Interferometric Sensors for Application in the Bladder and the Lower Urinary Tract

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ABSTRACT

Improved patient comfort and the need for better quality diagnostic information provide the motivation for new sensor development for the urinary tract. Optical sensors based on single mode fibre optics offer unique advantages in terms of access and miniaturization. We report the design, manufacture and evaluation of a diaphragm based sensor to give better than 10 mbar pressure sensitivity. The diaphragm is formed from a medically compatible material and it's geometric parameters set to give the desired resolution. The rear surface of the diaphragm has a thin aluminum coating such that an interference signal can be detected between the light reflected from the diaphragm and the distal end of the fibre. A number of approaches have been investigated for the analysis of the signal from the sensor using broadband illumination where minimizing overall system cost has been a major driver as well as achieving the required performance. A comparison of the techniques is given and experimental data presented with validation of sensor deflection from a white light interferometer.

Keywords: Interferometry, fiber sensors, extrinsic Fabry-Perot, pressure sensor.

1. INTRODUCTION

Miniaturized pressure sensors are of general interest for in-body use, particularly to complement minimally invasive interventions where the entry point is either along some body duct, such as the urethra or arterial system. The general design requirements are robustness with respect to clinical handling, biocompatibility and sensitivity to absolute pressures in the region of around 1% of ambient atmospheric, i.e. 10 mbar. The particular application addressed here is for sensors which might be used to enhance urodynamic analysis by permitting several point measurements along the urethra with minimal disruption to flow, where an additional design requirement is that the device should eventually be producible in a way that would allow the sensing elements to be disposable.

Bladder dysfunction is a relatively common condition causing troublesome symptoms that often have a significant impact on the quality of life of the sufferer. The symptoms may afflict both sexes of any age. Conventional urodynamic techniques measure the differential pressure between the bladder and rectum either through polymer tubes to external transducers or (less commonly) via miniature transducers [1, 2]. Careful data validation is needed as sampling holes or the sensor may contact the bladder or urethra walls. More fundamentally, the physical size of the probes, between 2 and 3 mm diameter, affects patient physiology during bladder voiding and filling, and the possibility of using arrays for enhanced diagnostic information is very limited.

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Optical fiber based pressure sensors offer a number of advantages over other forms of sensor. Single mode fibers provide a miniaturized length scale working from a basis of the fiber cladding diameter of 125 µm which can be combined with a micro-fabricated sensor body. Single point optical access is achieved by constructing the sensor in a reflection arrangement where a single fiber provides both the input optical wave and the signal wave to be passed to a detector. For high resolution measurements a low finesse Fabry-Perot interferometer arrangement has become popular with various geometries of sensor produced for specific applications [3-5]. In these configurations, a flexible diaphragm is located between the distal end of a fiber and the environment to be measured. A reflective coating on the back of the diaphragm reflects the input wave back towards the fiber, see figure 1. Pressure resolutions to 5 mbar and an extended measurement range, to 5 bar, have been reported using multiple DFB lasers [3]. However, for medical applications the cost of laser based interrogation and analysis systems is prohibitive particularly as a motivation for the current work is in the generation of sensor arrays in order to provide increased diagnostic information that has hitherto been unavailable.

In this paper we describe the design, fabrication and evaluation of a fiber Fabry-Perot pressure sensors using bio-compatible materials to give a pressure resolution of <10 mbar whilst minimizing the cost of the optical system.

