

High-resolution imaging of ultrasonic fields through a multimode optical fiber

Benjamin Keenlyside^{1,2,*}, Dylan Marques^{1,2,3}, David B. Phillips⁴, Edward Zhang,¹
Paul Beard,¹ and James Guggenheim^{1,2,3}

¹*Department of Medical Physics and Biomedical Engineering, [University College London](#), London, United Kingdom*

²*Department of Cardiovascular Sciences, [University of Birmingham](#), Birmingham, United Kingdom*

³*Department of Electrical, Electronic and Systems Engineering, [University of Birmingham](#), Birmingham, United Kingdom*

⁴*Department of Physics and Astronomy, [University of Exeter](#), Exeter, United Kingdom*



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High-resolution ultrasound and photoacoustic imaging with flexible, lightweight probes is highly desirable for applications ranging from handheld clinical diagnostics to minimally invasive endoscopy and industrial inspection in confined environments. Piezoelectric transducer arrays, while widely used, face limitations in resolution, bandwidth, and scalability. To overcome these limitations, we present an approach that combines optical ultrasound detection using a Fabry-Pérot ultrasound sensor (FPUS) with light delivery through a single multimode optical fiber (MMF), utilizing wavefront shaping for beam control. This approach offers the capability to combine high-resolution imaging, a compact and flexible form factor, and software-defined reconfigurable array read-out. We present a prototype imaging system in which an FPUS is interrogated through a 200- μm -core-diameter MMF supporting ~ 2000 spatial light modes, and use this system to map the ultrasound fields produced by known transducers. We demonstrate high-resolution, high-sensitivity ultrasonic field mapping, paving the way to a new class of versatile handheld imaging devices and endoscopes in medicine and industry.

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I. INTRODUCTION

High-resolution ultrasonic and photoacoustic imaging using flexible, lightweight probes is highly desirable in many application areas. These include handheld or minimally invasive medical imaging, intraoperative surgical guidance [1], and industrial inspection in confined or hard-to-reach environments [2]. Achieving this combination of high-resolution imaging with a flexible and lightweight form factor remains challenging: conventional piezoelectric transducer arrays, while compact and widely used, face limitations in spatial resolution, miniaturization, scalability, and detection bandwidth [3]. Improving on the resolution, bandwidth, and element count in handheld probes could transform their utility in medicine and industry, enabling, for example, clinical assessment of microvascular structure and function beyond the resolving power of

existing probes. Simultaneously, increasing the resolution and miniaturization of endoscopes could unlock new applications in minimally invasive disease monitoring (e.g., assessment of gastrointestinal cancers or vascular plaques), surgical guidance, or nondestructive inspection through narrow openings.

To address this unmet need, we present an imaging approach that combines optical ultrasound detection with light delivery and beam scanning through a single multimode optical fiber. This advance enables versatile, high-performance imaging in a format suitable for handheld or endoscopic use, offering advantages in miniaturization, scalability, and versatility over the existing state of the art.

The sensor used for optical ultrasound detection is a Fabry-Pérot ultrasound sensor (FPUS) [4]. Compared with other sensor types (e.g., piezoelectric arrays or CMUTs) [5], these sensors offer an exceptional combination of high resolution ($< 50\ \mu\text{m}$), broad bandwidth ($> 20\ \text{MHz}$), and the potential for massively multiplexed detection ($> 10^5$ array elements) [6]. This capability has made the FPUS attractive in areas such as ultrasound field characterization [7], pulse-echo ultrasound imaging [8], and photoacoustic imaging in both microscopy [9] and tomography [6] modes, with applications including the assessment of breast cancer, diabetes, and arthritis identified [6].

*Contact author: b.p.keenlyside@bham.ac.uk

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In structure, an FPUS is a compressible optical micro-cavity formed by two highly reflective planar mirrors separated by a transparent polymer layer [4]. When illuminated by a laser, the cavity acts as an interferometer, with wavelength-varying reflectivity that sharply drops near specific resonance wavelengths determined by cavity thickness. Incident ultrasound waves cause transient changes in thickness, shifting the resonance wavelengths and modulating the reflected optical power. By interrogating the FPUS at a bias wavelength near resonance, the local acoustic pressure can be measured. In practice, this means a planar FPUS can be used as a virtual array of highly sensitive, broadband, and miniature ultrasound detectors by scanning a focused laser beam across the sensor and making measurements of the reflected power at each position.

