PROCEEDINGS OF SPIE

SPIEDigitalLibrary.org/conference-proceedings-of-spie

L-band CYTOP Bragg gratings for ultrasound sensing

Christian Broadway, Kyriacos Kalli, Andreas Theodosiou, Michal Zubel, Kate Sugden, et al.

Christian Broadway, Kyriacos Kalli, Andreas Theodosiou, Michal Zubel, Kate Sugden, Patrice Megret, Christophe Caucheteur, "L-band CYTOP Bragg gratings for ultrasound sensing," Proc. SPIE 10681, Micro-Structured and Specialty Optical Fibres V, 1068109 (9 May 2018); doi: 10.1117/12.2307129

SPIE.

Event: SPIE Photonics Europe, 2018, Strasbourg, France

L-band CYTOP Bragg gratings for ultrasound sensing

Christian Broadway[1], Kyriacos Kalli[2], Andreas Theodosiou[2], Michal Zubel[3], Kate Sugden[3], Patrice Megret[1], Christophe Caucheteur[1]
[1] Electromagnetism and Telecommunication Department, University of Mons
[2] Nanophotonics Research Laboratory, Cyprus University of Technology
[3]Aston Institute of Photonic Technologies, Aston University

ABSTRACT

Polymer optical fibre (POF) has been receiving increasing attention for sensing applications. The fundamental properties of POF such as PMMA deliver at least an order of magnitude in improvements over silica fibres, though practical difficulties create additional complexity. POF has the potential to deliver lower acoustic impedance, a reduced Young's Modulus and a higher acoustic sensitivity within the megahertz region. In contrast, existing piezo-electric transducers have an inherent narrow acoustic bandwidth and a proportionality to size that causes difficulties for applications such as endoscopy within the biomedical domain.

POF generally suffers high attenuation per kilometre at telecommunications wavelengths, limiting fibre lengths to mere centimetres. However, CYTOP, a graded index perfluorinated polymer, is a commercially certified product allowing the use of telecoms region technology and tens of meters of fibre without exceeding acceptable losses. With an effective refractive index between 1.32 and 1.33, it is fundamentally better placed for applications using water or a similar media for acoustic coupling.

We demonstrate ultrasonic detection at 5,10 and 15 MHz using a TFBG within a CYTOP fibre in the telecoms region and the latest knowledge in POF handling and connectorisation. This first step in the use of CYTOP demonstrates the viability of the sensor and paves the way towards further advances towards its eventual application.

Keywords: Polymer optical fibre, fibre sensors, biomedical applications, ultrasound detection, fibre Bragg gratings, CYTOP

1. INTRODUCTION

Ultrasound is widely used for a range of diagnostic applications in the medical domain, from foetal ultrasound imaging to guiding invasive procedures. Ultrasound in this domain is known as biomedical ultrasound, typically encompassed in the 20 KHz – 50 MHz region. For many applications of diagnostic ultrasound, multiple modalities are proposed for use simultaneously, for instance, photoacoustic and ultrasonic tomography within a single endoscopic probe [1].

The industry standard detectors for ultrasonic generation and detection are piezo electric transducers. These detectors are based upon crystals or ceramics and rely on resonance effects, forcing a limitation on frequency bandwidth and linking sensitivity proportionally to the size of the sensing element. Furthermore, they are sensitive to electromagnetic interference, a cause for concern for certain applications.

Optical fibres have the potential to be promising replacements for piezo electric transducers as detectors, delivering immunity to electromagnetic interference and unimpeded by a relationship between sensor size and sensitivity. The broad family of optical fibres offer a wide variety of additional benefits and downsides, offering varying materials, geometry and sensing modalities, which can generally be consolidated into the two sub-categories of polymer and silica fibre. While silica fibre in general suffers lower optical attenuation and is less mechanically robust, polymer fibres exhibit higher optical attenuation but are both more robust and more sensitive to fundamental constraints. As a consequence, polymer fibres are generally used in sensing applications and are most useful where the aforementioned fundamental constraints are to be measured, such as pressure, a valid argument in favour of ultrasound detection.

