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## Fabry Perot polymer film fibre-optic hydrophones and arrays for ultrasound field characterisation

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**Abstract.** An optical ultrasound sensing method based upon the detection of acoustically-induced changes in the optical thickness of a Fabry Perot (FP) polymer film sensing interferometer has been developed as an alternative to piezoelectric based detection methods for ultrasound measurement applications. The technique provides an inherently broadband (~30MHz) response and excellent detection sensitivities (<10kPa), comparable to those of piezoelectric PVDF transducers. An important distinguishing feature however is that the sensing geometry is defined by the area of the polymer sensing film that is optically addressed. As a result, very small element sizes can be obtained to provide low directional sensitivity without compromising detection sensitivity – a key advantage over piezoelectric transducers. It also means that, by spatially sampling over a relatively large aperture, a high density ultrasound array can readily be configured. Other advantages are that, the sensing element can be inexpensively batch fabricated using polymer film deposition techniques, has the ability to self-calibrate, is electrically passive and immune to EMI. A range of measurement devices using this type of sensor have now been developed. These include a miniature (0.25mm o.d.) optical fibre hydrophone for *in situ* measurements of diagnostic and therapeutic medical ultrasound exposure. By rapidly scanning a focused laser beam over a planar FP sensor, a notional array of 3cm aperture, 50µm element size and 200µm interelement spacing has also been demonstrated for rapid transducer field mapping applications. It is considered that this ability to fabricate acoustically small, highly sensitive receivers in a variety of configurations offers the prospect of developing a valuable new set of ultrasound measurement tools.

### 1. Introduction

For accurate ultrasound field measurement, sensors that are small compared to the acoustic wavelength are required to provide an omnidirectional response and avoid spatial averaging – sub 20µm element sizes are required for an isotropic response at 10MHz for example. Achieving adequate signal-to-noise ratios with piezoelectric elements of this size presents a significant challenge due to the reduction in sensitivity with decreasing element area. A solution may lie in the use of optical methods whereby an optical beam is used to address a sensor. Since the dimensions of the addressing beam can, in principle, be as small as the optical diffraction limit of a few microns and the sensitivity is generally independent of element size there is the prospect of overcoming the sensitivity-element size

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limitations of piezoelectric transducers. Various methods have been investigated in both single element detector and array configurations for ultrasound field measurement. Perhaps the simplest approach is based on the detection of pressure-induced changes in reflectivity at the tip of an optical fiber<sup>1</sup> or an extended planar surface<sup>2</sup>. Low sensitivity, however, tends to limit its application to the measurement of relatively high amplitude sources, such as shockwave lithotripters. More sensitive methods employ interferometry. Methods based upon the detection of acoustically-induced changes in the optical thickness of a multilayer dielectric stack<sup>3</sup> and a polymer film Fabry Perot (FP) sensing interferometer<sup>4</sup> stand out as being particularly suitable for ultrasonic metrology. Both have been shown to offer wideband detection sensitivities, comparable to PVDF detectors, and can be configured as practical and robust measurement tools in the form of optical fibre hydrophones<sup>5,6,7</sup> for point measurements and 2D arrays<sup>8,9</sup> for field mapping applications. In this paper, an overview of the fabrication, underlying transduction mechanisms and acoustic performance of Fabry Perot polymer film sensors is presented. Two specific measurement devices that employ these sensors are described; a 2D planar array and an optical fibre hydrophone.