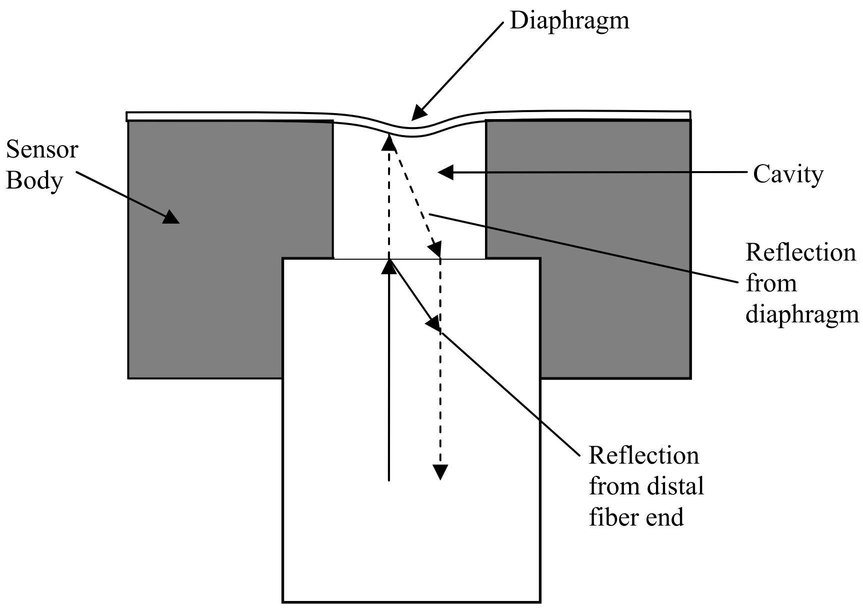

Figure 1 – Schematic of low finesse extrinsic fiber Fabry Perot pressure sensor

2. SENSOR DESIGN

The membrane diaphragm design is aimed at giving good sensitivity (measured as maximum deflection of the membrane) for minimal body diameter. An elastomer has been chosen for the membrane as it gives good deflections (for a given diameter and thickness) and can be fabricated easily and directly onto silicon [6] by spin coating onto the back-side of a wafer followed by etching (e.g. by deep reactive ion etching – DRIE) of a "window" into the silicon. A medical grade RTV silicone (Nusil MED10-6605) was chosen for its relatively low uncured viscosity and the relationship between coating thickness and spin speed established [7]. It was observed that thicknesses down to 2 µm could reliably be produced by controlling spin speed and time, but that there was a tendency for the diaphragms to suffer from discontinuities at the lowest thicknesses. A preliminary parametric study of a series of such membranes of varying diameter and thicknesses of 5 µm and 30 µm [7] has shown that the bending stiffness (measured by central deflection per unit increase in pressure) is a bit less sensitive to diaphragm dimensions than linear elastic, small defection theory would suggest [8]:

$$\frac{y}{p} \propto \frac{d^4}{t^3}, \qquad (1)$$

where y is the central deflection for a differential pressure across the sensor of p, d is the diaphragm diameter and t is the diaphragm thickness. It was not the intention in this work to further develop the mechanics of such diaphragms, but rather to assess the possibility of optically addressing the diaphragm to measure its central defection and to compare this with the technique used in the above parametric study, which was to employ a white-light interferometric profiler, obviously not a suitable method for use in situ. Accordingly, two aspects of the design were fixed, the diaphragm diameter by virtue of available sensor bodies and the diaphragm thicknesses. Both silicon sensor bodies (fabricated as described in [3]) and latterly fiber connector ferrules trimmed to a suitable length between 1 and 2 mm were used. The extra surface area provided at the end face of the ferrule when compared to the silicon sensor bodies was found to be advantageous in forming a seal between the diaphragm and the sensor body. The diaphragm thicknesses used were 20 µm and 40 µm, chosen to give adequate sensitivity in the exemplar sensors, combined with good handling properties for manual assembly. The diaphragms were spin-coated onto copper foil which was subsequently etched using standard PCB techniques to give the desired suspended diameter. The copper foil with adhering silicone was then trimmed, leaving sufficient overlap to bond the copper onto the sensor body. In a final microfabricated design, this step would be obviated by spin-coating onto silicon, etching the silicone and then dicing the wafer to the desired size as indicated above [6]. A thin aluminum coating was applied to the inside of the diaphragm to provide the return signal. The fiber was bonded into the ferrule or silicon sensor body using epoxy whilst observing broadband illuminated fringes live on a spectrometer.

3. OPTICAL READOUT

The system in figure 2 was implemented to interrogate the pressure sensitive optical cavity formed. Light from a 50 W tungsten halogen lamp was coupled into a single mode fiber and split into approximately equal parts at a fused coupler. One output arm goes directly to the pressure sensor with the other arm angle cleaved. Hence the spectrometer (Ocean Optics HR2000-USB, 12 bit intensity, resolution 0.23 nm) detects the interference of the light reflected at the distal end of the fiber and from the back of the diaphragm. A typical interferogram obtained is illustrated in figure 3 and it can be seen that high visibility fringes can be obtained.

