Background-free fibre optic Brillouin probe for remote mapping of micromechanics

YUCHEN XUANG,1 CARIN BASIRUN,2 JOSHUA CHOU,2 MAJID E. WARKIANI,2 PETER TÖRÖK,1,3 YINGYING WANG,4 SHOUFEI GAO,4 AND IRINA V. KABAKOVA5,*

1Department of Physics, Blackett Laboratory, Imperial College London, Prince Consort Road, London SW7 2BW, UK
2School of Biomedical Engineering, Faculty of Engineering and IT, University of Technology Sydney, Ultimo, NSW 2007, Australia
3Nanyang Technological University, School of Physical and Mathematical Sciences, 637371, Singapore
4Institute of Photonics Technology, Jinan University, Guangzhou 510632, China
5School of Mathematical and Physical Sciences, University of Technology Sydney, Ultimo, NSW 2007, Australia

*irina.kabakova@uts.edu.au

Abstract: Brillouin imaging (BI) has become a valuable tool for micromechanical material characterisation, thanks to extensive progress in instrumentation in the last few decades. This powerful technique is contactless and label-free, thus making it especially suitable for biomedical applications. Nonetheless, to fully harness the non-contact and non-destructive nature of BI, transformational changes in instrumentation are still needed to extend the technology’s utility into the domain of in vivo and in situ operation, which we foresee to be particularly crucial for widespread usage of BI, e.g. in medical diagnostics and pathology screening. This work addresses this challenge by presenting the first demonstration of a fibre-optic Brillouin probe, capable of mapping the micromechanical properties of a tissue-mimicking phantom. This is achieved through combination of miniaturised optical design, advanced hollow-core fibre fabrication and high-resolution 3D printing. Our prototype probe is compact, background-free and possesses the highest collection efficiency to date, thus providing the foundation of a fibre-based Brillouin device for remote, in situ measurements in challenging and otherwise difficult-to-reach environments in biomedical, material science and industrial applications.

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1. Introduction

Brillouin spectroscopy has received much attention in the past decade as a novel non-contact and label-free method for mechanical probing of tissues, cells and biomaterials with cellular and subcellular resolution [1,2]. Unlike contact methods such as elastography, micro-rheology and atomic force microscopy (AFM), Brillouin spectroscopy is minimally invasive since it makes use of light scattering by thermally-induced pressure waves and have minimal effect on tissues and cells at low light powers. The information obtained is characteristic of relaxation behaviour of materials at high frequencies (GHz) and can be related to the material’s compressibility and viscosity [3]. When combined with a standard confocal microscope, it is then possible to map these properties in 3D with optical resolution via scanning. The resultant Brillouin images are therefore hyperspectral and contain richer, more specific information owing to its mechanical contrast [4].

To date, Brillouin imaging (BI) has already been applied to a plethora of areas in life sciences, such as cells [5], human cornea [6], biofilms [7], zebra fish embryos [8], tumour spheroids [3], porcine cartilage [9] and assessment of atherosclerotic plaque formation in mouse arteries [10]. In the last case, a quantifiable difference between healthy and diseased arteries has been
demonstrated on \textit{ex vivo} samples, indicating that BI could guide cardiologists’ medical assessment of the plaque’s vulnerability and may assist in prediction of rupture likelihood. Specifically, instead of the conventional angiographic techniques, which produce low-resolution, non-specific images of coronary arteries [11], the development of a fibre-based diagnostic tool with enhanced imaging capabilities, such as the mechanical specificity and high spatial resolution offered by BI, is envisioned to be a truly disruptive technology. In general, BI is highly attractive for being an all-optical modality, which allows for non-contact mechanical testing, at a distance. Therefore, the possibility of Brillouin \textit{in vivo} and \textit{in situ} measurements has continued to drive the development of a flexible, fibre optic probe useful for scenarios that require remote and versatile operation. Figure 1(A) presents just a few examples of the key application areas where a fibre-optic Brillouin probe can be transformative, where mechanical sensing can be enriched with real-time detection of both spectral and spatial information. Apart from biomedical motivations, on the other side of the spectrum, a robust Brillouin probe would also prove to be invaluable in applications where the remote monitoring of (bio)mechanical properties is essential, such as for quality control and material characterisation in bioprinting and industrial processes [12–16].