Micro-Structured and Specialty Optical Fibres V, edited by Kyriacos Kalli, Alexis Mendez, Christian-Alexander Bunge, Proc. of SPIE Vol. 10681, 1068109 · © 2018 SPIE
CCC code: 0277-786X/18/$18 · doi: 10.1117/12.2307129

In recent years, polymer optical fibres (POF) are increasingly used and deployed with progressively better results. In the last decade, polymer fibre has moved beyond the benchmark poly-methyl methacrylate (PMMA) to the point where approximately a half dozen different polymer types are available in various states of development and characterization. Endlessly single mode fibres now exist and fibre connectorisation techniques have been put in place that deliver lower losses. The growing body of progress on polymer is driven by the fundamental characteristics of said fibre, best expressed through the most characterized variety of PMMA. PMMA typically has an elastic limit of 10% and a Young's Modulus of 3.2GPa [2] with specific positive results with relation to acoustic sensitivity [3]. It is important to stress that all polymer fibres have varying characteristics that include both advantages and disadvantages and that for a given application, it is important to consider the entire range of polymers to identify which presents the best proposal overall.

The progress within the domain of POF has led to their testing for a wider range of applications. These range from the fundamentals of temperature [4] and strain [4] to fuel level monitoring [5], dosimetry [6], ultrasound [2] and humidity [7]. To date, both PMMA and TOPAS, a cyclic olefin co-polymer, have been preliminarily examined for ultrasound detection, in interferometric [3] and later fibre Bragg grating (FBG) sensing modalities [8]. Other polymers such as CYTOP [7], polycarbonate [9] and Zeonex [10] have not yet been considered.

For ultrasonic detection, the key parameters for consideration are sensitivity to pressure, a refractive index close to that of water, the capability to inscribe high reflectivity FBGs, a mechanically stable sensor, acceptable optical losses, robust connectorisation results, a compact sensor size (beneficial for certain applications) and a wide bandwidth in the frequency domain. CYTOP, a commercially certified graded index perfluorinated polymer, has the most favourable refractive index and a comparable fibre diameter to other polymer fibres when stripped of its polycarbonate cladding. The response of CYTOP to humidity has yet to be documented, potentially offering a further advantage over the highly sensitive PMMA and offering as a consequence a better option. However, while CYTOP has many potential advantages, it is a fibre that is relatively unknown despite being commercially produced and has difficulties that need resolving before it will be a viable commercial ultrasound sensor. At the present time, CYTOP fibre is exclusively multimode, connectorisation has yet to be demonstrated in publications, has a protective cladding that is more difficult to remove than that of PMMA and identical sensor reproducibility is possibly a concern.

This paper represents the start of work towards the delivery of a CYTOP FBG based ultrasound sensor. We demonstrate 5, 10 and 15 MHz ultrasonic detection using a tilted fibre Bragg grating (TFBG) in reflection within a CYTOP fibre, utilising the latest in connectorisation and stabilisation techniques. Our sensor is the first published polymer fibre ultrasound detector implemented in the L band and demonstrates no technological disadvantage when compared to POF NIR detection. We demonstrate conceptual viability and draw conclusions beneficial to future development.

2. METHODOLOGY, FIBRE PREPARATION AND EXPERIMENTAL SETUP

The ultrasound sensor presented in this paper is based on a CYTOP polymer optical fibre. CYTOP is a commercially certified graded index polymer fibre sold in a small range of diameters by Asahi Glass Corporation. It is clad in polycarbonate to deliver a total diameter of 500 µm. The polycarbonate cladding is both protection and a support, CYTOP itself is prone to naturally rolling itself into loops and the jacket ensures that like other optical fibres, the fibre will remain straight unless acted upon. The sensor in this paper is based on a 120 µm CYTOP core and a 380 µm polycarbonate outer coating.

It is important to consider two primary factors for the eventual sensor, refractive index and optical attenuation. CYTOP has an effective refractive index of approximately 1.33, while polycarbonate has a refractive index in excess of 1.5. The refractive index of polycarbonate brings the combined fibre into the same refractive index range as PMMA and TOPAS. As opposed to other polymer fibres, CYTOP has the lowest optical losses in the L band, with a given attenuation of 0.25 dB/m. Polymer fibres are generally used in the region of 800nm where losses are typically lower so that a given fibre may have a total length of several tens of centimetres. CYTOPs unique properties allow the use of more commonly available L-band equipment along with a maximum length of tens of meters. This length increase allows a wider range of applications and a wider range of sensing modalities, this being one of the primary motivations for long term development of the aforementioned polymer type.

