A Compact Polymer Optical Fibre Ultrasound Detector

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ABSTRACT

Polymer optical fibre (POF) is a relatively new and novel technology that presents an innovative approach for ultrasonic endoscopic applications. Currently, piezo electric transducers are the typical detectors of choice, albeit possessing a limited bandwidth due to their resonant nature and a sensitivity that decreases proportionally to their size. Optical fibres provide immunity from electromagnetic interference and POF in particular boasts more suitable physical characteristics than silica optical fibre. The most important of these are lower acoustic impedance, a reduced Young's Modulus and a higher acoustic sensitivity than single-mode silica fibre at both 1 MHz and 10 MHz. POF therefore offers an interesting alternative to existing technology.

Intrinsic fibre structures such as Bragg gratings and Fabry-Perot cavities may be inscribed into the fibre core using UV lasers. These gratings are a modulation of the refractive index of the fibre core and provide the advantages of high reflectivity, customisable bandwidth and point detection.

We present a compact in fibre ultrasonic point detector based upon a POF Bragg grating (POFBG) sensor. We demonstrate that the detector is capable of leaving a laboratory environment by using connectorised fibre sensors and make a case for endoscopic ultrasonic detection through use of a mounting structure that better mimics the environment of an endoscopic probe. We measure the effects of water immersion upon POFBGs and analyse the ultrasonic response for 1, 5 and 10 MHz.

Keywords: Polymer Optical Fibre Sensors, Fibre Sensors, Polymer Optical Fibre, Endoscopy, Biomedical Applications , Ultrasound

1. INTRODUCTION

Ultrasound has many applications within the domain of medicine, from functional and anatomical diagnoses to therapeutic treatment. One emerging area of research that is gaining greater prominence is the use of opto-acoustics for endoscopic imaging. While ultrasound has relatively poor soft tissue contrast, opto-acoustics delivers relatively speckle free imaging with high contrast and high ultrasonic resolution [1]. Opto-acoustic imaging relies on the absorption of laser pulses, causing the rapid thermos-elastic expansion and contraction of the targeted tissue, enabling the emission of ultrasound to provide functional information on physiological parameters such as haemoglobin concentration. Detectors should be ultra-wideband in the frequency domain, as a typical opto-acoustic spectrum ranges from ~ 100kHz to ~ 50MHz. For the specific application of endoscopic opto-acoustic imaging, the dimensions of the detector is very important, as endoscopic probes are becoming more and more miniaturised while seeking to retain equal or higher sensitivity. Recent publications show endoscopic probe diameters ranging from 2.5mm [2] down to 1.1mm [3], enforcing a sensor of small size that may also be required to function in parallel with other imaging modalities, such as MRI and conventional ultrasound.

The current detectors for opto-acoustics are piezo-electric transducers, which are highly sensitive but have a number of weaknesses. Transducer sensitivity is proportional to its size, they offer a relatively limited bandwidth and are susceptible to electromagnetic interference [4]. These issues are inherent to the nature of piezo electric transducers and render them a sub-optimal solution.

Examining the three issues of high sensitivity with small size, ultra-wide bandwidth and EM immunity; optical detection appears to be a promising proposal. All optical detectors are unaffected by EM interference, they typically provide higher sensitivity than piezo electric transducers without a proportionality to size and there are many potential detection schema that can be attempted to deliver a desired result. These different options include fibre Bragg gratings [5], Fabry-Perot cavities [6], interferometric fibre sensors [4] and micro-ring resonators [7]. Optical fibres in particular provide the

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potential for a customisable bandwidth and have a diameter typically expressed in hundreds of microns. Optical fibres therefore show strong potential for opto-acoustic endoscopic imaging.

Optical fibres can be divided into silica and polymers. Both families of fibre have a range of geometries, materials and dopants suitable for a wide range of potential applications. The most common polymer in use is Poly-Methyl-Methacrylate (PMMA) and a number of studies provide figures from which a valid comparison can be made between Silica and Polymers. Bilro et al gives the Young's Modulus of PMMA as 3.2 GPa compared to 72 GPa for Silica [8] and Peters lists the elastic limits of PMMA and silica as 10% and 3% respectively [9]. Our prior work shows the acoustic sensitivity of a micro-structured polymer optical fibre (mPOF) is more than 9 times higher than a single mode silica counterpart over a 1-10 MHz range [10]. MPOF can deliver endlessly single-mode fibres that are preferable as single-mode POF is not widely available and mPOF presents a degree of geometrical customisation.

