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Rapid spatial mapping of the acoustic pressure in high intensity focused ultrasound fields at clinical intensities using a novel planar Fabry-Pérot interferometer

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Measurement of high acoustic pressures is necessary in order to fully characterise clinical high-intensity focused ultrasound (HIFU) fields, and for accurate validation of computational models of ultrasound propagation. However, many existing methods are unable to withstand the extreme pressures generated in these fields, and those that can often have high noise levels. Here, a robust sensor, based on a planar Fabry-Pérot interferometer with hard dielectric spacer and mirrors, was used to measure acoustic pressure in the field of a 3.3 MHz single element spherically focused bowl transducer. In preliminary measurements, peak positive pressures of 27 MPa, and peak negative pressures of 14 MPa were measured. The noise equivalent pressure scaled with the adjustable dynamic range of the system between 50 kPa for pressures up to 8 MPa and 235 kPa for measurements up to 70 MPa. This makes the system suitable for measuring low pressure regions of the field as well as the high focal pressures. The -3 dB bandwidth of the sensor was 600 MHz, and the effective element size was 25 µm, which makes the sensor well suited to the measurement of the highly nonlinear and localised high-pressure focal regions generated in HIFU fields. Waveforms were acquired at a rate of 200 Hz, several orders of magnitude faster than can be achieved with a hydrophone scanning system. This sensor represents a critical improvement in measurement capability for HIFU fields in terms of dynamic range, bandwidth, noise equivalent pressure, and acquisition speed.

HIFU, Field mapping, Fabry-Pérot, Acoustic pressure

I. Introduction

The measurement of high intensity focused ultrasound (HIFU) fields is critical in monitoring the stability of clinical ultrasound therapy systems, in validating models of ultrasound propagation, and in understanding these complex acoustic fields and their bioeffects. In order to fully characterise clinical therapeutic ultrasound fields, they must be measured at clinical levels where extremely high pressures of up to 100 MPa are generated in the focal region [1]. In the ideal case, a suitable sensor for these conditions must be robust enough to withstand the extreme conditions, and have a high dynamic range and low noise levels suitable for measuring the high-pressure focal region and low pressures elsewhere. It must also have a wide bandwidth to capture high frequency harmonics, and a small element size to avoid spatial averaging in the narrow focal region and to provide an omnidirectional response. At present, there are no sensors that fulfil all of these criteria.

Currently available acoustic pressure sensors used for the measurement of ultrasound fields fall into two categories: piezoelectric sensors and optical sensors. Piezoelectric hydrophones are easily damaged by cavitation, and by heating and direct mechanical effects. While purification and degassing of the test medium (usually water) can increase the threshold for cavitation, and use of a low duty cycle will reduce heating, damage is likely to occur eventually. The pressure range of most piezoelectric sensors is limited, often to 10 or 20 MPa. This is due to the dynamic range of components such as preamplifiers, as well as damage thresholds, and it limits them to the measurement of relatively low amplitude fields [2]. While piezoelectric hydrophones with element sizes as small as 40 µm are available, decreasing sensor size leads to a lower sensitivity and the frequency response can be non-uniform. Robust piezoelectric hydrophones have also been reported, but have been limited to large element sizes [3]–[5]. Fibre optic hydrophones (FOPHs) are suitable candidates due to their small size and wider bandwidth. Bare cleaved fibre tip FOPHs have been used to measure pressures up to 80 MPa. They are robust and have the advantage that they can easily be cut to form a new fibre tip when damaged [6], [7]. However, their high noise equivalent pressure (NEP) of more than 0.5 MPa renders them unsuitable for mapping regions of lower pressure.