Within the fibre, a tilted fibre Bragg grating (TFBG) was formed at Cyprus University of Technology using plane by plane femtosecond inscription [11]. One premise for using TFBGs as a first step is their ability to couple light into the cladding and back again, potentially rendering ultrasonic detection easier without removing the cladding. It was deemed worthwhile to leave the cladding untouched to determine if the removal of the cladding delivered a significant improvement at a later date, to avoid in depth work with CYTOP prior to establishing functionality and to avoid damaging the gratings through cladding removal. Furthermore, additional work would have required to create a mount maintaining CYTOP flat but without the application of strain and without varying the force of said strain.

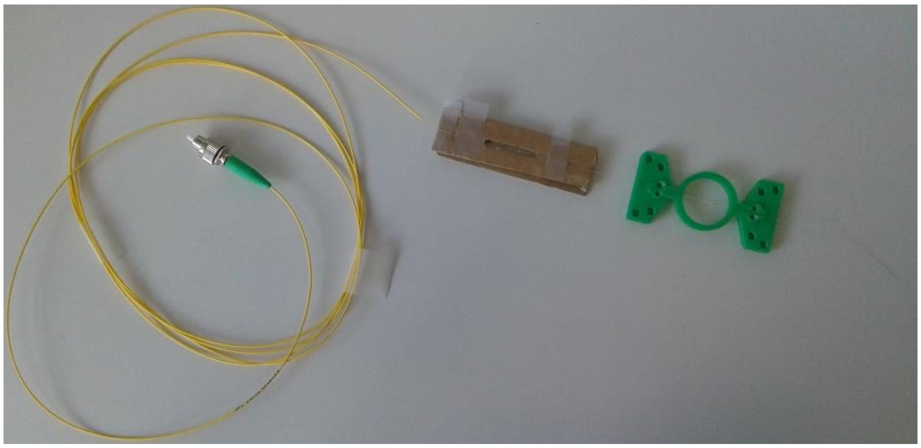

Figure 1. CYTOP TFBG connected to SMF-28 silica fibre using the UV curing method. The joint is secured to a cardboard support and the fibre placed within a custom designed mount

In the first instance, functionality of the TFBG was established using free space coupling techniques. Verifying the strength of the grating using an optical spectrum analyser (OSA), our own connectorisation procedure was developed and carried out to deliver low loss splicing on a level comparable to the latest in PMMA connectorisation (circa 3dB losses per connection). In this procedure, the CYTOP fibre was spliced to standard SMF-28 silica fibre using Norland optical adhesive, which is cured in stages until a stable and robust connection is delivered. The advantages of curing in stages are a lower resting time prior to use and the possibility of re-aligning in between stages to optimise the spectral results. In the case of CYTOP, a stronger adhesive is used to maintain the CYTOP side of the splice from deviation during experimentation, which had in the past led to grating loss and replacement with mode superpositioning. This had the side effect of reducing the number of stages in which optimisation was possible, but also securing the optimised signal faster and allows for a successful connection within 20 minutes at the most. As CYTOP is cleaved at room temperature, this time is not significantly longer than a silica connection, although this method has yet to be adapted for angled connectors.

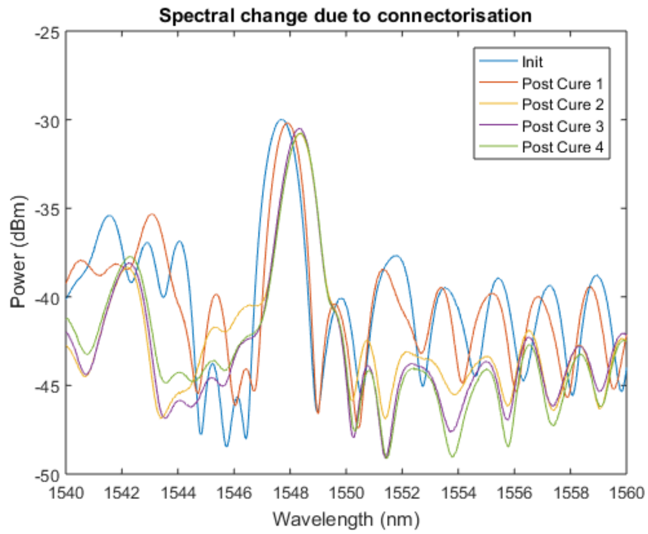

Figure 2 - The evolution of spectral amplitude and central wavelength through connectorisation to SMF-28 fibre using the UV curing method.

