers use health tracking on devices such as FitBit or Apple Watches. Indeed, the Apple Watch Series 6 was launched with the slogan “the future of health is on your wrist”.

More specialized medical wearables have allowed patients with chronic diseases to better manage their conditions. They have also enabled some powerful emerging trends, such as remote patient monitoring, improved home-based care and telehealth – all of which have accelerated during the COVID-19 pandemic.

This growth will require the development of increasingly sophisticated sensor technology to improve device performance. These improvements will encompass several aspects, including miniaturization, cost, ease of integration with other products, and the quality and type of physiological measurements. Sensors that are integrated into clothing, for example, would be less intrusive and more comfortable to use than standalone devices. In other cases, the challenge is to produce hospital-quality measurements that are reliable across a range of body types and physical activities. As an example, one issue that has been raised for optical sensors that rely on light transmission through the skin is that different results are obtained depending on the skin color [1].

Optical fiber sensors are particularly well suited to wearable applications more generally [2]. They can be used to measure a wide range of physical parameters such as temperature [3], mechanical strain [4], and pressure [5]. They can be made compact and lightweight, have a large bandwidth, and can operate in a wide range of environments [6]. These properties make optical fiber sensor-based devices highly suitable for biomedical applications [7], [8]. Importantly, they can also be incorporated into fabrics.

Indeed, fiber devices have become increasingly sophisticated, including electrical, optical, and mechanical components. These now include fibers that incorporate diodes, microelectromechanical systems (MEMS), memory elements and energy storage [9], [10]. The increasing complexity of these devices has been described as a “Moore’s Law for Fibers” [9].

However, most of these fiber devices and systems are fabricated from materials (such as glass and stiff plastics) that are much more rigid than most biological tissues. This limits their potential for some physiological measurements, such as those based on conformal contact with the skin or measuring small forces. Materials with lower Young’s modulus allow for a greater response to external perturbations and recently a growing number of flexible fibers [11], [12], [13], [14], [15], [16], [17], [18], [19], [20], [21], [22], [23], [24] have been demonstrated.

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for implementation in robotics [11], [12] and wearables [13], [14], [15], [16].

To achieve the low Young’s modulus, various materials have been used, e.g., PDMS [16], hydrogels [17], and various elastomers [11], [12], [18], [19], [20], [21] and diverse fabrication techniques, i.e., molding [11], [12], [16], [17], [20], extrusion [21], spinning polymerization [15], and thermal drawing [18], [19]. The combination of materials and fabrication techniques determined the specific mechanical properties (e.g., Young’s modulus and elastic limit) as well as the maximum achievable fiber length. What all previous reports have in common is the use of solid waveguides where the light propagates in the material itself and the achievable deformation is fully controlled by the material properties.

We recently reported on the fabrication, by fiber drawing, and characterization of hollow-core polyurethane (PU) — a thermoplastic elastomer — optical fibers, which operate by capillary guidance, relying on glancing incidence reflection [22], [23], [24]. Waveguides that confine light into an air core to achieve lower transmission loss date back sixty years [25]. Historically, the complexity of the structures has increased from capillaries to band-gap guiding fibers [26]. However, there has been a recent trend moving towards simpler structures, mostly driven by the advancements on antiresonant fibers [27], all the way back to hollow capillaries [28]. While relatively high loss, such fibers can operate effectively over short distances on the scale of the body and are well suited to integration into clothing. The remarkable material properties of PU allowed high levels of elongation and large deformation (compression and/or bending) that resulted in significant transmission losses [22]. Thus, the sensitive detection of pressure variation or deformation was possible through a simple optical intensity measurement.

The choice of a fiber structure containing a significant air fraction provides a further means to tune the mechanical properties and effectively increase the sensitivity to deformation and in particular to transverse applied pressure. It also adds a degree of freedom to the control of the optical properties.

Hence, the significantly enhanced sensitivity over a greater range of deformations and relative inexpensiveness of hollow-core PU fibers are their key advantages over commercial silica and solid polymer fibers. When compared with electronic transducers and standard capacitive sensors, such fibers offer an additional benefit of noise reduction when measuring pressure due to the smaller area of contact with the measurement site. The significant breaking strain (of up to 600% for thermoplastic PU, TPU) also makes polymer fibers much more rugged than electronics in wearable applications. In this article, we report two examples of the use of these highly flexible hollow-core fibers as respiration and pulse measurement sensors. In both cases the sensing relies on intensity variation of the transmitted light caused by fiber deformation. We show that breathing can be measured with the integration of a fiber at the chest and that the cardiac pulse wave can be measured accurately with a sensor on the wrist and on the ankle. We believe that the high level of fidelity provided by these novel fibers and their inexpensiveness are the key advantages that will ensure their widespread use in next-generation wearable devices and enable them to replace standard optical fiber sensors in the field of heart rate monitoring [29].