In most existing FPUS systems, the interrogation beam is scanned across a centimeter-scale aperture using a series of rotating mirrors and free space optics [4]. While these setups can produce highly detailed images, their size and complexity make their integration into handheld or endoscopic devices challenging. Additionally, the structure of the interrogation beam (e.g., spot size) cannot be adjusted in these systems without hardware modifications. Recent interest in minimally invasive imaging using FPUS technology has produced systems in which these sensors are interrogated through multicore fiber bundles [10] or rigid endoscopes [11], with *in vivo* imaging capability demonstrated.

An alternative approach, taken in this work, is to interrogate an FPUS through a multimode optical fiber (MMF), capable of supporting many ($>10^5$) independent channels through a single, narrow (<1 mm), flexible, and lightweight fiber. Beam scanning and structuring through the fiber is then achieved using a technique known as wavefront shaping to precisely control light transmission through the fiber [12].

Imaging through MMF offers numerous advantages. In addition to their high mode density and flexibility,

MMF are also low-cost and tolerant to extreme conditions, including strong electromagnetic fields [13] and ionizing radiation [14], relevant, for example, for disposable clinical probes or MRI-compatible imaging systems. For these reasons, MMF-based imaging using wavefront shaping has gained traction across various optical microscopy modalities [15–18], including probes incorporating multiple modalities [19]. Extending this technique to perform high-resolution acoustic field mapping through MMF would represent a significant new capability in this area. Furthermore, the programmable beam control offered by wavefront shaping could allow the creation of fully reconfigurable ultrasound detection arrays, e.g., switching between a matrix or linear probe, or between arrays optimized for superficial or deep photoacoustic imaging.

The concept of scalable all-optical ultrasound imaging through MMF is illustrated in Fig. 1. An input field can be found that, via propagation through MMF, evolves into a beam that focuses to a particular location on the FPUS. Thus, by coupling a sequence of precalibrated optical fields into the MMF, a focused beam can be scanned across the FPUS, enabling ultrasonic field mapping. A preliminary demonstration of this approach [12] showed that spatially resolved ultrasound measurements could be made using an FPUS interrogated in this way. However, that work relied on a genetic algorithm [20] for wavefront optimization, requiring sequential optimization for each position. This slowed calibration and limited the number of interrogated points. As a result, only four positions were sampled, precluding high-resolution field mapping or imaging.

In this work, we demonstrate high-resolution ultrasound field mapping through MMF using a transmission matrix (TM) based approach to wavefront shaping [21]. This method enables parallel calibration of the wavefronts for thousands of focusing positions, allowing interrogation of over 6000 points on the FPUS. We then evaluate our system by mapping fields from known ultrasound sources and discuss the implications for future devices using this approach.

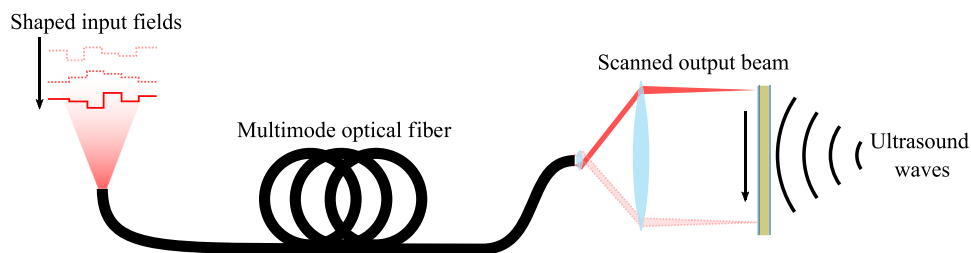


FIG. 1. Concept of an FPUS interrogated through a multimode optical fiber using wavefront shaping. Coupling an appropriately chosen wavefront into the fiber results in a focused beam directed to a particular position on the sensor. Scanning this focus across the sensor using a sequence of different wavefronts enables ultrasound fields to be mapped.

II. METHODS

A. Experimental setup

The basis of the experimental system used to interrogate an FPUS through an MMF is the ability to generate input wavefronts that produce focused spots on the sensor surface. This is achieved using a spatial light modulator (SLM) to control the phase of the input beam. Each wavefront is designed so that the modes in the fiber interfere constructively at a chosen location. To determine the required wavefronts, the transmission matrix of the fiber is measured. A schematic of the experimental setup used to achieve this is shown in Fig. 2.