As the above figure demonstrates, our connectorisation technique performs well, with losses of 0.7 dBm for the grating profile itself and a noise level reduction in excess of 5 dBm, rendering an initially noisy spectrum closer to the ideal of a single peak and flat baseline. Furthermore, the entire process of connectorisation took 10 minutes, demonstrating connection speeds approaching that of silica fibre. However, it is important to note that due to the curing gel used, leaving the completed connection to rest and finish curing for at least an hour is vital for stability and as such, the gap between silica and polymer connection times is still too large to state that it is a strong competitor at the present time. In order to improve this technique, it will be necessary to transition from UV curing to direct entry into a connector, where the curing gel serves only to hold the fibre in place and the rest time could perhaps be avoided.

Post-connectorisation, it is vital to establish correct mechanical controls so that the impact of curvature can be mitigated so as not to affect the reflected profile of the TFBG. FBGs are point sensors, but using a multimode fibre causes mode super-positioning due to induced curvature. In practice, this can obscure, corrupt and even imitate the profile of an FBG, despite the point sensor nature of an FBG and its theoretical isolation from optical propagation with the fibre. We have designed and 3D printed a mount designed to securely hold a clad CYTOP fibre with coarse pressure adjustment. This mount consists of a base designed to prevent deformation and two top sections that serve to maintain the fibre in position and deliver the controllable pressure. As can be seen in the figure below, with the exclusion of the mounting wings on the base and screw-hole additions, the structure is of a comparable size to a large ultrasonic probe.

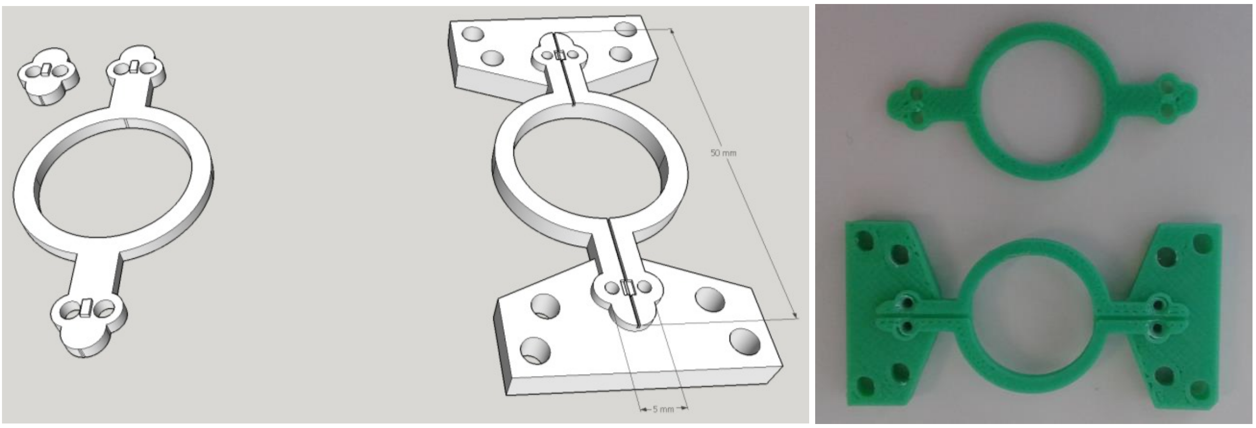

Figure 3 - Mount design (left) and printed example prior to use (right)

By mounting the CYTOP fibre within this structure, ultrasound is only directly targeted at the fibre area where the TFBG is present, with a large degree of tolerance for the angle of incidence, in the event of there being a preference for directionality. This preference was considered likely given the potential for refraction between core and cladding materials and taking into account potential diffraction style effects from the use of a TFBG.