## 2. Sensor Fabrication

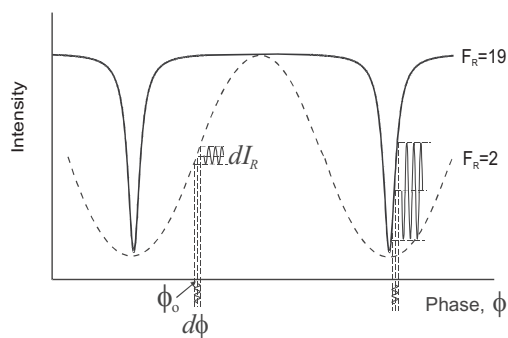
The sensor consists of a polymer or glass planar substrate, or the tip of an optical fibre, on to which a partially reflective metallic or dielectric coating is deposited. This is followed by the vacuum deposition of a polymer (Parylene C) film spacer of thickness in the range 5-50 $\mu\text{m}$ , depending on the bandwidth required. A near fully reflective coating is then deposited on to the polymer film spacer. The two reflecting layers either side of the spacer form the mirrors of the FP interferometer. Fabricating the sensors in this way enables them to be mass produced at low unit cost with high repeatability and has been shown to provide an extremely rugged sensing element.

## 3. Transduction mechanism

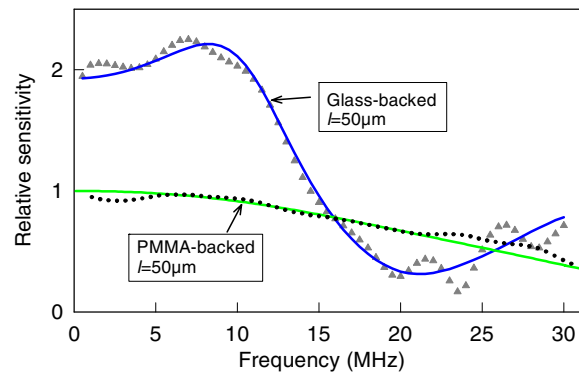
The transduction mechanism comprises two processes. Firstly, assuming the elastic limits of the polymer sensing film are not exceeded, an incident acoustic wave produces a linear change in the optical thickness of the film. Secondly, the resulting optical phase shift  $d\phi$  is converted to a reflected intensity modulation  $dI_R$  via the intensity-phase transfer function (ITF) of the interferometer as shown in figure 1. The ITF is not strictly linear although by appropriately setting the phase bias  $\phi_0$  or working point of the interferometer by tuning the laser wavelength<sup>4</sup> or angle of the incident beam<sup>3,10</sup>, small acoustically induced phase shifts can be detected with acceptable linearity.

## 4. Performance

*Sensitivity:* There are two components to the overall sensitivity. First, the slope of the ITF at the working point  $\phi_0$  gives the sensitivity of the reflected intensity to a change in phase, the phase sensitivity in  $\mu\text{W}/\text{rad}$ . This depends on the intensity of the incident light and the reflectivity finesse  $F_R$  of the interferometer. Second, the sensitivity of the phase to a change in the acoustic pressure, the acoustic sensitivity in  $\text{rad}/\text{MPa}$ , which depends on the elastic and photoelastic properties of the polymer film. Typically this is 0.15  $\text{rad}/\text{MPa}$  for a 50 $\mu\text{m}$  thick rigid backed Parylene film<sup>7</sup>. The product of the phase and acoustic sensitivities provide the overall sensitivity in  $\mu\text{W}/\text{MPa}$ . With knowledge of the minimum detectable intensity modulation reflected from the sensor (a function of the laser and detector noise characteristics), a value for the noise-equivalent pressure (NEP) can then be determined. Typical wideband (25MHz) peak NEPs for low to medium finesse FPIs are around 10kPa<sup>4,7</sup>. Recent developments have shown that NEPs as low as 0.1kPa (comparable to that of a 1mm diameter PVDF element) can be achieved by using a sensor of higher finesse ( $F_R=19$ ) and a novel photodiode configuration that enables much higher laser powers to be used without saturating the detector<sup>11</sup>.



**Figure 1.** Intensity-phase transfer functions (ITF) for Fabry-Perot interferometers of reflectivity finesse  $F_R=2$  and  $F_R=19$



**Figure 2.** Experimental (dots) and theoretical (solid lines) frequency responses of polymer (PMMA) and glass backed  $50\mu\text{m}$  FP sensors<sup>4</sup>.