II. METHODS

A. Fiber Fabrication

The capillary fibers were fabricated with a heat stretching process similar to that used in the fabrication of microstructured polymer optical fibers [30]. The fabrication consisted of a single step drawing process. The key to successfully using the fiber drawing method with low Young’s modulus materials such as PU is a quasi-zero drawing tension [31]. Two different types of fibers were fabricated: transparent and black fibers. For the transparent fiber, the preform used was a PU tube (FB85-TPU-Clear Grayline LLC) with an outer diameter of 6.375 mm and an inner diameter of 4.78 mm. The fabrication was performed with a set drawing temperature of 240 °C and a feeding velocity of 30 mm/min. The resulting capillary fibers had an outer diameter of 1.5 mm and an inner diameter of 1 mm. For the black fibers, a black PU tube (FB85-TPU-Black Grayline LLC) with an outer diameter of 6.35 mm and an inner diameter of 4.78 mm was used. The fabrication was performed with a set drawing temperature of 195 °C and a feeding velocity of 40 mm/min. The resulting capillary fibers had an outer diameter of 2.5 mm, and an inner diameter of 1.7 mm. Characterization of the stress-strain response, and bend losses of the capillary are reported in the Appendix (Figs. 7 and 8).

B. Breathing Sensor

The sensor is composed of a 20 cm long transparent PU capillary fiber, a 633 nm CW laser diode for the illumination and a Hamamatsu S5972 IR + Visible Light Si PIN photodiode to measure the optical intensity. The laser source, polymer fiber and photodetector were positioned and then fixed on two aluminum holders using commercial cyanoacrylate-based adhesive. The optical setup was then attached to a regular elastic bandage. Data acquisition was performed with an Arduino UNO board connected to a laptop.

C. Pulse Measurement

A 3D printed frame was used to mount the detector and light source. Intermediate conventional polymer fibers couple light between the black PU sensor fiber (2 to 5 cm long) and the light source (650 nm 6 mm 5 mW laser diode) and detector (Hamamatsu S5972 IR + Visible Light Si PIN Photodiode, Through Hole TO-18). The PU fiber extends on the outside of the 3D-printed frame. The device has a screw-adjustable pad in contact with the fiber, allowing for adjustment of the pressure with which the fiber is in contact with the wrist. The power supply for the light source and the detector as well as the data acquisition were external to the wearable device. The signal was collected with a data acquisition card (National Instruments USB-4431) and processed in real time. Pulse waveforms were recorded with the sampling rate of 10 000 samples per second, which ensured that the time resolution was much higher than what was needed to determine any of the characteristic timescales of the system.
The signal to noise ratio of the measurements spanned between 13 and 20 dB depending on the measurement. The system was optimized to maximize the electrical dynamic range, which was noise limited. Filtering was applied, removing frequencies below 0.2 Hz, to remove distortion due to movement, and above 45 Hz, to remove mains noise. The system was stable against temperature and power fluctuations and not affected by modal noise. The drifts were generally slow and therefore resulted in a slowly varying envelope to the signal that could be easily filtered out if necessary (see also Fig. 9 in the Appendix). The high sensitivity of the sensor and good design of the mechanics ensured that the change in amplitude was dominated by the deformation of the fiber rather than the environment.

Measurements were performed on volunteers in age groups spanning between 20 and 60 years old and repeated several times to ensure reproducibility. The testing was performed with prior approval from the University of Sydney Human Research Ethics Committee and with informed consent from all participants, as required by the relevant guidelines and regulations.

III. RESULTS

A. Breathing and Foot Cadence Measured at the Chest

Respiration is a key vital parameter that can be used to monitor and improve athletic performance [32], [33] and medical monitoring. Changes in respiratory rate appear earlier compared to other vital signs such as heart rate and blood pressure [34].

The subject wore a chest strap incorporating an optical fiber sensor [Fig. 1(a)] and the intensity of the light transmitted through the fiber was monitored as a function of time. With the subject stationary, the transmission of the fiber oscillates with the periodicity of breathing, where the signal decreases during inhalation and increases during exhalation [Fig. 1(b)]. To make sure the oscillations are a true representation of the respiration, the subject held their breath and the signal transmitted through the fiber stopped oscillating.

To determine whether other factors such as movement influence the result, the sensor was tested whilst the subject was using a treadmill. Fig. 1(c) shows the results for various running speed settings. The breathing periodicity can be clearly seen for all cases. As the treadmill speed increases, a higher frequency oscillation appears. Analysis of the frequency of these oscillations shows they correspond to the walking or running cadence. For instance, for 7 km/h (1.94 m/s) in Fig. 1(d) the second peak in the frequency spectrum is at 2.43 Hz, which corresponds to the subject with the step length of 0.8 m making 1.94/0.8 = 2.45 steps per second.