Higher sensitivity and lower noise levels can be achieved with a Fabry-Pérot interferometer based FOPH [8]. A Fabry-Pérot interferometer consists of two plane parallel mirrors separated by a cavity or spacer. Incident light is multiply reflected from the mirrors and the multiple beams interfere as they return. When the interferometer is placed in an acoustic field, the spacer thickness is modulated by the acoustic pressure. This causes a phase shift between light reflected from each mirror, which maps to a change in the reflected optical power. Fibre mounted and polymethylmethacrylate (PMMA) backed planar Fabry-Pérot sensors have been previously described for ultrasonic pressure and temperature measurement [8] and for photoacoustic imaging [9]. The planar sensors are formed of sputtered dichroic mirrors separated by a vacuum deposited Parylene-C polymer film spacer. Different spacer thicknesses have been employed, chosen to optimise the sensitivity and bandwidth for the given application. For example, as part of the photoacoustic scanner described in [9], a sensor with a 38 µm spacer had a noise equivalent pressure (NEP) of 0.21 kPa over a 20 MHz measurement bandwidth, and -3 dB bandwidth of 22 MHz. The bandwidth increased to 39 MHz with a 22 µm spacer at the expense of increased NEP of 0.31 kPa. These noise levels are extremely low compared to the 50 kPa NEP associated with a 75 µm PVDF needle hydrophone.

While very suitable for use in photoacoustic imaging when high sensitivity and extremely low noise are required for the detection of low amplitude signals, this polymer spacer construction is not robust to high pressures and has a small dynamic range (a few MPa). To overcome these limitations for use with HIFU, the interferometer can instead be formed from sputtered SiO_2 hard dielectric mirrors and spacer with a thickness of a few micron. This construction results in a robust sensor with a flat frequency response over 100s of MHz, and a large dynamic range [10]. The dynamic range of the sensor can also be optimised to enable measurement of both low and high pressures. By adjusting the power of the incident light source, the signal generated by returning light at the photodiode detector can be scaled to use the whole dynamic range of the photodiode. The associated noise equivalent pressure scales with this optimisation, so it remains low at 50 kPa over a 20 MHz bandwidth for low pressure measurements (up to 8 MPa), increasing to approximately 235 kPa over 20 MHz at the upper limits of measurable pressure of 70 MPa. The element size of the sensor is optically defined by the spot size of the incident laser light source (25 to 65 µm) which minimises spatial averaging and gives a broad directional response. Here, we present preliminary measurements of a HIFU field obtained with a planar Fabry-Pérot interferometer of hard dielectric construction.

II. Methods

A. Fabry-Pérot sensor and scanning system

The acoustic pressure was measured using a previously described optical scanner [9], with a hard dielectric planar Fabry-Pérot interferometer. The sensor was formed of a wedged glass substrate with 3.9 µm thick evaporated hard dielectric (SiO_2) spacer and dichroic mirrors. The -3dB bandwidth of the sensor was 600 MHz, the linear detection region was 70 MPa, and the effective element size was 25 µm. To interrogate the sensor, a 1550 nm focused laser beam (TSL-510, Santec Corporation, Japan) was raster scanned across the sensor using a galvanometer mirror scanning system with an acquisition rate of 200 Hz. To maximise sensitivity and linearity of the sensor, the wavelength of the incident light is tuned to the maximum gradient of the interferometer transfer function (ITF), the relationship between phase and reflected optical power, at each point. The reflected light was measured by a InGaS photodiode and the resulting voltage signal digitised and stored for each point to build up a 2D map of the time varying acoustic field distribution. The laser interrogation power was adjusted between scans to maximise the photodiode voltage.

Pressure calibration factors over the range of interrogation laser power settings were derived from measurements of the acoustic pressure generated by a 3.5 MHz pulsed ultrasound check source (Panametrics 500PR + V381, Olympus Industrial, UK), previously characterised with a calibrated PVDF needle hydrophone (Precision Acoustics, Ltd., Dorchester, UK).