While fibre sensors can be of various natures, there are several advantages to the use of intrinsic fibre structures, such as fibre Bragg gratings (FBG) [5]. These structures are inscribed into the core of a given fibre and modulate the refractive index of the fibre section to create a tailored response centred around a central, or Bragg, wavelength. Intrinsic structures have a customisable optical spectral response, can be highly reflective and act as point detectors. The point detection approach allows for multiple structures to be inscribed in the same fibre at different wavelengths and have a small array of sensors within a single fibre. We have previously presented initial ultrasonic measurements using an mPOF sensor that leverages the unique strengths of FBGs [11]. In order to eliminate the facet as a cause of interferometric signals and to demonstrate a compact detector, connectorised sensors are required.

In this contribution, we present a compact in fibre ultrasonic detector based upon a POF Bragg grating (POFBG) sensor. We demonstrate that the detector is capable of leaving a laboratory environment by using connectorised fibre sensors and make a case for endoscopic ultrasonic detection through use of a mounting structure that better mimics the environment of an endoscopic probe. We demonstrate connectorised fibres that can be manufactured with a higher repeatability and lower losses than previously published and measure the effects of water immersion upon POFBGs. Using the ultrasound detector, we measure and analyse the obtained ultrasonic response for incident ultrasonic waves at 1, 5 and 10 MHz.

2. EXPERIMENTAL SETUP AND METHODOLOGY

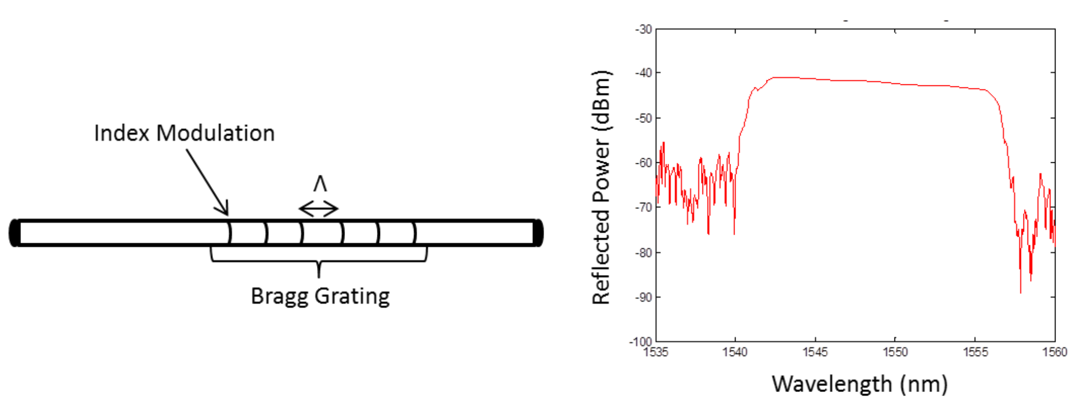

Figure 1. Overview of FBG Sensing and the effect on the transmitted spectrum

An inscribed FBG is a periodic modulation of the refractive index of the fibre core, producing a stop band in the transmission spectrum around a central wavelength, known as the Bragg wavelength. In reflection, it produces a pass band around the Bragg wavelength, as can be seen in figure 1. Parameters such as the grating period, initial refractive index, incident power of the laser light and length can completely alter the profile of the grating to provide a tailored response to the specific application.

When the fibre is exposed to forces such as strain, pressure and temperature, there is a corresponding shift in the refractive index of the fibre, causing a variation in the phase of the FBG profile itself. We tune a laser to the 3 dB point of the FBG profile that generates an amplitude variation at the detector output due to the phase shift of the grating profile. The resultant view from an oscilloscope can be observed as the green modulation in figure 2.

