Elastic Optical Fibres for Wearable Devices

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Abstract

Wearable devices are becoming increasingly common, addressing needs in both the fitness and the medical markets. This trend has accelerated with the growth in telemedicine, particularly after COVID-19. Optical fibre-based sensors are an ideal candidate technology to be included in such monitoring devices as they can measure a large number of parameters and they are easily integrated in wearables. However, their use has been limited due to their sensitivity being too low compared to the one needed to obtain information from the external perturbations of the human body. This requires highly deformable fibres. In this paper, we describe the use of a hollow polyurethane optical fibre sensor, operating through capillary guidance, to monitor breathing and cardiac activity. These fibres can be conformal to the skin and could potentially be incorporated into clothing. The extreme deformability of the material combined with the hollow nature of the fibre provide the necessary sensitivity to capture the pressure changes associated with breathing and arterial pulses. They allow us to recover and identify features of the cardiac pulse wave with high fidelity. In order to reduce sources of error and to increase the signal to noise ratio, we use black fibres. We discuss some unusual features of these fibre sensors, including the transfer function and how to configure for a monotonic response. Using these unusual fibres, we demonstrate wearable sensors for respiration and cardiac signals that produce physiologically rich data.

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 like smart watches to more specialised devices such as continuous glucose monitoring, fall detection devices, and sleep trackers. These devices allow a range of functionalities. For example, continuous measurement allows the wearer to monitor and hence better manage their health, but it can also be used to help in diagnosis and detection of potentially dangerous anomalies, some of which constitute medical emergencies and require rapid intervention.

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. 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”. The CEO of Apple, Tim Cook was quoted saying “I really believe that if you zoom out to the future and then look back and ask, ‘What has Apple’s greatest contribution been?’ it will be in the health and wellness area.” (ref. 1)

More specialised 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 since COVID-19.

These trends seem set to continue. A recent report (ref. 2) found that US consumers’ use of wearables such as Apple Watch and FitBit had increased from 9% to 33% over a four-year period, while a market report (ref. 3) predicts a compound annual growth rate of 26.8% for medical wearables from 2021 to 2028.

This growth will require the development of increasingly sophisticated sensor technology to improve device performance. These improvements will encompass a number of aspects, including miniaturisation, 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 colour (ref. 4).

Of particular interest, both for potential integration into clothing and for not relying on transmitting signals through the human body, optical fibre sensors are also well suited to wearable applications more generally (ref. 5). They can be used to measure a wide range of physical parameters such as temperature (ref. 6), mechanical strain (ref. 7), and pressure (ref. 8). They can be made compact and lightweight, have a large bandwidth, and can operate in a wide range of environments (ref. 9). These properties make optical fibre sensor-based devices highly suitable for biomedical applications (refs. 10, 11).

To date, optical fibre sensors have almost exclusively been fabricated from glass (typically silica) and stiff plastics (usually PMMA). However, these materials have a relatively high Young’s modulus, typically larger than 1 GPa, which limits their potential for measuring small forces, matching those typically observed in human tissue, and integrating practically into clothing. Materials with lower Young’s allow for a greater response to external perturbations and recently a growing number of flexible fibres (refs. 12-27) have been demonstrated for implementation in robotics (refs. 12, 13) and wearables (refs. 14-17).

To achieve the low Young’s modulus, various materials have been used, e.g., PDMS (ref. 17), hydrogels (ref. 20), and various elastomers (refs. 12, 13, 18, 19, 21, 22) and diverse fabrication techniques, i.e., moulding (refs. 12, 13, 17, 20, 21), extrusion (ref. 22), spinning polymerization (ref. 16), and thermal drawing (refs. 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 fibre length. What all previous reports have in common is the use of solid waveguides, i.e., 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 fibre drawing, and characterization of hollow core polyurethane (PU) — a thermoplastic elastomer — optical fibres, which operate by capillary guidance, relying on glancing incidence reflection (refs. 23-26). While relatively high loss, such fibres can operate effectively over short distances on the scale of the body and are well suited to integration into clothing. The remarkable properties of PU material allowed for high levels of elongation and large deformation that resulted in significant transmission losses (ref. 23, 24). Thus, the sensitive detection of pressure variation or deformation was possible through a simple optical intensity measurement.

The choice of a fibre 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 transversal applied pressure. Moreover, it also adds a degree of freedom to the control of the optical properties.

In this paper, we report two examples of the use of these highly flexible fibres. These examples demonstrate that continuous monitoring of parameters critical for health monitoring can be achieved non-invasively due to the high sensitivity of the fibres, and that sensors requiring fibres of various lengths are achievable with a simple capillary guidance. We show that breathing can be measured with the integration of a fibre at the chest level and that the cardiac pulse wave can be accurately measured with a sensor on the wrist and at the ankle.