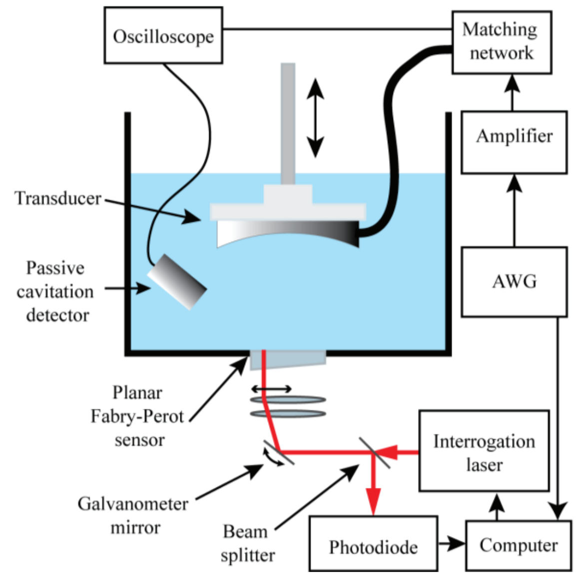

Fig. 1. Schematic of experimental configuration

B. Ultrasound source and driving system

The sensor, which had an area of 50 mm by 30 mm, was mounted in the base of a temperature controlled water bath. An acoustic field was generated by a single element focused bowl transducer (H101, Sonic Concepts, Bothell, WA, USA) mounted in the water bath filled with degassed, deionised water. A diagram of the experimental arrangement is shown in Fig. 1. The transducer had an active area diameter of 64 mm and focal length of 62.5 mm. The transducer was driven with a 4 cycle burst at 3.3 MHz. Input signals were generated by an Agilent 33522A Arbitrary Waveform Generator (Agilent, Berkshire, UK) before amplification by an E&I A300 RF power amplifier (Electronics & Innovation Ltd., Rochester, NY). This was coupled to the transducer via an impedance matching network. The transducer drive signal was monitored using an Agilent oscilloscope probe and Agilent DSO-X3204 oscilloscope (Agilent, as above). A 10 mm diameter 4 MHz PVDF transducer was used for passive cavitation detection (the signal and spectrum were monitored using the oscilloscope).

III. Results and Discussion

Areas of up to 20 mm by 20 mm were scanned with step sizes of 50 µm to 200 µm with scans containing up to 40,000 measurement points. All scans were performed in single acquisition mode and scan times were on the order of 1 minute.

A. Measurements at different drive levels

Field scans were performed at a range of transducer drive levels from 38 Vpp to 225 Vpp in the focal plane of the acoustic field. Peak positive pressures of up to 27 MPa were measured, with peak negative pressures up to 14 MPa. Pressure waveforms and spectra measured in the focal region are shown