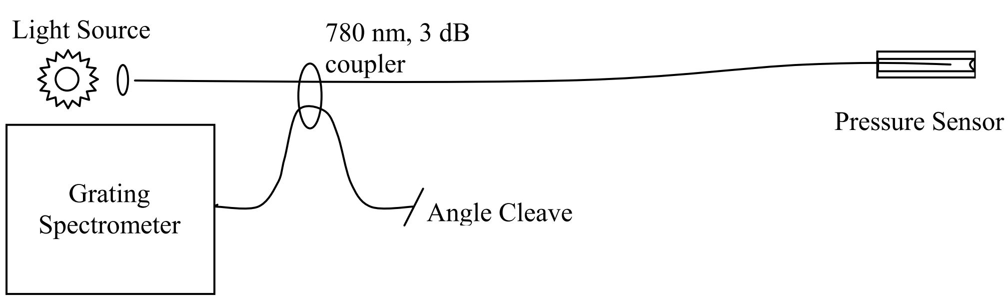

Figure 2 – Single mode fiber pressure sensor interrogation system

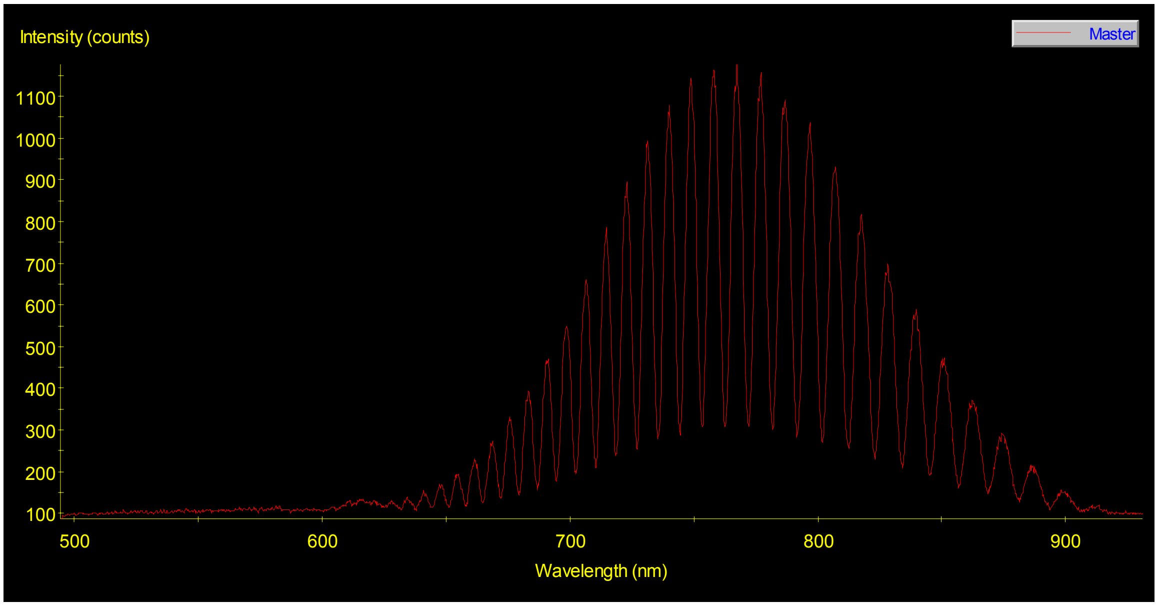

Figure 3 – Example spectrum obtained from a typical sensor

4. SIGNAL PROCESSING

The output of the spectrometer is an array of intensity values that are non-uniformly spaced with respect to the optical wavelength. Three data processing schemes have been compared using a 2-beam model of the interference pattern produced. The interference phase, $\Delta\phi$ is given by:

$$\Delta\phi = \frac{4\pi}{\lambda} l \tag{2}$$

where λ is the optical wavelength and l is the cavity length between the distal end of the fiber and the back of the diaphragm. With an inverse dependence between phase and wavelength it is difficult to apply either Fourier transform or phase evaluation methods. Therefore, the intensity array is re-sampled using spline interpolation to obtain an intensity array as a function of optical frequency in which the fringes are equally spaced. The first analysis approach uses a Fourier transform of the interferogram and performs a Gaussian fit to the peak obtained in the power spectrum. Having identified the dominant fringe spacing the cavity length is determined. Secondly, a phase stepping algorithm has been evaluated that employs 11 intensity samples at 90º steps [9, 10] with the intensities found using an interpolation scheme from the intensity against frequency array. Finally, the Carré algorithm has been used taking 4 intensity samples at equal phase steps [11].