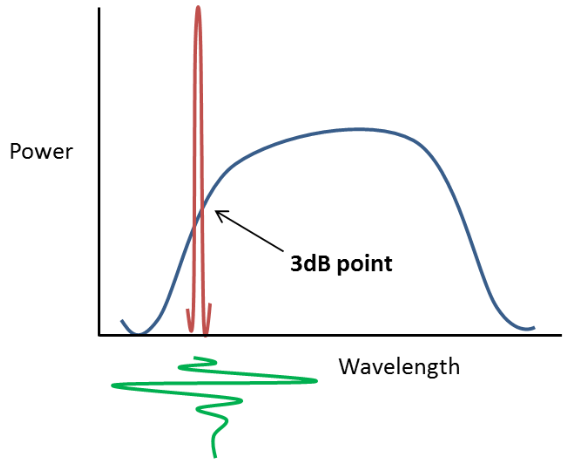

Figure 2. Sketch of a reflected spectrum profile of an FBG (blue), the output of a tuneable laser tuned to the 3 dB point of the grating (red) and the modulation of the output amplitude of the laser as the fibre is affected by an incident ultrasonic wave (green)

First, we examine the grating profile using an Exalos SLED and an Optical Spectrum Analyser. With the peak power and 3dB point found, we determine whether the power and the shape of the grating profile are good enough to provide good quality ultrasonic measurements or not. When we have a grating that we are satisfied with, we replace the SLED with an external cavity tuneable laser (Sacher, Lion TEC-500) and the optical spectrum analyser by a detector connected to an oscilloscope. In this case, a photodiode was used (Thorlabs PDA10A), as shown in figure 3.

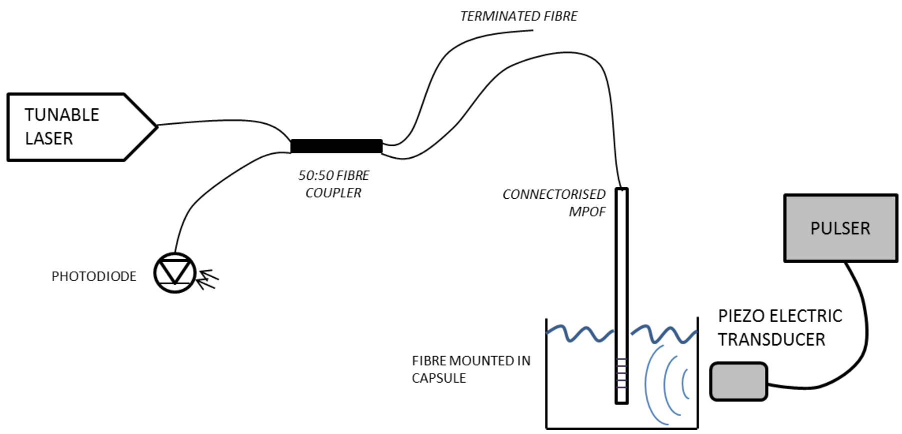

Figure 3. Equipment setup for Ultrasonic detection using intrinsic fibre structures

With the new laser and detector schema in place, we tune the laser to the 3 dB point of the grating and mount the fibre for testing. In order to demonstrate the viability of the mPOF sensors for endoscopic opto-acoustic imaging and avoid artifacts caused by placing a surface behind the fibre, we designed and developed a 3D printable capsule-like structure that serves as both a mount and represents the form of a 5mm endoscopic probe.

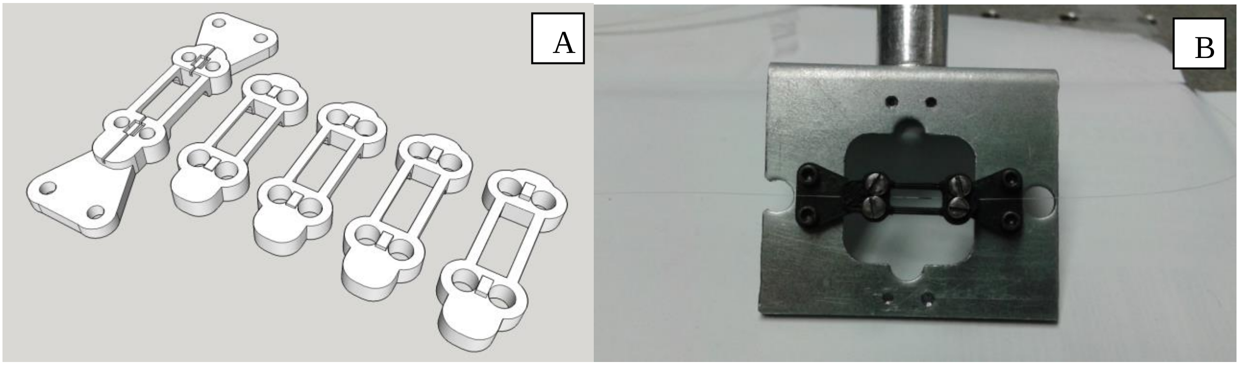

Figure 4. Design of capsule base and top sections with varying depths of grip sections (A) and an FBG secured in the actual mount, ready for testing (B).