![Fig. 1. Schematic design of the Brillouin fibre probe and potential applications. A) The Brillouin fibre probe can be integrated with existing technologies or serve as a novel solution for a wide range of applications. B) Schematics of a flexible, fibre-based Brillouin system. The inset shows a hollow core fibre and a system of miniaturised lenses to enable remote, contact-free measurements.](image)

In our previous work, we explored two fibre probe designs based on step-index fibres that use either bidirectional or separate path geometries for light illumination and collection [17].
Although the bidirectional fibre probe was found to theoretically have an order of magnitude higher light coupling efficiency than the dual-fibre design and, in general, presented a more simple and elegant solution, this concept has suffered a significant drawback. Namely, preliminary results have shown that the background signal, generated by spontaneous Brillouin and Raman scattering in the fibre itself, creates unwanted spectral features far stronger than the Brillouin signal generated by the sample, making this fibre probe design unfit for the purpose of imaging [17]. On the other hand, the dual-fibre design was free from the fibre background and produced good spectroscopic results on standard liquids [17]. The off-axis collection geometry, however, was intrinsically inefficient due to the splitting of optical paths, and hence resulted in large loss in collection efficiency and off-axis aberrations in the probe’s focusing optics. Consequently, neither design could be considered as a working solution applicable for micromechanical mapping of semi-transparent biological materials and tissues. More recently, a time-resolved, fibre-based Brillouin probe was proposed by La Cavera et al. for viscoelastic analysis of liquid samples [18]. The probe functions, however, more akin to a fibre sensor and is not suitable for 3D mapping. Specifically, in order to generate the signal, the probe tip was required to contact the sample to facilitate local heating.

Here we present a non-contact Brillouin fibre probe capable of 3D background-free micromechanical mapping of biologically relevant materials and that utilises a new conceptual design based on a hollow core fibre (HCF). The HCF confines light without the conventional total internal reflection guidance, but rather by utilising anti-resonant guidance in the central, low refractive index hollow-core region surrounded by capillaries in the cladding [19,20]. Since the core region of the HCF consists only of air (Fig. 1(B)), the problem of the fibre scattering background in a bidirectional Brillouin probe is in theory remedied inherently. In addition, we incorporate a new nodeless design that has been recently introduced [21], which has been shown to possess sufficiently low propagation and bending losses (0.2dB/m with bending radius of 8cm at 532nm) that help to achieve high collection efficiency in a flexible, Brillouin fibre imaging system, making the throughput comparable to the free-space optical Brillouin microscopy set-ups [22]. To demonstrate the performance of our HCF-based BI system, the surface micromechanical profile of a tissue phantom (consisting of a hydrogel and water) was measured. This work marks the first key milestone in the new, attractive subfield of flexible and miniature BI systems, with far-reaching implications in a wide range of medical, industrial and bioengineering applications where real-time, in situ micromechanical characterisation is required.

2. Methods

2.1. Miniature fibre probe design

The main considerations governing construction of our fibre probe are summarised in this section with detailed discussion of the overall optical system design presented in Supplement 1. A miniaturised objective has been designed for the dual purpose of the sample illumination and the back-scattered light collection at the sample-facing side of the set-up. The overall dimensions of the miniaturised optics and housing components were designed specifically to be compatible with endoscopic devices, namely to be less then 3 mm in diameter. Another important consideration for the mini-objective was to match the low numerical aperture (NA) of our HCF (NA~ 0.02). Finally, the optics was required to have a sufficient working distance of greater than 1 mm to enable non-contact probing of the sample. These considerations resulted in the final design, consisting of two GRIN lenses positioned in series (G2P10 and GRIN2906, Thorlabs). The arrangement of the lenses were optimised using Zemax OpticStudio (see Supplement 1 for details) as schematically presented in Fig. 2(A).