In the simulations an interferogram is generated for a particular centre wavelength and spectral bandwidth of the source (defined by the full width half maximum, FWHM) at the same wavelength resolution as obtained from the spectrometer. A fringe visibility of 0.5 has been assumed. Multiplicative intensity noise has been added based on a statistical measure of the noise typically found from CCD detectors. The cavity length has been calculated using each of the techniques described above over the range 29 to 31 μm in 0.05 μm steps. From the results the standard deviation of the cavity length error has been determined. This process was repeated for source spectral bandwidths between 25 and 200 nm. The results are summarised in figure 4 below. It can be seen that for each technique the expected trend of improved cavity length resolution with increasing source bandwidth is observed. The phase stepping techniques give largely equivalent performance despite the fact that the number of intensity samples used differs by a factor of two. However, significantly better performance is obtained by utilizing the entire intensity spectrum in the Fourier transform approach. The diaphragm geometry, diameter 125 μm, thickness 40 μm, gives a pressure sensitivity of approximately

1.5 nm/mbar hence neglecting other error sources the cavity length resolution achievable can be expected to give the desired pressure resolution of <10 mbar.

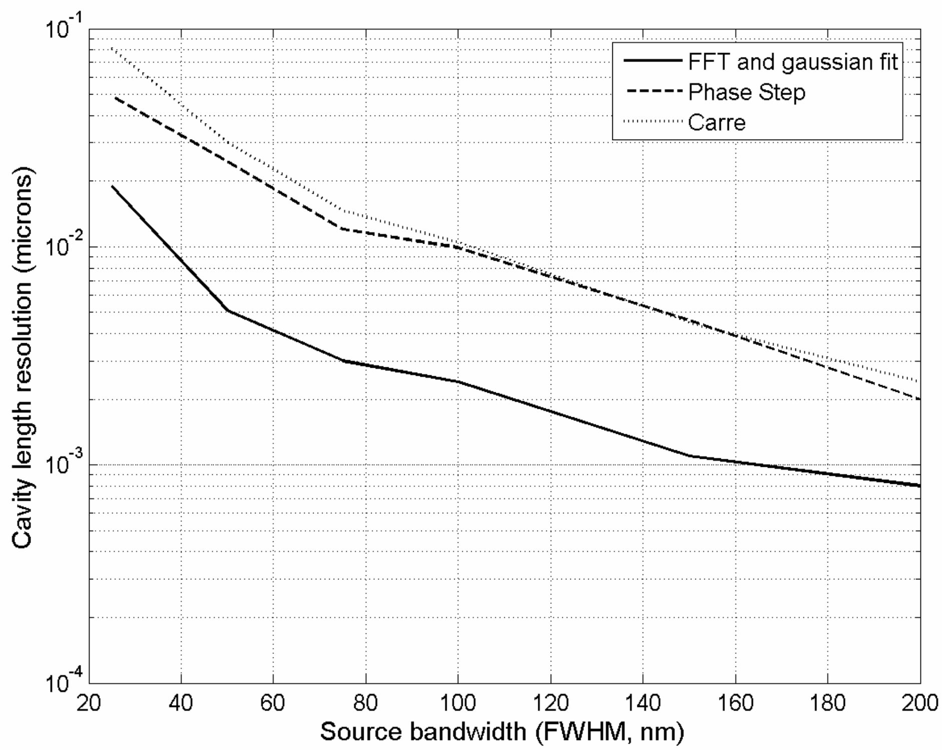

Figure 4 – Diaphragm displacement resolution (1 standard deviation) from simulated data using a number of analysis algorithms and as a function of the spectral bandwidth of the source

5. RESULTS

A detailed understanding of the diaphragm deflection has been obtained by white light interferometry (Zygo). This analysis was performed prior to bonding a fiber into the sensor body such that pressurization could be achieved in a suitable rig. A representative result is given in figure 5a). The expected spatial profile is observed however the data quality is relatively poor despite the addition of an aluminum coating to the exterior of the diaphragm and is believed to originate from the heterogeneous nature of the silicone surface, illustrated from a high magnification image of the surface in figure 5b). The result shown in figure 5a) shows that the deflected form of the sensor extends over a diameter of approximately 215 µm – significantly larger than the 125 µm diameter hole in the ferrule. Hence it can be concluded that the silicone diaphragm is not bonded completely across the end of the ferrule. Analysis of the peak deflection as a function of pressure over a 50 mbar range show that a pressure sensitivity of 13.5 nm/mbar has been obtained.