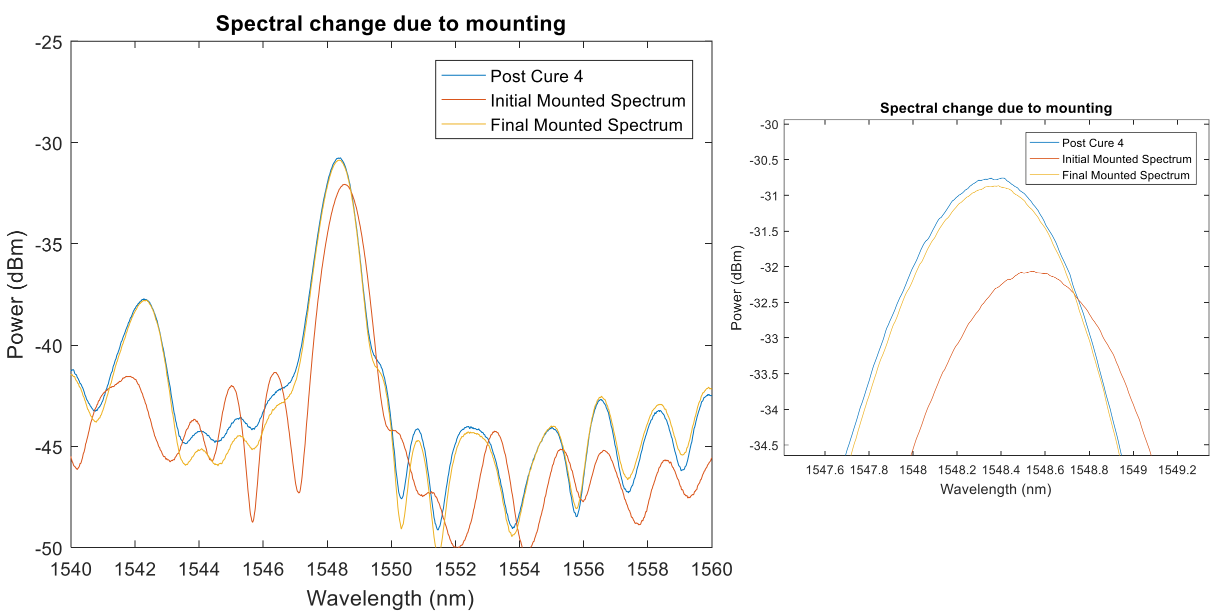

Figure 4 - Spectral changes due to the mounting procedure within the 3D designed structure (left) and a zoom of the TFBG peak (right)

We can observe that upon mounting, the spectral changes are much more significant than the connectorisation losses, with 2 dBm of losses before the applied pressure from the mount is tuned to return the central peak to its pre mount position. We can see that with good fidelity, mounting induced losses are in the order of 0.1 dBm and the pressure induced on the fibre is still enough to maintain it in position when exposed to water and ultrasound. Once the fibre was tested within the mount, an attempt was made to detect incident ultrasound.

The principle of detection is based on the use of the TFBG as an edge filter in reflection. This method effectively tracks changes in the amplitude of a tuneable laser set equidistantly between maxima and minima caused by wavelength shifts that are induced by the incident ultrasonic wave. These amplitude changes effectively trace out the incident signal which can then be subjected to analysis.

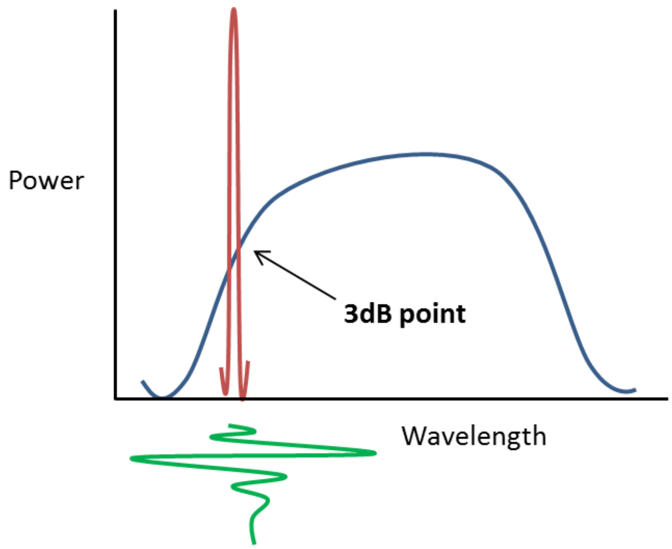

Figure 5 - The spectral profile of the FBG (blue) is stabilised and the 3dB point identified on one of the slopes. The tuneable laser (red) is tuned to this point. Incident ultrasonic waves cause a shift in the wavelength and an oscilloscope output (green) representing the initial wave.