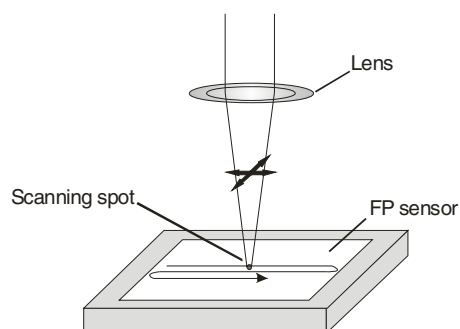
*Linearity:* The highest pressure at which the sensor is linear depends on the finesse of the FPI and the thickness and backing of the polymer film. A low finesse sensor,  $F_R=2$ , with a  $25\mu\text{m}$  thick rigid backed sensing film is linear to within 10% up to about 11MPa with a dynamic range of 60dB<sup>7</sup>. This reduces to 1MPa for a sensor of finesse  $F_R=19$  although the dynamic range actually increases to 69dB - the reduction in the lower limit of detection due to the increased finesse more than compensates for the reduction in the upper limit of linear detection.

*Frequency response and directivity:* The angle and frequency dependent response depends on the acoustic properties of the sensor film, the backing material and of the surrounding fluid as well as the area of the illuminated spot. Assuming thickness mode operation, the normally incident frequency response for a polymer backed sensing film will drop to zero when the acoustic wavelength equals the thickness of the FP sensing film. A polymer backed  $50\mu\text{m}$  Parylene sensor film has a 3dB bandwidth of approximately 20MHz as shown in figure 2. By using a thinner spacer of  $10\mu\text{m}$ , a bandwidth in excess of 100 MHz could readily be obtained whilst retaining acceptable (kPa) NEP. Whether it will be possible to obtain true optical diffraction limited effective element sizes of a few microns remains to be seen. However, directivity measurements<sup>7</sup> of a sensor illuminated by a  $6\mu\text{m}$  diameter beam suggests that effective radii down to at least  $50\mu\text{m}$  (and possibly significantly lower) can be achieved.

## 5. 2D array for ultrasound field mapping

By illuminating a large area of a planar FP polymer film sensor and detecting the reflected output beam with a photodiode or CCD array, or a mechanically scanned photodiode<sup>8</sup>, a 2D ultrasound receive array can be realised. An alternative approach is to scan a focussed laser beam over the sensor head (figure 3) and detect the reflected output beam with a single detector. This offers the prospect of synthesising an array of near arbitrary aperture and geometry and optical diffraction limited element sizes and interelement spacings.

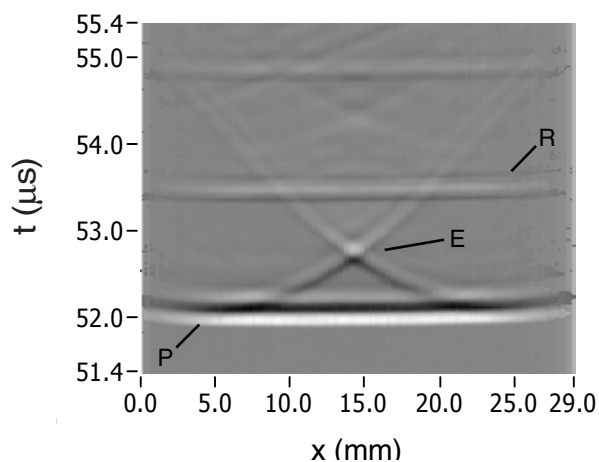
A PC controlled system that employs 2 high speed galvanometer mirror to rapidly scan a focussed laser beam over the sensor has been developed. At each point of the scan, the optimum phase bias  $\phi_0$  is located by tuning the laser over the FPI free spectral range to recover the ITF and then returning to the wavelength that corresponds to the peak value of ITF phase derivative, the point of maximum sensitivity. The detected acoustic waveform is then captured with a digitising oscilloscope (DSO) and downloaded to a PC. To correct for variations in sensitivity from point to point over the sensor head



**Figure 3.** Rapid spot-scanning of FP sensor to synthesise an array

due to imperfections in the FPI or changes in incident laser power, the captured waveform is divided by the corresponding value of the ITF phase derivative (the phase sensitivity) - in this way the system can be regarded as being self calibrating.