The capsule is 21mm in length, 5mm wide and 5mm deep, with a viewing window 8mm long, 5mm wide and 2.4mm in depth. The capsule is mounted to a frame by means of triangular attachments and top and bottom are secured by screw holes in the capsule. These attachment points are for re-usability in a lab environment and have no other function. The base of the capsule has indentations on either side of the viewing window underneath the central fibre groove and the top sections have matching extrusions of variable depths. These allow the modification of pressure on the fibre in order to avoid major profile changes and to keep the fibre in the same position.

Once the fibre is correctly mounted, the structure is immersed in water and any changes are accounted for. Finally, we generate ultrasonic waves onto the fibre by using 1, 5 and 10 MHz ultrasonic transducers (Panametrics V303, V326 and V327) which are excited by a square pulse generator (Panametrics 5077PR).

3. DISCUSSION OF RESULTS

We present a PMMA mPOF fibre sensor based on a Bragg grating for ultrasound detection. The grating is a scanned FBG, 5.7 mm in length and has a bandwidth of 0.09 nm. The results presented in this paper were obtained using a fibre with the profile and reflected power depicted in the below figure. We can observe that the slope from the peak to the 3 dB point is curved at the top and then a near-linear descent on the left side, with a strong reflectivity when observed using an SLED through our experimental setup.

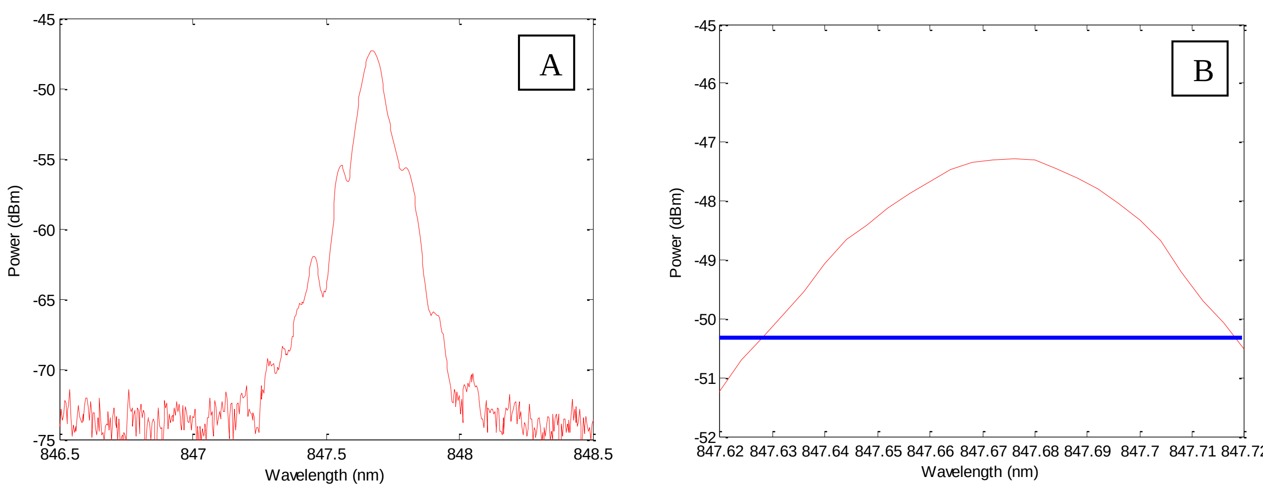

Figure 5. Reflected profile of the FBG (A) and profile showing the peak with the 3DB point demarcated by a blue line (B)

The FBG in question is present in an FC-PC connectorised fibre using a technique that demonstrates typically lower losses and higher repeatability than previously demonstrated [12]. One end of the fibre is etched using acetone to a suitable diameter and then pulled through the connector until the thicker diameter further down the fibre causes it to become stuck inside the connector. The process is monitored through an optical spectrum analyser and when a suitable result is obtained, the fibre is attached using UV glue to hold the fibre in place. The protruding fibre from the connector is then cleaved, yielding a result similar to the one shown in Figure 6. This technique gives an approximate success rate of 50% and typical losses of 2-5 dB.