At the sample plane, this mini-objective lens thus focuses to a diffraction limited spot radius of 5.081 µm, with NA=0.08 and working distance of 1.04 mm. The focal plane is conjugated with the entrance of the HCF, which acts as a pinhole with a diameter of ~20 µm. The simulated
back-coupling efficiency is maximised in this arrangement to be 81.0%, after taking into account the systematic reflection loss.

In summary, our new miniaturised fibre probe design achieves not only higher spatial resolution but also at least an order of magnitude higher collection efficiency than what was possible with our dual-fibre design in the past [17]. The tolerance of the design to misalignment and fabrication imperfections was verified by performing Monte Carlo simulations (Supplement 1) of 20 different configurations within defined manufacturing ranges (axial misalignment ≤ 0.5 mm, relative tilt ≤ 1°). The focal spot was still found to be diffraction-limited on average, while the worst-case scenario produced an RMS radius of 6.98 µm. In addition, these large tolerances only produced a minor decrease of 8% in fibre coupling efficiency when compared to the ideal case. To fix the relative alignment of the distal optics in place, a custom-designed sleeve was 3D printed in-house using high precision stereolithography (Fig. 2(B)). The tolerance of the fabrication process allowed for an alignment accuracy of ±15 µm, which was well within the predicted fibre probe design tolerances.

2.2. Fibre-based Brillouin imaging set-up

Figure 3(A) illustrates the complete set-up for HCF-based Brillouin imaging. In brief, the laser (Torus 660 nm, Laser Quantum) produces a collimated beam with a diameter of 1.66 mm and maximum power of 120 mW. A linear polariser (LP) is placed directly after the laser input to ensure a linear polarisation state, which is crucial to facilitate polarisation control in the system later on. A non-polarising beam-splitter (BS) with a split ratio of 90:10, for transmission and reflection respectively, then splits the light into a signal and reference arms and directs the latter into the spectrometer (TFP1) for alignment. The spectrometer used in this work is a tandem Fabry-Perot (FP) interferometer (TFP-1) supplied by JRS Scientific Instruments.

The beam transmitted by the non-polarising BS is then expanded to twice of its diameter with the two lenses as described in Supplement 1. A polarising beam splitter (PBS), whose transmission axis is aligned with the orientation of the LP, is expected to transmit ≥ 90.0% of the incident light. A quarter waveplate (λ/4) placed at 45 degrees with respect to LP then converts
Fig. 3. Experimental configuration of the fibre-based Brillouin set-up. A) The optical set-up schematics (not to scale). B) The distal probe assembly. C) The beam profile and scanning electron microscopy image of the HCF.

the incident linear polarisation state to circularly polarised state. A single achromatic doublet (f=75mm, AC254-075-A-ML, Thorlabs) is then used to couple light to the hollow core fibre which was in-house fabricated to produce a mode field diameter (MFD) of approximately 20 µm and NA=0.02 at 660 nm. The properties and performance of the fibre are comparable to that described previously [24]. The fibre structure and beam profile after the fibre are shown in Fig. 3(C).

The entire length of approximately 3 metres was used initially to ensure single mode operation, the fibre was stripped and cleaved at 90° at both ends and carefully mounted onto two 3-axis flexure stages (MAX313D, Thorlabs) for fine alignment. The input fibre coupling efficiency was measured to be >60%. The distal probe assembly consists of two GRIN lenses (G2P10, GRIN2906, Thorlabs), which were inserted into the transparent, 3D printed sleeve (Microfluidics BV-003) for mounting. A photograph of the assembled probe is presented in Fig. 3(B). The same optics serves both to illuminate and collect back-scattered light from the sample, which is back-coupled into the HCF (collection efficiency approx. 70%) and sent back along the same light path. Once the scattered light arrives at the waveplate, it is rotated back into a linear polarisation state, but at a direction orthogonal to LP, which causes the PBS to direct the back-scattered light towards the spectrometer arm, where the large doublet lens (f=300mm, AC508-300-AB-ML, Thorlabs) finally focuses it into the spectrometer for analysis.