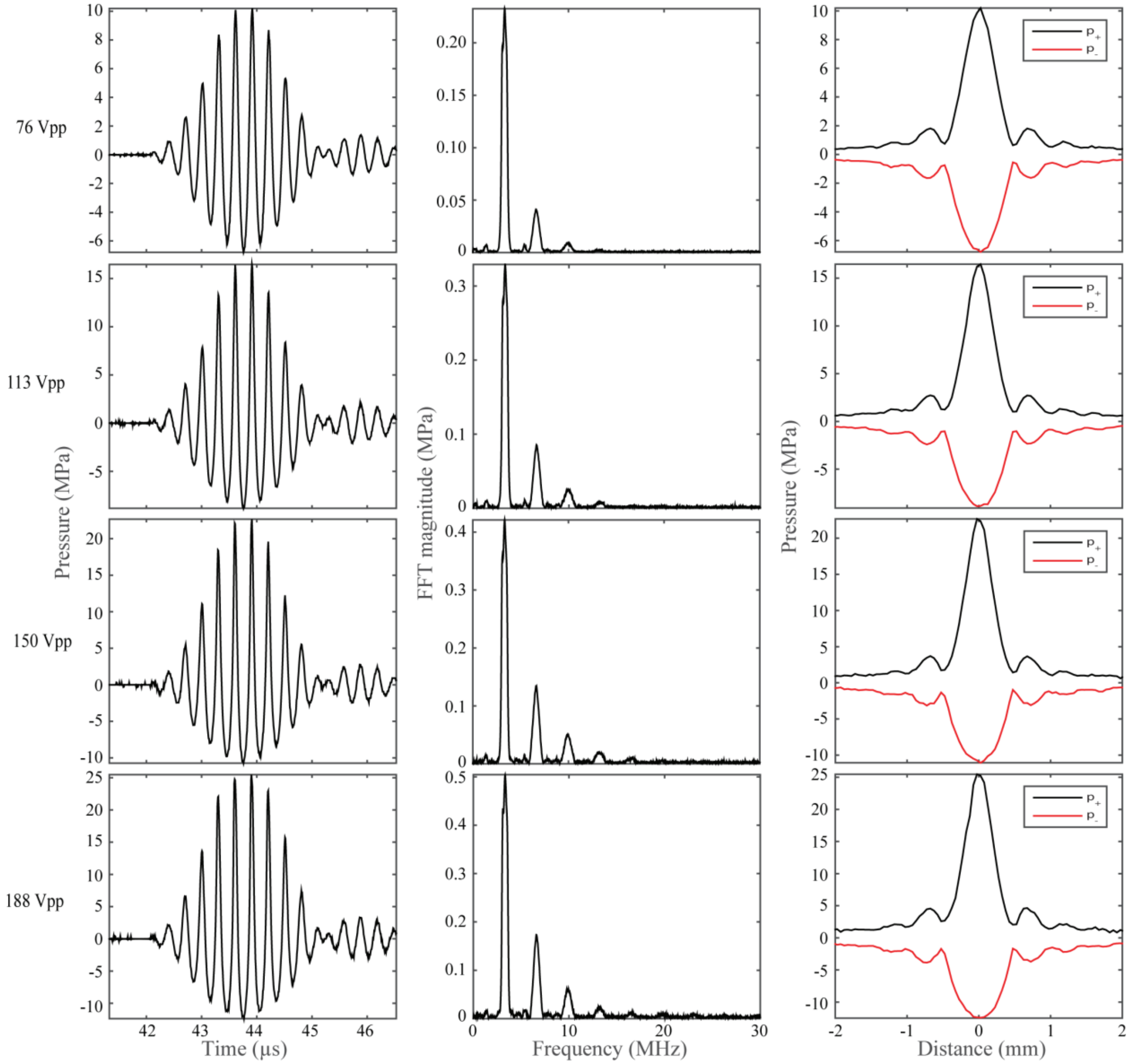

Fig. 2 Single shot focal waveforms, their spectra (smoothed by fitting with a spline) and transverse beam profiles at the focal plane measured at drive levels of 76 Vpp, 113 Vpp, 150 Vpp and 188 Vpp. Scan parameters: step size 50 µm, scan area 6 mm by 6 mm, scan time 72 s.

in Fig. 2. These data show the acoustic waveform becoming increasingly nonlinear as the drive level is increased, with increasing amplitude of the harmonics visible in the spectra. The extremely wide bandwidth and flat frequency response of the hard dielectric Fabry-Pérot sensor make it ideal for measurement of highly nonlinear ultrasound signals which contain many harmonics, especially when compared to the 40 – 60 MHz bandwidth typical of conventional PVDF hydrophones. Temporal peak values were extracted from the time varying pressure waveforms at each point in the scan. Transverse peak positive and peak negative pressure profiles at each of the drive levels are shown in the right hand column of Fig. 2. The small element size of the sensor helps limit spatial averaging which could cause underestimation of spatial peak pressures in the focus as the peak narrows with increasing drive level [1]. The transverse beam profiles show increasing difference between the temporal peak positive and negative pressures, and several side lobes are visible. In further measurements not shown here, pressures as high as 49 MPa were measured. When the drive level was increased further, cavitation occurred. However, there was no indication of damage to the sensor following this, and measurements were continued at lower drive levels.

B. Measurements at different axial positions

2D field scans were performed at several planes at axial distances of up to 5mm from the focal plane. 2D plots of the peak positive and negative pressure are shown in Fig. 3. The transducer was driven at 152 Vpp resulting in a peak positive pressure of 22.7 MPa and peak negative pressure of 11.3 MPa at the focus. All scans in this data set were performed with the same interrogation laser power, which enabled a maximum measureable pressure of 37 MPa and a NEP of 160 ± 18 kPa (calculated over a 20 MHz bandwidth). At each of the planes there is an axial maximum surrounded by several side lobes. The smallest clearly visible side lobes have amplitudes of