The system has been demonstrated by mapping the output of a pulsed 3.5MHz, 25mm diameter planar PZT transducer over a 3cm line-scan, in steps of 200 $\mu$ m using a 50 $\mu$ m diameter optical spot. Figure 4 shows the initial plane wavefront P emitted by the transducer followed by the inverted edge wave components E originating from the transducer perimeter. These appear as a characteristic “X” shaped feature with a signal maximum at the centre of the scan where the scan line passes through the transducer axis. R is the reflection of P from the back surface of the 3.85mm thick glass stub on to which the polymer sensing film was deposited.



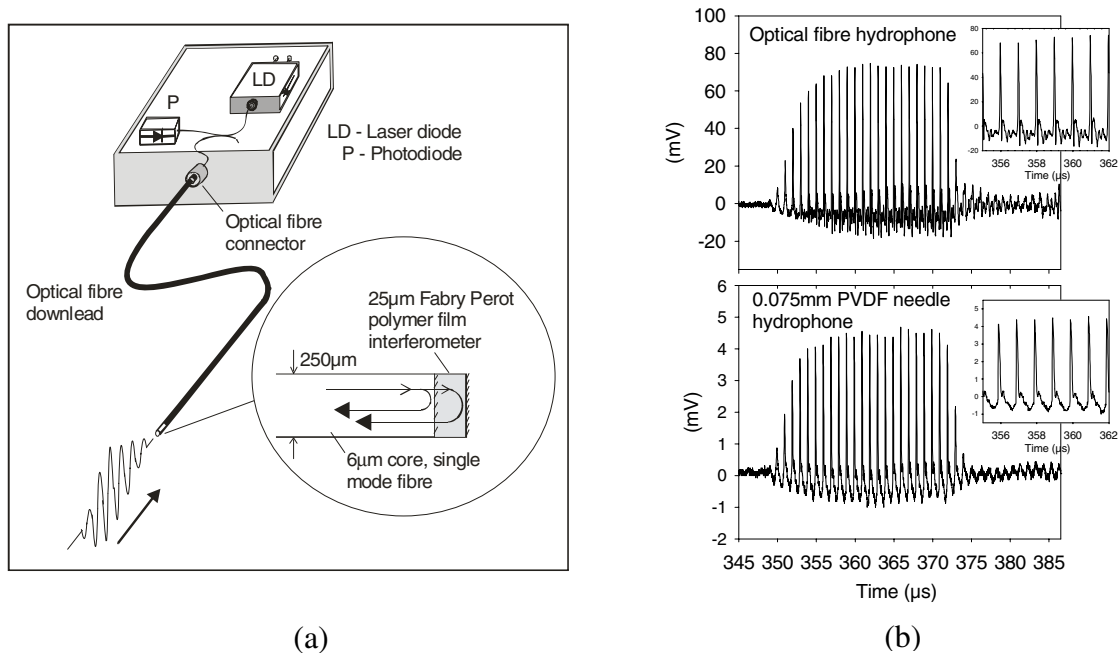
**Figure 4.** Line scan of a pulsed 3.5 MHz plane wave PZT transducer. P – initial plane wave, E - edge wave, R – reflection from back surface of 3.85mm glass FPI backing stub. Transducer-sensor separation=5.2cm. Scan increment  $dx=0.2$ mm, spot size=0.05mm.