The TFP1 system is equipped with an adjustable entrance aperture, ranging from 0.07 mm to 1.3 mm in size. Just as for a grating spectrometer, in a FP system, both the spectral resolution and the throughput are also influenced by the shape and size of the input aperture [23]. In the interest of maximising throughput and hence SNR of the device, the largest aperture size (1.3 mm) is to be used for all experiments and the spectral response of the system can be characterised by acquiring the spectrum of a known substance, namely distilled water at room temperature (see Section 2.4).

2.3. Sample preparation

The spectroscopy samples used in this study are either pure, distilled water or hydrogels prepared in-house. The former can be easily deposited in the same V-groove mount that holds the probe assembly on the flexure stage, where surface tension alone is sufficient to hold the droplet in
place. The latter is the GelMA (0.25% LAP) hydrogel, printed using a BioX bioprinter from Cellink, using the droplet protocol available. The best printing parameters were determined to be at 28°C and a pressure of 30 kPa. The extrusion time was adjusted to be 1 second to account for the sample size. The droplet sample was crosslinked by exposure to UV light (405 nm UV module) available on the BioX and it was exposed for approximately 10 minutes at a distance of 15 cm from the sample. The final mixture produced a hydrogel with approximately 10% in solid fraction.

For spatial scanning, a combination of the two materials was used to create a ‘phantom’ with spatially contrasting mechanical properties. A droplet of the GelMA hydrogel, roughly 5 mm in diameter was first prepared on a microscopy slide as described above. Next, a multilayer tape with a square window for the droplet was placed on the slide. A small amount of distilled water was then dropped with a pipette within the window area to immerse and hydrate the gel. A 0.14 mm microscopy cover slip was finally placed on top to seal the structure and prevent water evaporation. Special care was also taken during this process to remove any air bubbles within the water-hydrogel construct.

2.4. Spectral response deconvolution

In order to retrieve biologically relevant linewidth information from the obtained spectra, the effect of the optical system needs to be taken into account. Following the theoretical framework proposed by Török and Foreman [25], the detected spectral intensity (D) can be considered as the convolution of the true signal (S) with the response function of the instrument (R), which is based on the assumption of a linear system.

\[ D = S \otimes R \]  

As discussed above, the spectrometer used in this work is commercial and does not exhibit signs of large optical aberrations, while the official specification also states that the instrumental response function attains a contrast of \(10^{10}\). This combined with the fact that common Brillouin spectra obtained using the instrumentation follow a Lorentzian lineshape almost perfectly, allows the further assumption of a dispersion-limited system, which implies that the Eq. (1) can be simplified to the convolution of two Lorentzian functions. Additionally, the spectral region-of-interest (ROI) selected in this work is located towards the centre of the detector and is described by enough pixels for the resultant spectrum to be considered as finely pixelated. Hence by assuming a practically infinite detector and negligible pixelation effect, the linewidth of the true signal can be determined by first calibrating the instrumental response function. This was accomplished by measuring a distilled water sample, whose viscoelastic properties are well-documented [26] and can be used to generate a theoretical spectrum (\(S_w\)) by relating these quantities to the phonon relaxation time [27], thus the response function of the system is given by:

\[ \bar{R} = \frac{\bar{D}_w}{\bar{S}_w} \]  