Results

Breathing and foot cadence measured at the chest

Respiration is a key vital parameter that can be used to monitor and improve athletic performance (refs. 27, 28) as well as to help prevent deterioration of medical conditions, as changes in respiratory rate appear earlier compared to other vital signs such as heart rate and blood pressure (ref. 29).

The subject wore a chest strap incorporating an optical fibre sensor [Figure 1(a)] and the intensity of the light transmitted through the fibre was monitored as a function of time. With the subject stationary, the transmission of the fibre oscillates with the periodicity of breathing, where the signal decreases during inhalation and increases during exhalation [Figure 1(b)]. This is due to the expansion and contraction of the subjects’ chest, which accordingly deforms the fibre, which due to its elasticity, follows the shape of the chest. To make sure the oscillations are a true representation of the respiration, the subject held their breath and the signal transmitted through the fibre stopped oscillating.
The experiment shows the ability of the sensor to measure breathing by detecting chest movement. However, to determine whether other factors, e.g., movement, influence the result, the sensor was tested whilst the subject was using a treadmill. Figure 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, in Figure 1(d), at 7 km/s, the second peak is at 2.43 Hz, which corresponds to a step length of 0.8 m, appropriate for this subject.

![Figure 1: Experimental demonstration of PU fibre wearable for measurement of respiration: (a) photograph of the wearable on the subject; (b) measurement of respiration in resting condition; (c) measurement at various running speeds (2 km/h – top, 7 km/h – middle, 11 km/h – bottom); (d) analysis of the 7 km/h running trace.](image)

**Pulse measurement measured 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) (refs. 30-39), and/or by the analysis of the statistical occurrences of the pulses, i.e., the pulse rate variability (PRV) (refs. 40-42). With such analysis, it is possible to derive information, among others, about hypertension (refs. 33, 34, 39), diabetes (refs. 41, 43), cardiac output (ref. 38), and mental stress (ref. 35).

This demonstration addresses the continuous monitoring of the pulse at the wrist and at the ankle. In the implementation at the wrist, the fibre is placed crossing the radial artery perpendicularly, light is coupled into one end of the fibre, and the changes of the optical power guided in the fibre are monitored with a photodiode. An image of the wrist-wearable device on the radial artery fixed by a Velcro strap is shown in Figure 2(a), while a picture of the black PU fibre used is shown in Figure 2(b).

A typical recording of the pulse waveform measured by the wrist-wearable device is shown in Figure 2(c). Detail of a single pulse is shown in Figure 2(d). The device provides a high-resolution measurement of the pulse waveform and resolving 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 well as the dicrotic notch, as annotated in Figure 2(d). The waveform obtained 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 (refs. 30-48). The performance of this sensor is further elaborated through the following investigations.

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 Figure 2(e), with details of a single pulse shown in Figure 2(f). Again, key physiological features are well resolved. Furthermore, as expected (ref. 39, 44, 45), the pulse shape is different in the two locations. As we were making simultaneous measurements at the wrist and ankle it was straightforward to determine the pulse transit time, as shown in Figure 2(g). From the pulse transit time (~92 ms) and the distance between the two points on the body (~77 cm) the pulse wave velocity is readily calculated as ~8.4 m/s, which is a typical value for healthy people (ref. 44, 46).

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 Figure 2(h) and a clear deformation of the pulses in shape and amplitude is observed. Whilst our purpose here is simply to show that 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 (ref. 45, 47, 48).

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 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 (ref. 41). A histogram of the pulse-to-pulse times acquired in 20 minutes of recording at rest is shown in Figure 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 is in agreement 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 (ref. 40)].
Discussion

The results presented here clearly show that the novel PU fibre 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 fibre sensors is that they can be multiplexed at various positions in the body, for example allowing measurement of the pulse shape at various distances from the heart. As we have shown pulse transit time, and hence pulse wave velocity, can be readily obtained, which opens up the possibility of a wearable capable of continuous unobtrusive monitoring of blood pressure.

However, the sensitivity of the fibres 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 behaviour 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.

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, for applications on these length scales, these aspects that are normally highly disadvantageous, are of little significance. If better fibre performance were desired, anti-resonant fibres 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 fibre geometry suggests that these structures would also make very sensitive sensors. They have been demonstrated in soft polymers (refs. 25, 26), albeit for THz frequencies. However, it remains a fabrication challenge to produce anti-resonant fibres in PU at small diameters, i.e., for structures operating in the visible range.

Other fibre 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 as bend, twist, pressure, stretch, etc. The sensitivity is thus a function of how readily the capillary is deformed, and this is determined by the Young’s modulus and the inner and outer diameters — a thin-walled capillary will deform much more readily than a thick-walled capillary. Thus, the sensitivity can be engineered over several orders of magnitude by simple design changes, effectively the wall thickness. The fibre length may also be a useful design parameter, depending on whether the force is applied at a point or distributed.