In this setup, a beam from a 1500–1600-nm-tunable-wavelength interrogation laser (Santec TSL 570) is expanded and used to illuminate a reflective liquid-crystal SLM (Meadowlark 1920 × 1152 PCIe), which provides wavefront control. The modulated beam is then coupled into the MMF using a series of three lenses arranged in a 6f configuration: the first lens is placed one focal length from the SLM to perform a Fourier transform of the modulated field, and the subsequent two lenses relay and demagnify this transformed field onto the proximal facet of the fiber. The MMF has a core diameter of 200 μm and a numerical aperture (NA) of 0.2, supporting ~2000 modes at a wavelength of 1550 nm. The NA of the third lens was 0.3, greater than that of the fiber. Combined with illumination of a sufficient area and number of pixels on the SLM, this ensures that any combination of fiber modes could be excited. The proximal facet of the MMF is slightly offset

from the optic axis of the system, such that the 0th diffraction order does not enter the fiber and is filtered out. On the distal side, a 4f imaging system and a beam splitter project a magnified image of the fiber output onto both an FPUS and a coplanar camera (Hamamatsu C10633), with a magnification of 25×. In this configuration, the camera effectively images the illumination on the FPUS and enables TM measurement.

The FPUS used in this work was essentially the same in structure as that described in Huynh *et al.* [6], comprised of a 20-μm-thick layer of Parylene-C between two dielectric mirrors of reflectivity 98.5%, with a glass backing. Deionized water was used as a coupling medium during the ultrasonic field mapping experiments.

B. Wavefront shaping methods

To determine the wavefronts needed to focus at any position on the illuminated region of the FPUS through the MMF, a transmission matrix measurement approach [21] was used. As optical transmission through the fiber is linear, the relationship between any input field E_I and the corresponding output field E_O can be expressed in the following form:

$$E_O = \begin{bmatrix} E_{o1} \\ \vdots \\ E_{oM} \end{bmatrix} = \begin{bmatrix} t_{11} & \cdots & t_{1N} \\ \vdots & \ddots & \vdots \\ t_{M1} & \cdots & t_{NM} \end{bmatrix} \begin{bmatrix} E_{i1} \\ \vdots \\ E_{iN} \end{bmatrix} = TE_I, \quad (1)$$

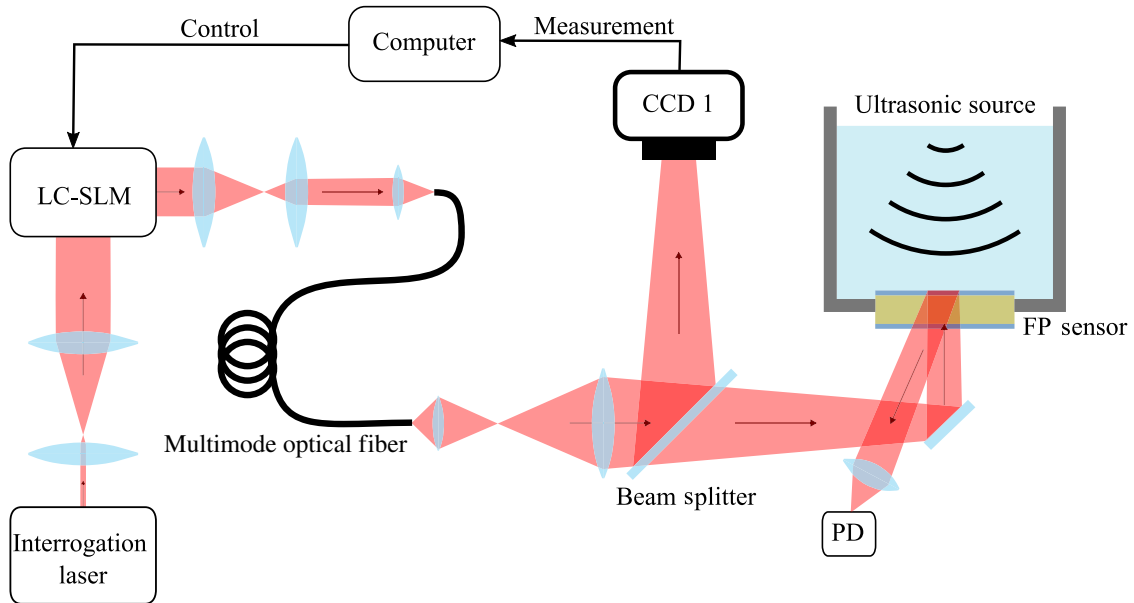


FIG. 2. Schematic of the experimental setup used to map US fields through MMF. The MMF used had a core diameter of 200 μm and a numerical aperture of 0.2. The FPUS used comprised a 20-μm Parylene-C spacer between two dielectric mirrors, with a glass backing layer. The beam envelope of the total illuminated area is shown, rather than individual focused beams formed by wavefront shaping.