The scanning time was 1s per point, limited by the time taken to download the acoustic waveform from the DSO to the PC via a GPIB interface. To reduce this, a data acquisition system, such as a PC digitising card with a sufficiently deep memory, that could store all the acoustic waveforms acquired over a scan on a single record and then download the entire record to the PC in a single step could be used. The point-point scan time would then be limited by the galvanometer and laser tuning speed and could be as low as a few ms. The dimensions of the current scan region were limited by the aperture of the focussing lens used and could readily be increased to that limited by the angular range of the galvanometers giving a sensing area of 5cm x 5cm. This concept offers a high speed, flexible

alternative to mechanically scanning a PVDF needle hydrophone for mapping the output of medical ultrasound transducers and arrays.

### 6. Optical fibre hydrophone for *in situ* ultrasonic field measurements

By depositing a 25 $\mu\text{m}$  FP polymer film sensor directly on to the cleaved end of a single mode optical fibre, a miniature wideband (20MHz) ultrasonic hydrophone probe can be realised (figure 5(a)). The sensitivity, linearity, frequency response and directivity have been compared to a range of PVDF needle and membrane hydrophones and are reported in reference 7.



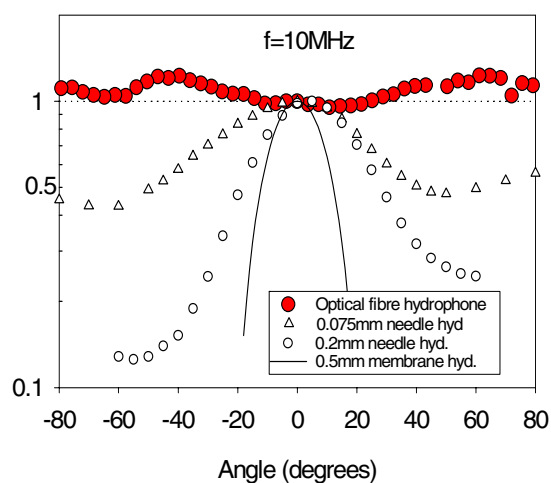
**Figure 5** (a) Schematic of fibre optic hydrophone system and (b) comparisons of optical fibre hydrophone output (top) with that of a 0.075-mm PVDF needle hydrophone (bottom) in response to a "shocked" 1-MHz toneburst. Insets show expanded timescale. No signal averaging was used.

Figure 5(b) shows a shocked toneburst detected by the optical fibre hydrophone and a 75 $\mu\text{m}$  PVDF needle hydrophone, the two devices showing comparable signal to noise ratios. Figure 6 shows an example of the directional response at 10MHz, illustrating a near isotropic response at this frequency on account of the small element size, which, to a first approximation, is defined by the dimensions of the 6 $\mu\text{m}$  core diameter of the fibre.

In addition to favourable acoustic performance, an important advantage over piezoelectric hydrophones is that the inexpensive nature of the sensor fabrication process offers the prospect of realising a disposable hydrophone concept for single use applications – eg for making *in situ* exposure measurements of diagnostic and therapeutic medical ultrasound where single use is required to avoid cross infection. The ability to sterilise the sensor downloads, the miniature, flexible probe-type configuration and low directional sensitivity are further advantages for this application.

### 7. Summary

FP polymer film ultrasound sensors can fulfill many of the requirements for measuring medical ultrasound fields. Wideband, high sensitivity, acoustically small sensors that can self-calibrate can be configured as 2D arrays for rapid field mapping applications or as miniature optical fibre hydrophones



**Figure 6.** Directivity of optical fibre hydrophone and PVDF needle and membrane hydrophones.

for *in situ* measurements. The low unit cost of the sensors means that they can be regarded as consumables that can be inexpensively replaced for single use *in vivo* applications or for measuring high intensity focussed ultrasound fields where there is a risk of damaging expensive PVDF hydrophones. To date, bandwidths have been limited to around 20MHz. Given the increasing use of high frequency ultrasound, such as that used in ocular medicine and harmonic imaging, future work will be targeted towards extending the bandwidth to 100MHz. As indicated in section 4 this can readily be achieved by reducing the polymer film thickness to 10 $\mu$ m whilst retaining high sensitivity by increasing the FPI finesse and illuminating with a higher laser power<sup>11</sup>.

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