On bonding the optical fiber into the sensor body a similar series of tests were performed and preliminary cavity length results are illustrated in figure 6. The pressure sensitivity here is 1.55 nm/mbar. The decrease in sensitivity arises as for these experiments a higher pressure was applied to the exterior of the sensor causing the diaphragm to deflect towards the fiber and hence the effective diaphragm diameter is restricted to the hole size in the ferrule of ~125 µm. From equation 1 the difference in pressure sensitivity between the white light interferometry measurements and those from the optical fiber system should scale with d^4, i.e. $215^4/125^4 = 8.75$. This is largely in keeping with the sensitivities obtained in the two sets of experiments. An absolute validation of sensor performance has not been attempted here due to the difficulty in obtaining reliable material properties for particular samples of the silicone used and furthermore these parameters, e.g. elastic modulus, vary significantly with strain and other parameters. The deviation in the measured cavity length from the best fit linear line, see figure 6, is attributable to experimental difficulties and achieving good stability. Even so a resolution, to 1 standard deviation, of 8 mbar has been achieved.

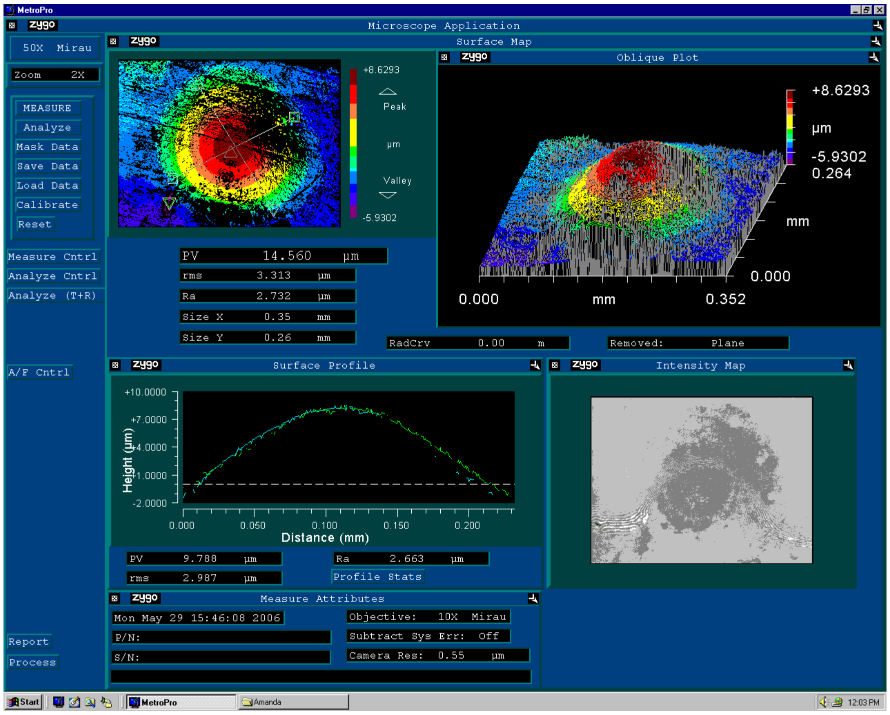

Figure 5a) - White light interferometer deflection profile