B. Pulse Measurement at the Wrist and at the Ankle

A high-fidelity measurement of the pulse over time gives access to substantial information about various health conditions. The desired information can be derived by analysis of the shape of the pulse, using the so-called pulse wave analysis (PWA) [35], [36], [37], [38], [39], [40], [41], [42], [43], [44], and/or by the analysis of the statistical occurrences of the pulses, i.e., the pulse rate variability (PRV) [45], [46], [47].

With such analysis, it is possible to derive information about conditions such as hypertension [38], [39], [44], diabetes [46], [48], cardiac output [43], and mental stress [40].
Fig. 2. Experimental demonstration of PU fiber wearable for measurement of the pulse. (a) Photograph of the wearable on the subject; (b) photograph of the black PU capillary fiber used – scale bar is 2 mm; (c) pulses acquired at wrist; (d) detail of a single pulse at wrist; (e) pulses acquired at ankle; (f) detail of a single pulse at ankle; (g) comparison of single pulse at wrist and ankle showing pulse time difference; (h) measured pulse during a cuff blood pressure measurement; and (i) histogram of pulse-to-pulse times over 20 minutes.

This demonstration addresses the continuous monitoring of the pulse at the wrist and at the ankle. In the implementation at the wrist, the fiber is placed perpendicularly over the radial artery [Fig. 2(a)]. Light is coupled into one end of the fiber, and the changes of the optical power are monitored at the other end with a photodiode. The black PU capillary fiber used is shown in Fig. 2(b). A close-up photograph of the wearable is shown in Fig. 3.

A typical recording of the pulse waveform measured by the wrist-wearable device is shown in Fig. 2(c). Detail of a single pulse is shown in Fig. 2(d). The device provides a high-resolution measurement of the pulse waveform and resolves details with clear correspondence to features of physiological significance, i.e., the foot of the pulse, the systolic and diastolic peaks of the pulse, as annotated in Fig. 2(d). The waveform allows the derivation of critical parameters of the pulse in the time domain and in the frequency domain, as well as relative amplitudes of the various features [35], [36], [37], [38], [39], [40], [41], [42], [43], [44], [45], [46], [47], [48], [49], [50], [51], [52], [53].

We performed a comparative measurement with an additional sensor applied to the ankle, simultaneously with measurement at the wrist. A typical pulse waveform at the ankle is shown in Fig. 2(e), with details of a single pulse shown in Fig. 2(f). Again, key physiological features are well resolved. The pulse shape is different in the two locations, consistent with literature [44].
[49], [50], due to differences in resistance and back reflection. As we were making simultaneous measurements at the wrist and ankle it was straightforward to determine the pulse time difference, as shown in Fig. 2(g). If the two sensors were on the same artery, this time difference would be the pulse transit time. From the pulse time difference (∼92 ms) and the distance between the two points on the body (∼77 cm, in this case calculated as the difference between the paths from the heart to the sensors) a pulse wave velocity equivalent is readily calculated as ∼8.4 m/s, which is a typical value for healthy people [49], [51].

To demonstrate the sensitivity of the response to physiological changes in the actual pulse waveform, a recording was taken while a cuff for blood pressure measurement (Omron BP5100) was inflated and deflated on the upper arm, cutting off and then releasing the blood supply to the wrist, and hence pulse signal to the sensor. The recording is shown in Fig. 2(h) and a clear deformation of the pulses in shape and amplitude is observed. Whilst our purpose here is simply to show that the sensor responds to the physiological change, analogous to stopping breathing in the previous measurement, there are some interesting features in the measured data. For example, at 42 seconds the diastolic peak disappears, while the systolic peak is still visible, before the flow of blood is cut off. Also, as the blood flow returns at 62 seconds for five beats the shape is quite different, resembling the oscillometric oscillations typical of a cuff measurement and related to the Korotkoff sounds [49], [51], [52].

We also looked at one simple, but very significant, parameter that can be obtained from the measurement: the change in time between consecutive pulses, i.e., the pulse rate variability (PRV). In the literature, PRV has been used in mental health assessment studies, in pharmaceutical research, in sleep studies, as well as in cardiovascular health and many more applications [46]. A histogram of the pulse-to-pulse times acquired in 20 minutes of recording at rest is shown in Fig. 2(i) and follows a Gaussian distribution. On average there is a new pulse every 1.05 seconds (about 57 beats per minute), with a standard deviation of 56.8 ms, which agrees with that expected for a short-term PRV in a healthy person – values of standard deviation of heart rate variability for long-term measurement, i.e., 24 h, less than 50 ms are classified as unhealthy [45].