where the deconvolution process has been performed on the detected water signal (\(D_w\)) in Fourier space, the \(\bar{X}\) notation is thus used here to signify the Fourier transform operation, which is implemented numerically with the use of Fast Fourier Transform (fft) in Matlab. To cater for experimental variabilities, the simulated and experimentally obtained signals of water are first peak-matched and normalised. Subsequent measurements of the linewidth of any unknown substance then requires the deconvolution of this response function. Mathematically, since the convolution of two Lorentzian functions also gives another Lorentzian function whose linewidth is given by the sum of the two constituent linewidths, in practice, the process to obtain the true
linewidth ($\Gamma_T$) is simplified to a subtraction:

$$
\Gamma_T = \Gamma_D - \Gamma_R
$$

where $\Gamma_D$, $\Gamma_R$ are the detected and response function linewidths respectively.

### 2.5. Data analysis

Data analysis is performed in Matlab using a custom-written program. The spectrometer commercial software (GHOST) outputs data which are already frequency-calibrated. These raw spectra are first pre-processed, specifically background subtracted and denoised using wavelet analysis according to protocols presented in [28], and then spectrally analysed using least-squares line fitting [29].

### 3. Results

Firstly, we characterise the spectral background of the HCF system to validate that it does not pose a problem for Brillouin microspectroscopy. To demonstrate this, multiple spectra were taken without any sample at the distal side with an incident power of 20 mW from the laser, of which approximately 12 mW was successfully coupled into the HCF, yielding an input coupling efficiency of ($\eta = 60\%$). The signal was acquired for over 2 minutes to achieve sufficient SNR for fitting and the results can be found in Fig. 4(A), represented by the blue solid curve. Most notably, the problematic, wideband background that is characteristic of Raman scattering in silica fibres [30] can no longer be observed. There are, however, some weak residual spectral features at two distinct locations within the free spectral range (FSR) of the spectrometer (from $-37.78$ GHz to $+37.78$ GHz). As the Brillouin doublet is symmetrically positioned either side of the laser frequency, only the positively shifted peaks (Anti-Stokes) are presented in the figures for conciseness. Namely, there are identifiable Brillouin peaks appearing at 10.82 GHz and further away at around 25 GHz. Upon closer inspection the latter consists of two closely spaced features at 23.93 and 25.74 GHz, respectively. Qualitatively, the more prominent features towards the edges of the FSR correspond to relatively fast phonon modes and appear to be characteristic of common glass materials [31], whereas the other background signal resembles that of plastic materials (e.g. poly-acrylate plastics) previously measured in our lab. It is then reasonable to assume that these features correspond to the Brillouin scattering in the glass anti-resonant structure that serves as an effective cladding in the fibre and in the outer, protective polymer jacket respectively.

To test this hypothesis, the optics for coupling into the fibre was deliberately defocused to decrease the coupling efficiency to $\eta = 10\%$, which led to a much larger amount of light in the cladding and outer jacket regions. The resultant spectrum acquired over the same amount of time is also shown in the same figure with an arbitrary offset for clarity (black solid curve), where a visible increase in intensity can be observed for all background features as expected, albeit not proportionally, due to the different propagation lengths of cladding and jacket modes in the HCF, which makes the estimation of effective scattering volumes difficult. Interestingly, apart from confirming the existence of at least two modes in the far region, the background spectrum also sees the introduction of an additional mode at 21.90 GHz. These different glass-induced Brillouin peaks can be attributed to the existence of higher order modes propagating in the fibre. Existence of such modes have been reported previously for short sections of the fibre with a similar design and can be attributed to incomplete fulfilment of the anti-resonance condition due to deviation of the fibre fabrication from the design parameters [32]. A sufficient length of fibre and some bending actually aid in preventing these higher order modes from propagating, which is another advantage of the HCF, since it allows for a robust device which focuses on mechanical flexibility considerations. These higher order modes (LP11, LP02, LP20) differ from the fundamental mode...
in terms of propagation constants and thus can be phase-matched to different phonon modes according to a continuous dispersion relation in the same glass material. All of the glass modes, however, do not present any significant hurdle to Brillouin microspectroscopy and imaging, since they are located sufficiently far away from the spectral region of interest (ROI = 12 GHz) where peaks for common soft materials are expected.