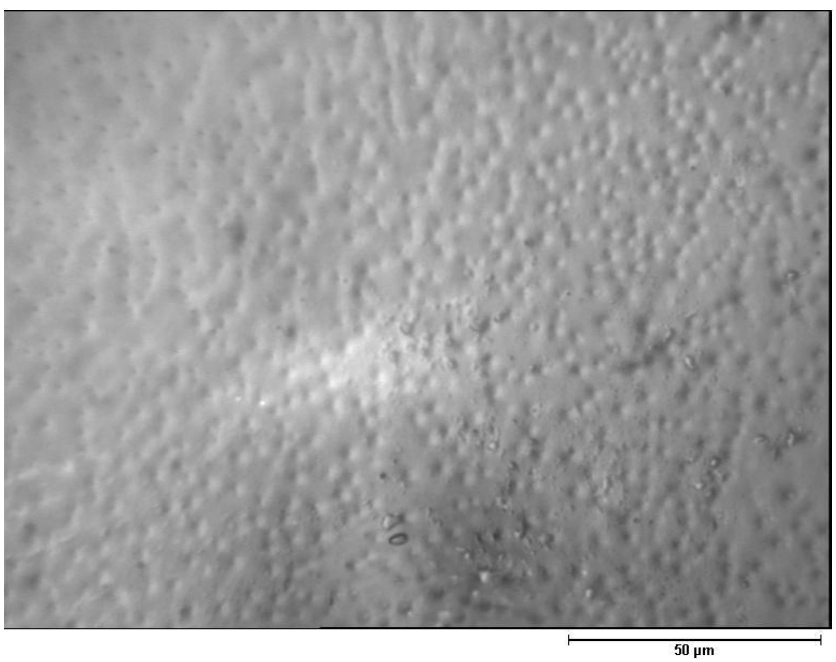

Figure 5b) - Microscopy image of the diaphragm surface

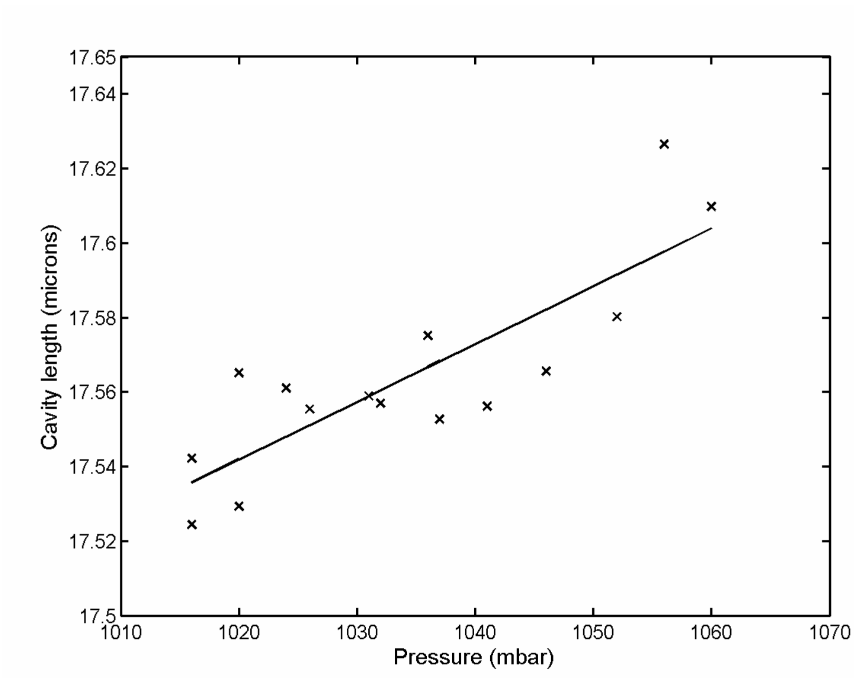

Figure 6 – Measured cavity length as a function of pressure from a sensor with a 125 µm diameter, 40 µm thick silicone diaphragm

6. CONCLUSIONS AND FURTHER WORK

A fiber based Fabry-Perot pressure sensor has been developed for application to urodynamics. It has been shown that bio-compatible silicone diaphragms can be fabricated on suitable dimensions to give a pressure sensitivity of 1.55 nm/mbar. The use of broadband illumination and Fourier based signal processing gives the resolution required in the pressure measurements whilst significantly reducing the cost of the interrogation system compared to laser based approaches. Furthermore, the simulations show that satisfactory resolution can be obtained using a source bandwidth of 50-75 nm (FWHM) thereby demonstrating that multiple sensors, at least 6-8, can be multiplexed to a single spectrometer for simultaneous measurement from a sensor array.

Further work is needed in a number of areas: to investigate the reliability and durability of the bond between the diaphragm and sensor body, to explore the temperature sensitivity of the sensors and investigate sensor performance over a wider range of operating conditions including in situ trials.

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