where $\mathbf{T}$ is the complex-valued transmission matrix with elements t_{mn} that describes the amplitude and phase contribution of the n th input mode to the m th output mode in chosen bases.

The input basis chosen here was a set of beams focused at different positions on the proximal fiber facet. These are generated by displaying a phase grating on the SLM, with each position corresponding to a grating with a particular period and orientation. The canonical basis provided by the pixels of the CCD camera was used as the output basis.

To measure the transmission matrix of the fiber in these bases, a focused beam was scanned across an array of positions on the proximal fiber facet using the SLM. For each input position, the phase and amplitude of the output field on the camera were recorded holographically. A copropagating reference [21] was generated by projecting a second focused beam at a fixed position on the fiber using the SLM. The advantage of this choice of reference is that no portion of the SLM (or beam) needs to be dedicated to the reference.

To focus light onto the k th camera pixel through the MMF, the required wavefront $\mathbf{E}'_I$ is computed using the transmission matrix, i.e.,

$$\mathbf{E}'_I = \mathbf{T}^\dagger \mathbf{I}_k. \quad (2)$$

Here, $\mathbf{I}_k$ is the k th column of the identity matrix. Assuming the transmission matrix is approximately unitary, its conjugate transpose $\mathbf{T}^\dagger$ serves as an effective inverse. The corresponding phase pattern to display is then obtained by taking the Fourier transform of $\mathbf{E}'_I$, since the SLM is located in the Fourier plane of the fiber's proximal facet. This Fourier transform yields a complex-valued pattern, with the amplitude information discarded by normalization to obtain a phase-only pattern for display on the SLM.

In this work, the fiber TM was measured with ~ 6000 input modes, requiring 24 000 measurements (four per mode to recover phase information via phase shifting holography [21]). The full calibration process took around 40 min, limited primarily by the 10 Hz read-out speed of the camera.

C. Ultrasound field mapping procedure

Ultrasound (US) field mapping was performed in essentially the same way as ultrasound measurements presented in prior work [12], apart from a bias wavelength estimation. In this work, the single bias wavelength used across the entire sensor surface was determined by illuminating the sensor with an unshaped wavefront and measuring an interferometer transfer function (ITF). The bias wavelength was then taken as the wavelength of maximum slope on the falling edge of the ITF. This bias wavelength provides the maximum change in FPUS reflectivity with acoustic pressure and, thus, the maximum sensitivity to ultrasound.

The TM of the fiber was measured at this bias wavelength (necessary due to the finite bandwidth of the TM), and the phase patterns required to scan a focus across the FPUS were calculated. Finally, these wavefronts were sequentially displayed on the SLM. For each wavefront, the time-varying power reflected from the FPUS was recorded using a photodiode connected to an oscilloscope, synchronized with an acoustic source via a trigger signal. Each of these measurements yields a single ultrasound time trace at a specific spatial position. Collectively, these traces form a complete spatiotemporal map of the ultrasound field.

III. RESULTS

A. Wavefront shaping system

To interpret the system's imaging performance, it is important to first understand the quality of the optical foci generated through the multimode fiber. The spatial resolution, sensitivity, and uniformity of the ultrasound field maps are all expected to depend on the ability of the wavefront shaping system to deliver well-focused beams to specific locations on the FPUS. To evaluate the system's focusing performance, experiments were performed in which foci were generated in the plane of the FPUS and the camera (equivalent to focusing at the distal fiber facet) and images of the resulting optical fields were recorded.

Using the wavefront shaping method described in Sec. II, a focused beam was raster-scanned across a region of the camera. For each focus position, images of the optical field were captured. To enhance dynamic range, a multiple exposure technique was employed: images taken at varying laser powers were combined into a composite image, allowing both high- and low-intensity features to be visualized simultaneously.