Having connectorised and mounted the CYTOP fibre, the stable spectral profile is measured using an OSA and a broadband light source. A tuneable laser (Agilent 81940A) is then set to the 3dB point of the reflected spectrum and the output amplitude detected by a photodiode (Menlo Systems FPD510). The output signal from the photodiode is displayed on an oscilloscope and should be in good agreement with incident ultrasonic signal. In this experiment, the incident ultrasound is a plane wave from unfocussed immersion transducers and a pulse generator (Olympus, Epoch650, V309, V311 and V319).

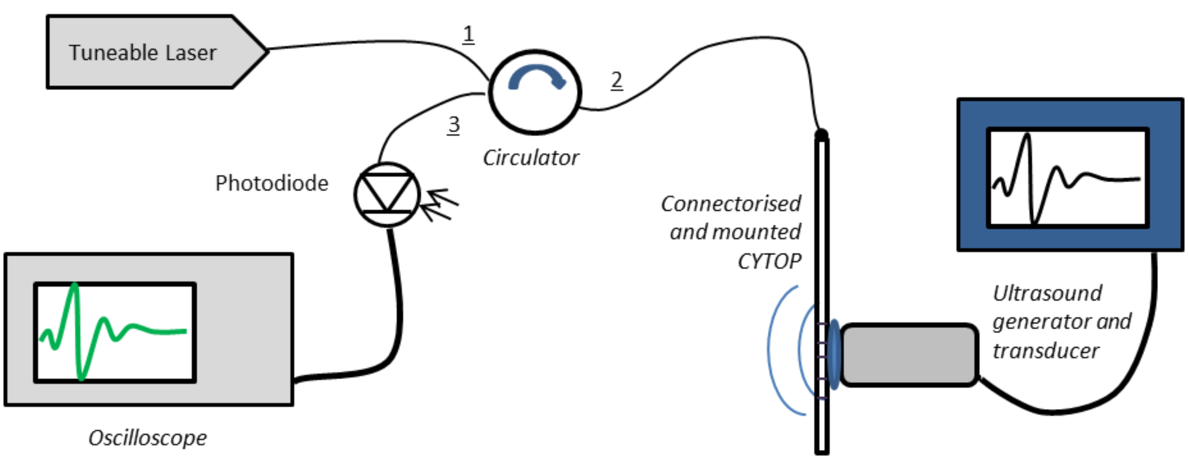

Figure 6 – Ultrasonic detection setup. Incident planar ultrasound is generated by a piezo electric transducer with water based acoustic coupling and detected by optical means through use of edge filtering.

For this experiment, ease of use dictated coupling using a small quantity of water placed on the surface of the transducer and brought into physical contact with the fibre. Physical manipulation confirmed the preference for a certain angle of incidence, which will be of great interest for comparison with an identical experiment without the polycarbonate cladding. Furthermore, these results demonstrate the viability of polycarbonate fibres for ultrasonic detection.

3. ULTRASONIC RESULTS AND ANALYSIS

We present ultrasonic detection using a TFBG within a polycarbonate clad CYTOP fibre. From the aforementioned setup, we obtained three ultrasonic responses with the same incident wave and fibre with an untracked angle variation due to the method used. These responses demonstrate the viability of CYTOP ultrasonic detection and preview the azimuthal directivity of the fibre in a general manner by observing different angles.

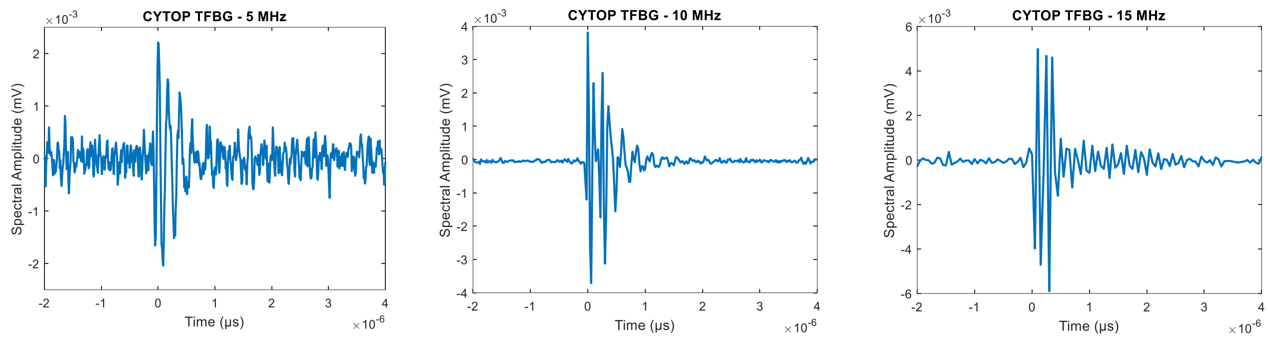

Figure 7 – The detection of incident 5, 10 and 15 MHz centred ultrasound using a CYTOP TFBG.