Another unusual aspect of the sensor is its transfer function. Intensity sensors generally show an approximately linear change in intensity over a wide scale of perturbation. However, for these fibres, depending on the sensor design, the range of approximately linear response may be limited, and indeed in some cases it may not be monotonic over the whole range. This can occur because of competing perturbations: compression and bending (both of which reduce the signal) or straightening (which increases it). If the fibre starts from an initially bent position, this can result in perturbations initially increasing the transmitted signal. The wearable device design needs to accommodate these aspects of the sensing fibre.

The typical transfer function of a PU fibre sensor, with an initial small curvature such that on placing on the wrist the central portion of the sensing fibre comes into contact first, is shown in Figure 3(a), together with the measured arterial pulse shapes associated with particular positions on the transfer function. The function has four distinctive regions: the ‘anomalous’ quasi-linear region (1) where the output signal grows with pressure due to the partial straightening of the fibre, the intermediate highly nonlinear region (2) about the function maximum, the ‘normal’ quasi-linear region (3) where pressure reduces the signal due to the fibre squashing, and the saturation region (4) with a strongly suppressed output. Only regions (1) and (3) allow accurate measurement of the arterial pulse waveform [Figures 3(b) and 3(d)] whereas regions (2) and (4) strongly distort the waveform [Figure 3(c)] and degrade the device sensitivity [Figure 3(e)] respectively. Region (2) does not appear useful as the response is not monotonic. For Regions (1) to (3) actual corresponding measurements are shown on the right side of the illustration. It should be noted that the monotonicity of the transfer function over the entire pressure range can be either by design or placement such that compression of the fibre always reduces the output signal, and the output represents the inverted pulse train waveform.
Figure 3: (a) Transfer function of bent TPU fibre and [(b)-{(e)}] effect of its different parts on periodic pulse waveform (f); (b)-(d) right: measured waveforms obtained when the sensor operates in regions (1)-(3) respectively.

An even more unusual aspect of the work relates to the fibre material. Optical fibres are, for obvious reasons, generally fabricated from low loss optical materials. However, the loss mechanism in capillary guidance is dominated by scattering from imperfections (including the perturbations being sensed) on the inner wall, and not by optical absorption of the material. With a capillary made of a transparent material, some fraction of that scattered light ends up being guided within the capillary wall and towards the detector, increasing the overall detected optical power, but reducing the actual signal to noise ratio. Better performance is generally found with black PU as the capillary walls absorb any extraneous light launched or scattered into them, so that the detector is solely illuminated by light that is transmitted through the hollow core.

The combination of waveguide materials (low Young’s modulus and colour), 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 vitals such as breathing and pulse as here reported, and opens possibilities for further applications given the large design parameters space. Moreover, integration with other wearable devices and analysis methods, such as artificial intelligence, could lead to full time monitoring of the human body conditions.

Materials and methods

Fibre fabrication

The capillary fibres were fabricated with a heat stretching process similar to that used in fabrication of microstructured polymer optical fibres (ref. 49). The fabrication consisted of a single step drawing process. The key to successfully using the fibre drawing method with low Young’s modulus materials such as polyurethane is a quasi-zero drawing tension (ref. 50). Two different types of fibres were fabricated: transparent and black fibres. For the transparent fibre, the preform used was a polyurethane tube (FB85-TPU-Clear Grayline LLC) with an outer diameter of 6.375 mm and an inner diameter of 3.175 mm. The fabrication was performed with a set drawing temperature of 240°C and a feeding velocity of 30 mm/min. The resulting capillary fibres had an outer diameter of 1.5 mm and an inner diameter of 1 mm. For the black fibres, a black polyurethane 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 fibres had an outer diameter of 2.5 mm and an inner diameter of 1.7 mm. As reference for the stretchability of the material used, we report in Figure 4 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.

Figure 4: Strain/stress curve of a PU capillary with 1.6 mm outer diameter and 1.2 mm inner diameter. The grey area indicated the maximum discrepancy of the measurements of the various samples.

Breathing sensor

The sensor is composed of a 20 cm long transparent PU capillary fibre, 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 fibre and photodetector were positioned then fixed on two aluminium 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.

Pulse measurement

The device is composed of a 3D printed frame, which hosts the detector and the light source. Intermediate conventional polymer fibres couple light between the black PU sensor fibre (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, Throughole TO-18). The PU fibre extends on the outside of the 3D-printed frame. The device has a screw-adjustable pad in contact with the fibre, allowing for adjustment of the pressure with which the fibre 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. Filtering was applied, removing frequencies below 0.2 Hz, to remove distortion due to movement, and above 45 Hz, to remove mains noise.

Data availability statement

Data will be available from the authors on request.

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Conflict of interest
The authors declare no conflict of interest.

Contributions

AS, ML, SF conceived the idea; AS, IR, AR, designed the experiment, acquired and analysed the data; All the authors contributed to the discussion of the results and preparation of the manuscript.

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