Figures 3(a)–3(d) present representative images of the optical fields generated by wavefront shaping, displayed on a logarithmic scale to reveal both the bright focal spots and the surrounding speckled background. The foci are diffraction-limited, given the numerical aperture of the fiber (diameter $d \approx 4 \mu\text{m}$ in the plane of the distal fiber facet). In Figs. 3(a)–3(c), the foci are located in the fiber core and the focus to background peak intensity contrast is > 1000 . The image in Fig. 3(d) shows a relatively weaker focus, located at a position on the fiber cladding, showing that, in this setup, focusing through the fiber cladding was also possible at lower efficiency. Note that the ability to focus through the fiber cladding is expected to vary significantly with conditions such as the length and bending state of the fiber (due to variable losses and coupling between core and cladding). As such, focusing through the cladding is not expected to be of practical benefit for imaging under most conditions.

To quantitatively assess focusing performance, the total power delivered to the sensor and the fraction of this power

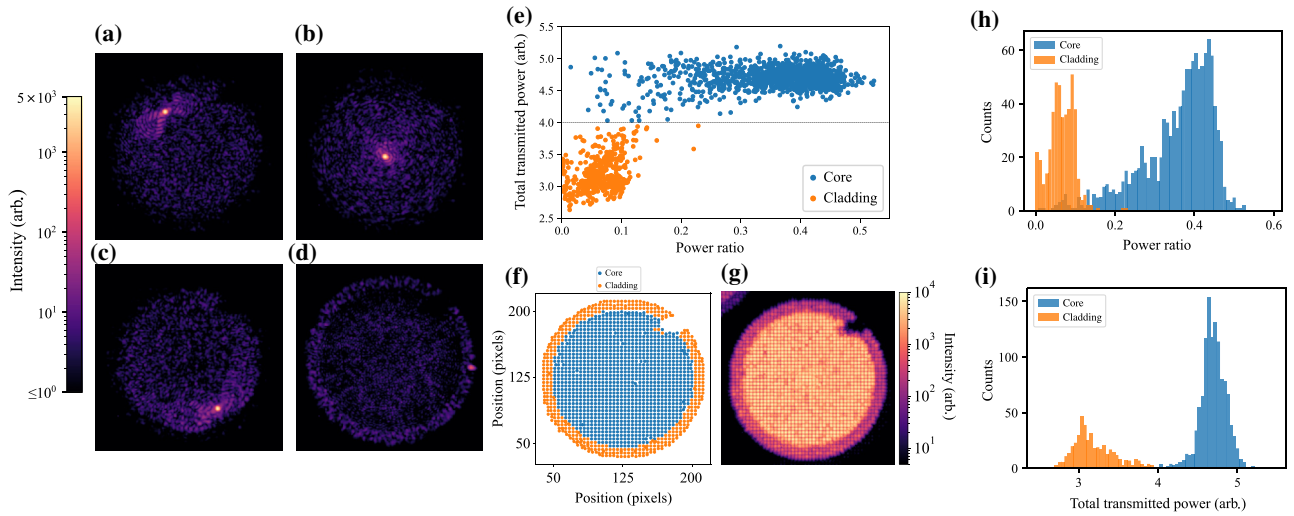


FIG. 3. Wavefront shaping performance. Example images of the fields produced when focusing through the (a)–(c) fiber core and (d) cladding, shown on a logarithmic color scale. (e) Total power transmitted through the fiber and the focal power ratio for a set of beams focused at different positions on the fiber tip. Foci have been divided into core and cladding foci based on the marked intensity threshold. The positions of the foci in each population are plotted in (f), compared with a maximum intensity projection of the images of each beam profile in (g). (h) Histogram of the achieved focal power ratio and (i) total transmitted optical power through the fiber, again with the core foci population separated.

contained in the optical focus were calculated. The focal power was defined as the summed intensity within a 3×3 -pixel region centered on the brightest pixel in each image. This region is equivalent to a 3.6×3.6 - μm region on the distal fiber facet, approximately equal to the diffraction limit given the numerical aperture of the fiber. These metrics are plotted for each focus in Fig. 3(e). To separate the foci transmitted from the fiber core from those transmitted through the cladding, we apply a specific threshold on the total transmitted power threshold, as marked in Fig. 3(e).