The mode associated with the polymer jacket of the HCF, however, presents inconvenience for fibre-integrated Brillouin imaging as its spectral region overlaps with the ROI. To reduce the background associated with the fibre jacket to a negligible level, the first 10 cm of the HCF at the proximal side of the fibre were stripped, which was sufficient to almost fully eradicate the plastic mode due to its short propagation lengths in the fibre. Additional spatial filtering was also implemented before the entrance aperture of the spectrometer, which induces a further 30% decrease in systematic throughput. Any non-background signals, however, are largely unaffected, as only the non-fundamental modes possess different numerical apertures (NAs) and generally traverse on the outer rim of the beam. Any remaining background was easily filtered by software during post-processing as the contaminants are known.

To calibrate the spectral response of the system, the signal from distilled water and hydrogel were recorded by the probe with 8 mW at the sample, and an acquisition time of 2 minutes to facilitate accurate line fitting. Both spectra are presented in Fig. 4(B). Spectral analysis of the measurement data for water produces an average frequency shift value of 6.01 GHz with a linewidth of 0.733 GHz. While the spectral shift agrees well with the value expected for the laser wavelength (6.05 GHz at 660 nm) at room temperature [33], the width of the peak is wider than the theoretical value due to instrumental broadening caused by the imaging system and the spectrometer optics [34]. As the linewidth can be linked to the viscosity of the liquid [35] and provides extra information from the sample, the known width of water can be used to deconvolve and retrieve the instrumental response function, assuming that the system is linear [25]. By performing deconvolution as described in the preceding sections, the response function was approximated to follow a Lorentzian distribution with a linewidth of 0.626 GHz in the spectral ROI. To validate this calibration procedure, the Brillouin parameters obtained from gelatin-based hydrogel (GelMA) are compared. The Brillouin shift and linewidth of the sample were determined to be 6.97 GHz and 1.25 GHz respectively. Taking into account the spectral

![Fig. 4. Brillouin spectra acquired using the hollow-core fibre optic probe, both the resulting fits (solid line) and the raw data (dots) are presented. A) Background spectra at different coupling efficiencies (η); B) Spectra of water and GelMA hydrogel acquired after background removal.](image-url)
response of the system, the true linewidth of the hydrogel is then 0.624 GHz, which agrees with the value calculated from the viscoelastic parameters of the GelMA hydrogel [36].

To demonstrate the ability of the probe to obtain spatial information as well as spectral information, surface profilometry of a hydrogel-water phantom was performed by mounting the sample on a translation stage. The hydrogel-water phantom has been created to mimic the mechanical properties of biological tissues, since both possess high water content and a similar range of mechanical responses at GHz frequencies. A sketch of the phantom morphology is shown in Fig. 5(A) and both a coarse scan and a fine scan were performed along the directions indicated in the schematics, with step sizes of 0.5 mm and 0.2 mm respectively. 10 mW of incident power was focused onto the mounted slide and the sample was brought to focus such that maximum signal strength was achieved. The starting position was identified visually so that

![Surface profilometry results of the hydrogel-water phantom. A) Illustration of the sample orientation, the axis of scanning (dotted line) and scanning directions (red and blue arrows for course and fine scans, respectively). B) Spectral heat maps of raw spectral data, with colourmap corresponding to an intensity range of 5-100 counts. C) Profile of estimated average Brillouin shift and linewidth values at different positions of the scan. Deconvolved values and relevant statistical analysis can be found in the text.](image-url)
the line scan would go through the centre of the hydrogel droplet. The acquisition time for each spectrum was chosen to achieve a consistent SNR ($\approx25$). As the gel region produces a slightly weaker signal due to higher turbidity, the acquisition time thus varied between 30-40 seconds per point. The surface profile of the droplet can be visualised by simply stacking the raw spectra obtained along the direction of scanning, forming ‘heat maps’ in the third dimension due to the relative changes in the position and intensity of Brillouin peaks.