We can observe a low noise floor that is typically less than 0.5 mV in amplitude and in two of the cases, negligible when compared to the signal strength. This is a significant improvement on past results with PMMA, where ultrasonic signals of less than 5 mV were considered noisy and of poor quality. The overall amplitude of the signals for CYTOP is less than hoped for and many artefacts are present in the signals, indicating acoustic propagation losses. This is primarily caused by the refractive index mismatch from the polycarbonate cladding interface between the CYTOP and the water. In contrast, despite the low signal strength, the 10 and 15 MHz signals show good agreement with the original incident wave shape and the expected output. Increased noise in the 5 MHz signal can be partially attributed to the stronger reverb at lower frequencies and indeed, goes part of the way to explaining the increased noise level relative to the other results.

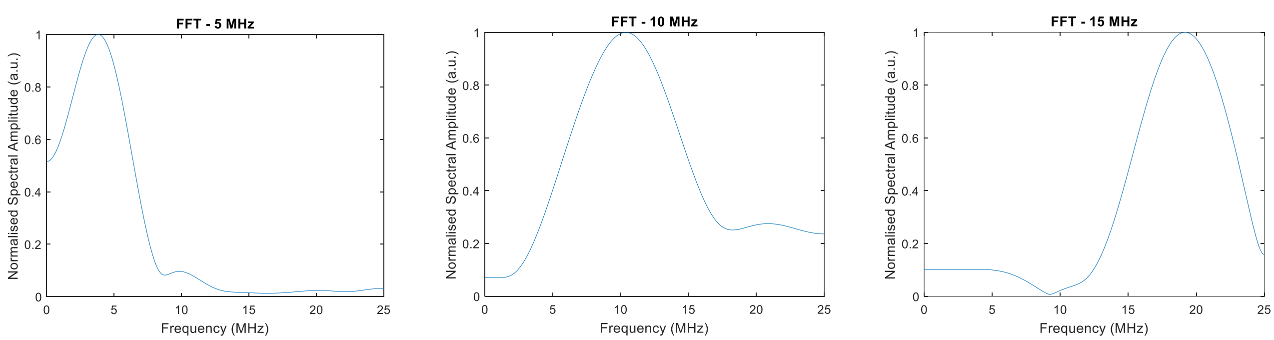

Figure 8 – Fourier transforms yielding the frequency response as detected by the CYTOP TFBG for 5,10 and 15 MHz

The frequency responses as a whole are in line with expectations, with respect to the narrow resonant bandwidth of piezo electric transducers. Indeed, the 5 MHz frequency response conforms well to expectations, while both 10 and 15 MHz transducers are somewhat surprising in that their bandwidth is much wider and in the case of the latter, not centred around 15 MHz. Nonetheless, we can derive from these results that CYTOP is firstly, capable of detecting ultrasound at least up to 20 MHz in frequency and secondly, that the frequency domain results tally closely with the stated output frequencies on the transducers. The primary cause for concern in this case is the larger than anticipated bandwidth, which merits further investigation. This could be due to a number of factors, of which the most likely contenders are the polycarbonate cladding partially diffracting the incident acoustic waves, the low detected signal meaning that a lot of the signal was lost, primarily in the resonant frequency and finally, that the higher frequency transducers are more wideband than in previous reports. For the latter, it is important to note that no characterization details are delivered with the transducers and as such a characterization by a calibration sensor is required.