Figure 3(f) shows a spatial distribution of the foci, color-coded according to the threshold-based classification. For comparison, Fig. 3(g) presents a maximum intensity projection of all the recorded beam profiles. The nearly regular grid of points, which are bounded by a circular aperture, corresponds to the fiber's cladding. A distinct inner boundary, visible in both Figs. 3(f) and 3(g), marks the transition between the brighter core foci and the dimmer cladding foci, confirming that the power threshold effectively distinguishes the two regions. A dark region is also visible in the upper right-hand side of these plots, caused by a defect on the distal fiber facet.

Figure 3(e) shows that the total power transmitted through the fiber is relatively constant between the core foci ($\lesssim 20\%$ variation), while the focal power ratio varies more significantly. To aid interpretation, these quantities are plotted in histograms in Figs. 3(i) and 3(h). The power ratio of the core foci exhibits a right-skewed distribution, with a modal power ratio of 0.4. The variability of the focal power ratio is partially attributed to the use of a copropagating reference for TM measurement, as discussed in Appendix A.

The results presented in this section demonstrate that the wavefront shaping system can form diffraction-limited foci at any position on the distal fiber facet, with all foci accompanied by a speckled optical background. The power ratio when focusing through the core was typically in the range 0.3–0.5. As this is less than 1, both the wavefront shaping-formed focus and background light are expected to contribute to the measured ultrasound signals.

B. Ultrasonic field mapping

To demonstrate high-resolution ultrasonic field mapping through MMF, experiments were performed in which the sensor was insonified with known ultrasonic fields, which were mapped by scanning an optical focus across the sensor.

1. Planar ultrasound transducer

First, the FPUS was insonified by a 3.5 MHz planar ultrasound transducer with a diameter of 50.8 mm, driven by a pulsed signal generator. This transducer was mounted such that its orientation could be adjusted, allowing the angle of the planar wavefront to be varied. To facilitate later sensitivity characterization, this transducer and signal generator had been previously calibrated, so that the pressure generated by the transducer was known (as detailed in Zhang *et al.* (2008) [4]).

The ultrasound field produced by this transducer was recorded across a ~ 5.4 -mm-diameter region of the FPUS. This involved raster scanning a focused interrogation beam across this region using wavefront shaping, with a step

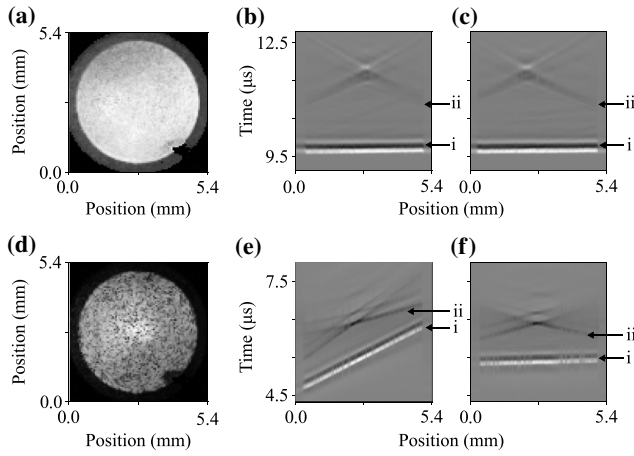


FIG. 4. Ultrasound field maps recorded using the FPUS interrogated through MMF. (a) Maximum intensity projection of the US field recorded when the sensor was insonified at normal incidence. Cross-sections through the recorded field along the (b) vertical and (c) horizontal axis of (a). (d)–(f) Maximum intensity projection and cross-sections as in (a)–(c), for the field recorded when the FPUS was insonified at an angle of $\sim 26^\circ$. The planar wavefront and subsequent edge waves are marked with (i) and (ii), respectively.

size of $60\ \mu\text{m}$. At each position, a time-resolved ultrasound signal was recorded, as described in Sec. II C. The full dataset provided a spatiotemporal map of the acoustic field. This procedure was repeated for multiple transducer orientations to capture wavefronts at different angles of incidence. The time taken to record each ultrasound field was approximately a minute, with this speed determined by the $\sim 100\ \text{Hz}$ refresh rate of the SLM.

Figure 4 presents two representative examples of recorded ultrasound field maps, each corresponding to a different angle of incidence of the planar wavefront. In Figs. 4(a)–4(c), the wavefront is normally incident on the FPUS. The maximum intensity projection [Fig. 4(a)] reveals two concentric regions of signal intensity, i.e., higher in the core and lower in the cladding, consistent with the fiber structure. Cross-sections [Figs. 4(b) and 4(c)] show a tripolar wavefront arriving at $\sim 9.5\ \mu\text{s}$, along with edge waves that converge near the center of the field of view and then diverge, matching the expected field pattern of the transducer.