Figure 6. FC-PC connectorised PMMA mPOF fibre

In order to accurately use a PMMA fibre while immersed, it is necessary to take into account the effect of water absorption on PMMA. Previous studies have shown the effects of humidity upon PMMA [13] but not the effect of water immersion. We provide a first assessment of water ontake through immersion for a PMMA mPOF FBG where the facets of the fibre are kept clear of the water to avoid compromising the integrity of the air holes.

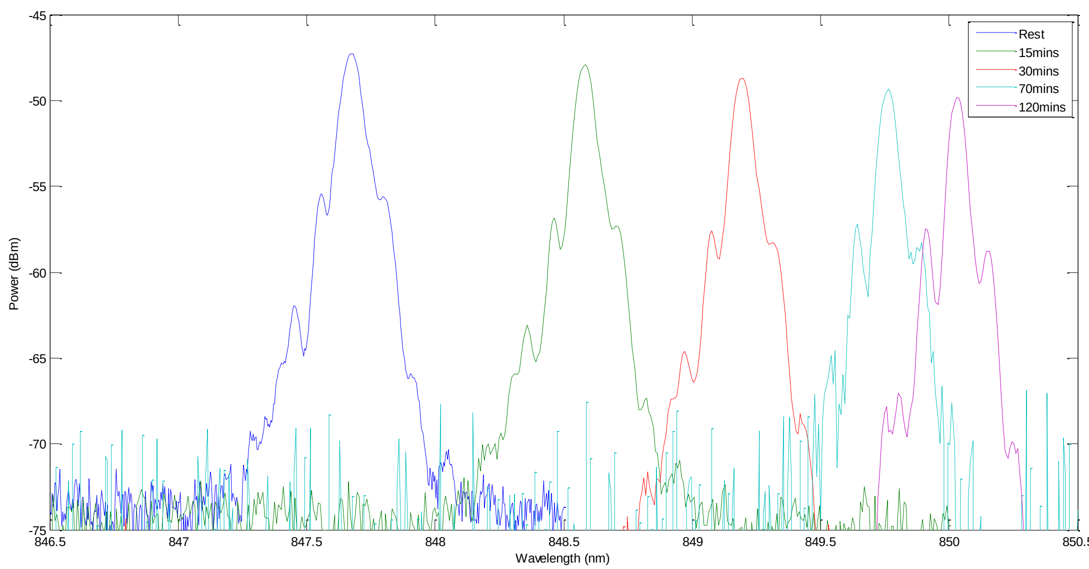

Figure 7. Reflected profile over a period of 2 hours while immersed in water

We show that water ontake due to immersion of the fibre in water at room temperature is a non-linear effect that decreases from 0.16 nm per 10 minutes to 0.016 nm per 10 minutes exposure. After 2 hours, the grating profile in question (PMMA mPOF, 130um fibre diameter) was considered stabilised.

We have obtained an ultrasonic response from immersion transducer induced ultrasound waves using water as the acoustic medium with transducers at 1, 5 and 10 MHz. The detector bandwidth is set to 20 MHz with no filtering or averaging and an output power of -17.6 dBm at the detector output.