4. CONCLUSION

This paper has presented ultrasonic detection within the 5-15 MHz frequency band using a TFBG within a CYTOP polymer optical fibre. We have demonstrated low loss connectorisation using the UV curing method and low impact stabilization in a custom built mount. Our sensor is the first published polymer fibre ultrasound detector implemented in the L band and the first CYTOP ultrasound detector in any region. We have detected and analysed the temporal and frequency domain results and drawn conclusions that trace a path for future work. Our analysis concludes that the results are of good quality but poor signal strength, with some aspects of the frequency domain that need greater study. Our sensor appears to function well within the 5- 20 MHz frequency band and as such is setting a positive case for a wideband polymer fibre sensor that can cover the entire biomedical ultrasound frequency band for multiple applications.

In the future, progress needs to focus on three aspects. Firstly, attention needs to be addressed to comparing with unclad CYTOP and removal of the cladding itself. Secondly, connectorisation needs to progress from the UV curing method to direct insertion into a connector, avoiding the many pitfalls of unprotected and flexible connections in a material that can degrade under certain conditions. Finally, the first major milestone is the use of a uniform FBG in unclad CYTOP and comparing it to other fibre types with the same modality. In the same vein, the characteristics of the incident ultrasonic waves need identifying and spectra established to compare with the received signal in greater detail.

5. REFERENCES

[1] W. Wei, X. Li, Q. Zhou, K. Shung,Z. Chen, “Integrated ultrasound and photoacoustic probe for co-registered intravascular imaging”, Journal of Biomedical Optics 16(10) (2011)

[2] C. Broadway, D. Gallego, A. Pospori, M. Zubel, D. J. Webb, K. Sugden, G. Carpintero and H. Lamela, “A compact polymer optical fibre ultrasound detector”, Proc. SPIE 9708, Photons Plus Ultrasound: Imaging and Sensing 2016, 970813 (15 March 2016)

[3] D. Gallego, D. Saez Rodriguez, D. J. Webb, O. Bang and H. Lamela, “Interferometric microstructured polymer optical fiber ultrasound sensor for optoacoustic endoscopic imaging in biomedical applications”, Proc. of SPIE Vol. 9157 (2014)

[4] W. Yuan, L. Khan, D. J. Webb, K. Kalli, H.K. Rasmussen, A. Stefani and O. Bang, “Humidity insensitive TOPAS polymer fiber Bragg grating sensor”, Optics Express/ Vol 19 No.20/ 26 September 2011

[5] C. Marques, G. Peng and D. Webb, “High performance liquid level monitoring system based on polymer fiber Bragg gratings embedded in silicone rubber diaphragms”, Micro-structured and Specialty Optical Fibres IV Proc. of SPIE Vol. 9507 (2015)

[6] A. Fernandez Fernandez, S O’Keeffe, C. Fitzpatrick, B. Brichard, F Berghmans and E. Lewis, “Gamma dosimetry using commercial PMMA optical fibres for nuclear environments”, 17th International Conference on Optical Fibre Sensors Proceedings of SPIE Vol. 5855 (2005)

[7] P. Savva, A. Theodosiou, M. Petrou, K. Kalli, “Concrete RH monitoring using embedded FBG sensors inscribed in multimode CYTOP fibre”, 5th Workshop on Specialty Optical Fiber and Their Applications (WSOF'2017) (2017)

[8] D. Gallego and H. Lamela,” Microstructured polymer optical fiber sensors for optoacoustic endoscopy”, Photons Plus Ultrasound: Imaging and Sensing Vol. 10064 (2017)

[9] A. Fasano, G. Woyessa, P. Stajanca, C. Markos, A. Stefani, K. Nielsen, H. K. Rasmussen, K. Krebber, O. Bang,” Fabrication and characterization of polycarbonate microstructured polymer optical fibers for high-temperature-resistant fiber Bragg grating strain sensors”, Optics Materials Express Vol.6 No.2. (2016)

[10] G. Woyessa, A. Fasano, C. Markos, A. Steffani, H. Rasmussen and O. Bang, “Zeonex microstructured polymer optical fiber: fabrication friendly fibers for high temperature and humidity insensitive Bragg grating sensing”, Optics Materials Express Vol. 7 No. 1 (2017)

[11] A. Theodosiou, A. Lacraz, A. Stassis, C. Koutsides, M. Komodromos, K. Kalli, “Plane-by-Plane Femtosecond Laser Inscription Method for Single-Peak Bragg Gratings in Multimode CYTOP Polymer Optical Fiber”, Journal of Lightwave Technology, vol. 35, no. 24, pp. 5404-5410, Dec.15 (2017)