Figures 4(d)–4(f) show the recorded ultrasound field when the planar wavefront is incident at an angle ($\sim 26^\circ$). The maximum intensity projection [Fig. 4(d)] again reveals concentric regions, with a stronger maximum signal intensity in the core. Cross-sections [Figs. 4(e) and 4(f)] display a tilted tripolar wavefront and associated edge waves, consistent with the angled insonification. Unlike the previous case, the signal intensity varies locally across the field of view. Given that the FPUS is insonified with a uniform ultrasound plane wave, this variation is likely due

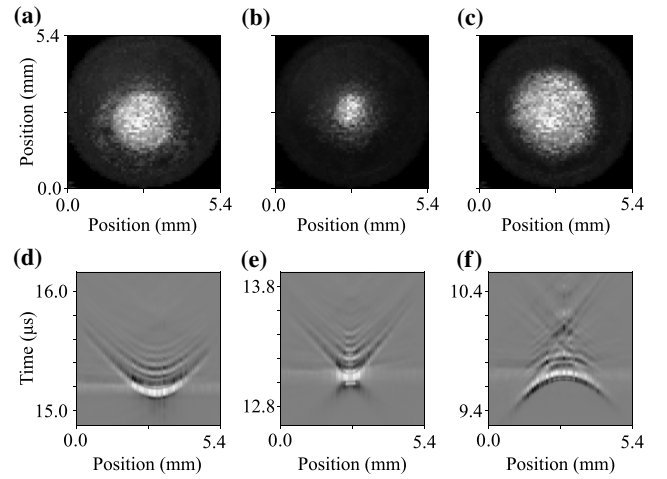


FIG. 5. Focused ultrasound field maps. (a),(c),(e) Maximum intensity projections and (b),(d),(f) cross-sections of the US fields recorded when the FPUS was insonified by a focused ultrasound transducer at three different transducer-sensor separation distances. The distances used were greater than (a),(b), approximately equal to (c),(d), and less than (e),(f) in relation to the transducer focal distance.

to variation in the sensitivity of the system between different interrogation beams. This variation, and its presence only under angled insonification, is explained by a spatial-averaging-based model of FPUS directivity, as discussed in Sec. IV.

2. Focused ultrasound transducer

To demonstrate field mapping of nonplanar ultrasound fields, the system was used to map the output of a high-frequency focused ultrasound transducer. The transducer had a center frequency of $30\ \text{MHz}$, a diameter of $12.7\ \text{mm}$, and a focal length of $f = 19.05\ \text{mm}$, and was mounted such that the distance between the FPUS and transducer could be varied. Again, the transducer was driven by a pulsed signal generator. The ultrasound field maps were recorded for different transducer-sensor separation distances in the same way as described previously, with the same spatial sampling density and field of view.

Three of these recorded ultrasound fields are shown in Fig. 5, with the sensor positioned in planes above, close to, and below the transducer focal plane. The maximum intensity projections of each of these ultrasound fields are shown in the top row of Fig. 5, with the corresponding field cross-sections shown in the bottom row. In each case, the cross-section is through the center of the region shown in the corresponding maximum intensity projection, along the vertically displayed axis.

The recorded field maps exhibit the expected features of a focused ultrasound beam. In the maximum intensity projections and cross-sections, weak spatial confinement is observed when the sensor is positioned above or below

the focal plane [Figs. 5(a), 5(b), 5(e), and 5(f)], while strong confinement is evident at the focal plane [Figs. 5(c) and 5(d)]. As anticipated, the wavefront curvature transitions from positive (with a leading positive-pressure peak) above the focus to negative (with a leading negative-pressure trough) below the focus.

In addition to these features, a weak planar wavefront is observed in each ultrasound field cross-section. The arrival time of this weak planar wavefront coincides with the most intense, shallow-angle portion of the curved wavefronts. This artifact arises from background light not confined to the wavefront shaping-generated focus, which is discussed in Sec. IV.