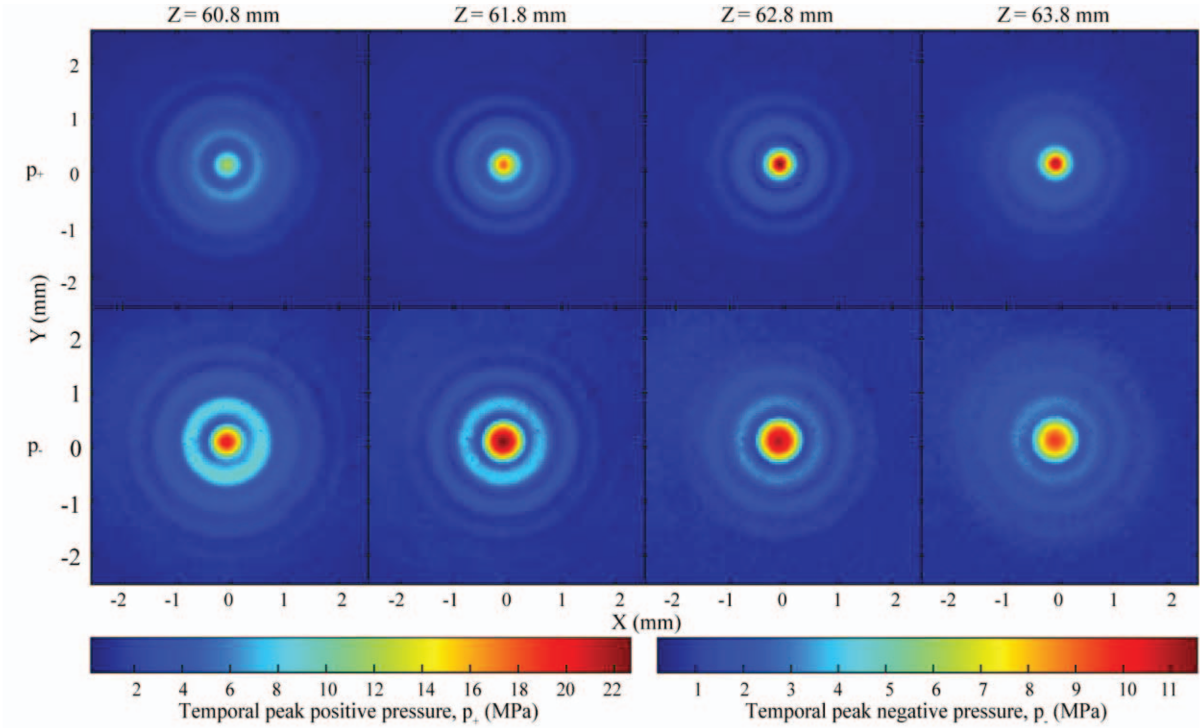

Fig. 3 2D field maps of temporal peak positive (top row) and temporal peak negative pressure (bottom row) at XY planes positioned at distances of 60.8 mm, 61.8 mm, 62. 8 mm (focal plane) and 63.8 mm from the origin of the transducer. Scan parameters: step size 50 µm, scan area 5 mm by 5 mm, scan time 50 s.

approximately 1.5 MPa. The sensitivity and dynamic range of the sensor is suitable for mapping both high and low pressure features of the field. The mapping of lower pressure regions could be further optimised by adjusting the interrogation laser power for each plane.

IV. Conclusion

A novel planar hard dielectric Fabry-Pérot interferometer is presented here for rapid measurement of acoustic pressures in HIFU fields. Two-dimensional maps of acoustic pressure were obtained with a scan rate of 200 Hz which gave scan times on the order of 1 minute. This acquisition time is very rapid when compared to the 4-8 hours required to run a hydrophone scan of the same size. Peak pressures of up to 27 MPa were measured, with NEP that scaled with the maximum measurable pressure over the range of interrogation laser powers. In the configuration described here, the dynamic range of the system was approximately 63 dB. This range could be increased by employing a lower noise interrogation laser. The extremely wide bandwidth of the sensor renders it ideal for measurement of highly nonlinear fields containing many harmonics. The small size of the sensitive element (25 µm) minimises spatial averaging. Overall, this technique represents a critical improvement in measurement capability for HIFU fields.

In further work, the range of pressures will be extended by improved purification and degassing of the water. A comparison will also be made between low-pressure hydrophone measurements and scans obtained with the Fabry-Pérot system.

V. References

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