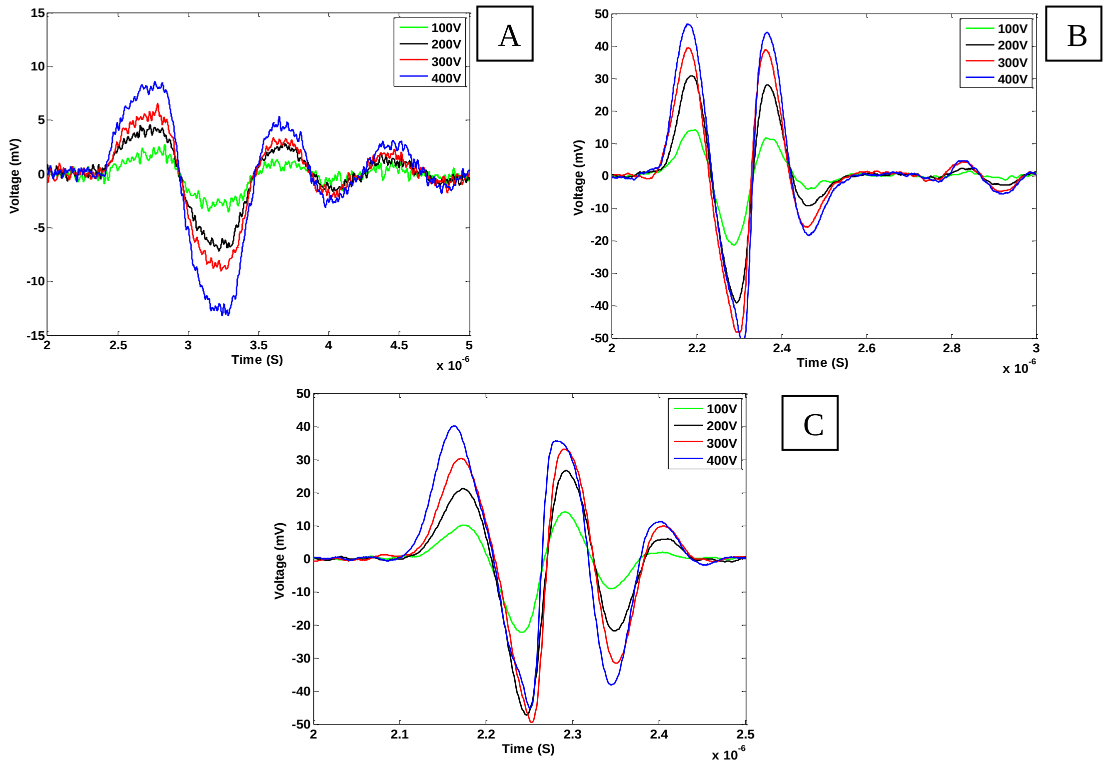

Figure 8. Ultrasonic response for 1MHz (A), 5MHz (B) and 10MHz (C) at excitation voltages between 100 and 400V

We can observe a clear response to the ultrasonic emission that corresponds to our predictions, even at the lowest possible excitation voltage with the least powerful transducer (1 MHz). For our weaker transducer there is notable noise regardless of the excitation voltage in question as the output amplitude approaches the noise level of the photodiode. We can observe that the 5 and 10 MHz responses demonstrate the impact the curved slope of the FBG peak when comparing the 300 V and 400 V responses for each transducer. In each case, we can see a non-linear increase in the output amplitude and indeed in the shape of the resultant pulse. At 2.3 µs with the 10 MHz transducer we are able to see the 400 V response leading the others and with a barely increased output amplitude, caused by the higher pressure applied to the fibre to saturate the detector.

4. CONCLUSIONS AND FUTURE WORK

We have presented a compact in fibre ultrasonic detector based upon a POF Bragg grating (POFBG) sensor. We have shown that the detector is capable of leaving a laboratory environment through connectorisation and have made a case for endoscopic ultrasonic detection through use of a mounting structure that better mimics the environment of an endoscopic probe. We have presented results on the effects of water immersion upon POFBGs that can guide the use of POFBGs and have presented and analysed the ultrasonic response for 1, 5 and 10 MHz.

While we have demonstrated the viability of polymer optical fibres, much more can be achieved by making several modifications to our detector schema. Firstly, modifications can be made to improve the amplitude of the ultrasonic response. We can replace the coupler with a circulator in order to deliver a 5dBm increase, however the cost is prohibitive in the wavelength region in question. We can improve the input power by acquiring a better laser, although the need for tuneability vs. cost must be well balanced. We can also use a detector with a lower NEP and/or a greater gain.

Secondly, while the saturation of the signal could be modified by simply using a less sensitive detector, we must choose between low sensitivity sensors with less saturation or high sensitivity sensors with low saturation thresholds. For opto-acoustics, high sensitivity sensors are preferable, even if this means a lower saturation threshold. Thirdly, we can test other polymers for sensing, such as TOPAS and CYTOP. These polymers benefit from distinct advantages that could also improve the performance of the detector.

Future considerations include a full characterisation of our detector that provides both acoustic sensitivity and directivity data. Opto-acoustic measurements are then the next logical step, along with initial experiments into sensor topology. Then, focus can return to improving the fibre sensor itself through the optimisation of sensor geometry, polymer type, sensor type (FBG, modulated FBG or Fabry Perot cavity). The resulting system would then be worth using for detailed opto-acoustic testing with a range of phantoms and mock in-vivo conditions.

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