To demonstrate that the pulse traces obtained are consistent with conventional measurements, we simultaneously measured the pulse at the wrist with the PU sensor and with a pulse oximeter (CMS-P PC Based Pulse Oximeter, Contec Medical System Co. Ltd.) on the middle finger of the same hand. The comparison is shown in Fig. 4 where the PU sensor reproduces the pulse shape with high fidelity both in its amplitude and time features, and apparently with higher resolution.

Measurements were performed on multiple subjects of various age groups. The results (Fig. 5) did not differ significantly in terms of the optical signal. However, physiological differences can be observed both in shape and heart rate.

These experiments show the immense potential of the sensing platform based on hollow-core PU fibers for cardiovascular system monitoring. The analysis of high-fidelity blood pressure waveforms obtained with the proposed fiber sensor provides a simple and inexpensive alternative to the existing heart rate monitoring technology based on a variety of much more complex optical fiber sensors [29].

IV. DISCUSSION

The simple capillary guides used in this work are unusual. Capillaries are rarely the waveguide of choice as they are lossy, highly susceptible to perturbations, and difficult to scale to smaller dimensions as the loss increases substantially as the inner diameter reduces. However, the transmission is adequate over the length scales used in this work. If better fiber performance were desired, anti-resonant fibers would improve the transmission and allow for smaller dimensions, lower overall optical loss, and hence reduced power requirements. The sensitivity of the anti-resonant guidance mechanism to the fiber geometry suggests that these structures would also make very sensitive sensors. They have been demonstrated in soft polymers [23], [24], albeit only for THz frequencies. However, it remains a fabrication challenge to produce anti-resonant fibers in PU at small diameters, i.e., for structures operating in the visible range.

Other fiber parameters are more easily changed: the inner and outer diameters and the length of the capillary. The primary sensing mechanism in this sensor is the additional loss caused by a perturbation to the structure through external force such
The sensitivity of the fibers to deformation will also result in unwanted signals that are not physiological in source. The nature of the results we have presented however allows this effect to be mitigated. All the signals we have investigated are periodic, with a characteristic periodicity and a relatively low deviation from it, i.e., narrow bandwidth. Such behavior allows us to filter out other sources of noise. The analysis of the signal in the frequency domain, connected with a large amount of collected data and aided by artificial intelligence, might also allow specific information about non-periodic features to be obtained, such as motion detection or falls.
V. CONCLUSION

The results presented here clearly show that the novel PU fiber sensor can measure physiologically relevant signals from the body. The fidelity of the signals allows for extended and combined analysis of the data according to the different methodologies described in the literature, each of which give insight into aspects of health conditions. One of the biggest advantages of the PU fiber sensors is that they can be multiplexed, by placing multiple sensors at various positions in the body, for example allowing measurement of the pulse shape at various distances from the heart, or the simultaneous measurement of multiple physiological parameters, which can be distinguished in the frequency domain given the different characteristic frequencies. This wearable sensor network or array would allow for example, real-time relationships between activity, cardiac and respiratory performance to be established. We have shown that pulse transit time, and hence pulse wave velocity, can be obtained using this system. This opens up the possibility of a wearable capable of continuous unobtrusive monitoring of blood pressure, as well as pulse shape.

The combination of waveguide materials (low Young’s modulus and color), its geometry (i.e., large air fraction), together with the sensor design (intensity based measurements, wearability and transfer function) allow the realization of unique, continuous, unobtrusive monitoring of vital signs such as breathing and opens possibilities for further applications given the large design parameter space. Moreover, integration of multiple conformable PU fibers sensors along with other wearable devices and analysis methods, such as artificial intelligence, could lead to continuous monitoring of a wide range of human physiological parameters.

APPENDIX

A. Stress-Strain Response of PU Capillaries

As reference for the stretchability of the material used, we report in Fig. 7 the stress-strain curve for a drawn clear PU tube with outer diameter 1.6 mm and inner diameter 1.2 mm. The measurement was performed on 50 mm long samples extended with a 20 mm/min rate on an Instron load frame. The measurement was repeated with 10 samples.

B. Bend Losses of Black TPU Fiber

The optical sensitivity to bending deformation can be measured as the light lost as a function of bending radius. This is one of the possible sensing mechanisms for this fiber. Fig. 8 shows the attenuation vs bend radius of a 2-mm-ID, 30-mm-long black TPU fiber.

C. Stability of Sensing Data Over Time

The comparison of the pulse waveforms at the beginning, in the middle, and after ten minutes of data acquisition (Fig. 9) showed no noticeable drift of the sensing data.

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