Figure 5(B) are such heat maps focusing on the Anti-Stokes peaks of the spectra that are generated by first denoising the spectral ROI [28]. It can be seen that there is a clear contrast that identifies the hydrogel region even with the coarser scan. Interestingly, the seemingly spurious features at the edges of the droplet are reproducible and can be explained by more solid layer of gel produced at the edges of the droplet during UV crosslinking stage. Quantitatively, it is also possible to plot the surface profile of the phantom in terms of the Brillouin shift and linewidth values, both are presented below in Fig. 5(C), by blue and yellow lines respectively. Apart from recreating the profile shape observed in the raw data, from the values obtained with the fine scan, the water region gives an average shift of 6.08 GHz, with a standard deviation of 0.02 GHz. In the hydrogel region, excluding the outliers at the edges, an average shift of 6.92 GHz and standard deviation of 0.04 GHz are calculated. In terms of the spectral widths, the fitting yields average linewidths of 0.0835 GHz and 0.489 GHz, with standard deviations of 0.0466 GHz and 0.067 GHz for water and hydrogel respectively, after removing the measurement system response function.

4. Conclusion

To the best of our knowledge, we have demonstrated, for the first time, a fibre optic Brillouin probe that can be used to simultaneously collect spectral and spatial information in a non-contact manner. The spectral performance of the system is directly comparable to that of a free-space system and no longer suffers from the background signal generated by spontaneous Brillouin and Raman scattering in the fibre that was detrimental in previous designs [17]. The probe is sensitive to both changes in the Brillouin shift and linewidth, which all contain information about the viscoelastic properties of the sample. The spatial sensitivity is facilitated by a custom-designed, miniaturised objective, which retains a sufficient working distance for non-contact signal collection while producing a focal spot of 10.16 $\mu$m, which governs the spatial resolution of the probe. A straightforward improvement could be to modify the optical design in order to achieve higher resolution for imaging applications. Such designs are commercially available and able to achieve an NA as high as 0.3 [37], which would make the resolution of the miniaturised system more comparable to the microscope equivalent. For some samples, rather than higher resolution, it is more attractive to have a high acquisition speed for real-time operation. For example, intravascular plaque detection only requires a spatial resolution on the order of 50-200$\mu$m, but requires acquisition speed that falls within the heart’s pulsatility rate (1 Hz) [38]. To improve on this particular aspect in future implementations, both the system throughput and the scanning mechanisms have to be enhanced. For instance, a VIPA (Virtually-Imaged Phase Array) based spectrometer can be incorporated which would enable acquisition speed on the order of tens of milliseconds per spectrum [22]. A fast, resonant fibre scanning variant also appears to be more favourable as they have already been demonstrated to function well with a confocal fibre system while maintaining the flexibility and dimension of a single fibre [39]. The imaging time, however, still fundamentally depends on the device SNR. In this vein, non-linear Brillouin modalities are emerging as promising alternatives [40,41], which can provide significant improvement to both the SNR and hence imaging speed. While the fibre optical instrumentation of a non-linear device would require further re-design and optimisation, the hollow core fibre is naturally affinitive to non-linear processes [19] and thus remains the ideal choice of waveguide. Finally, the current design can also be converted into a multimodal instrument. Fibre-based Raman probes have
been developed since the 1990s and thus benefit from being a more mature technology [42]. The inclusion of a Raman channel, for example by adding a ring of collection fibres around the HCF, would be an ideal addition to our probe. The chemical sensitivity of the Raman mode would allow the quantitative monitoring of water content in the sample, which has been demonstrated to be a dominant factor in the Brillouin shift values measured [43] and may help in the retrieval of pure mechanical properties in the future.

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See Supplement 1 for supporting content.

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