3. Sensitivity

To evaluate the sensitivity of the system, its noise equivalent pressure (NEP) was measured. The procedure for this NEP measurement follows that presented in Zhang *et al.* [4], using the same calibrated 3.5 MHz planar transducer used in the field maps shown in Fig. 4. The transducer was angled at 15° to the sensor, to prevent significant contribution from light outside the region of the focus for the recorded signals (confirmed by the absence of a strong planar artifact). This NEP measurement procedure was repeated across an array of ~ 3000 focus positions within the fiber core, and the modal RMS NEP was found to be 240 Pa over a 20 MHz bandwidth. This is slightly higher than values previously reported for planar FPUS [4]. A small reduction in sensitivity is expected, given that the focal power ratio is < 1 , which is consistent with this result.

IV. DISCUSSION

The primary outcome of this study is the demonstration that high-resolution, high-sensitivity ultrasound field mapping can be performed by interrogating an FPUS through MMF using wavefront shaping. In addition to this central result, several secondary findings and technical considerations warrant further discussion. These can be grouped into three main areas: (1) the directional response of the FPUS, interpreted through a spatial averaging model, (2) opportunities for improving wavefront shaping capability in terms of speed and focus quality, (3) practical design considerations for implementing this approach in real-world systems, and (4) the prospects for fully reconfigurable array read-out.

In terms of the directional response of the sensor, the results presented in Sec. III B showed that, for oblique ultrasound waves, sensitivity varied noticeably between interrogation positions while, for normally incident ultrasound waves, the sensitivity was relatively uniform. Additionally, weak planar artifacts were observed under certain conditions. Both effects can be explained using a simple model of FPUS sensitivity based on spatial averaging [22].

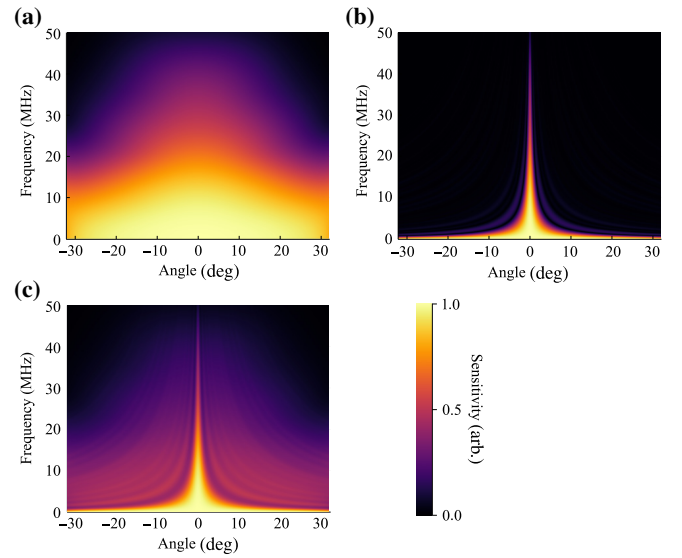


FIG. 6. Illustrative example of FPUS directivity profiles simulated using a spatial averaging model [22], under interrogation by (a) a 100- μm -diameter Gaussian beam, (b) a 5-mm-diameter speckle pattern, and (c) a beam composed of a sum of a 100- μm -diameter Gaussian beam and a 5-mm-diameter speckle pattern. In (c), each beam component carries 50% of the total optical power. The FPUS used in these simulations had a 40- μm -thick spacer layer, equivalent to a 20- μm -thick spacer with a glass backing [23]. Interrogation beam patterns were generated by modeling light propagation through a 200 μm core diameter, an NA of 0.2, and the MMF used in Jolab [24].

In short, this model states that the signal recorded by an FPUS is the weighted average of the ultrasound pressure across the illuminated region, with weights determined by the local optical intensity. As a result, large interrogation beams are primarily sensitive to ultrasound waves near normal incidence since contributions from oblique angles self-cancel. In contrast, small interrogation beams remain sensitive to ultrasound waves over a broad angular range. This behavior is illustrated in Figs. 6(a) and 6(b), which show simulated directional- and frequency-dependent sensitivity profiles for an FPUS interrogated using 100- μm - and 5-mm-diameter Gaussian beams, respectively.

Interrogating an FPUS through MMF using wavefront shaping introduces additional complexity due to the beam's dual structure: a tightly focused core and a broader speckled background. The focused component provides broad angular sensitivity while the background, illuminating a larger area, has a narrower directional response. Together, they yield a composite sensitivity profile: high near normal incidence (where both components contribute) and moderate over a wider angular range (where only the focus contributes). Figure 6(c) illustrates an example of this hybrid directivity. In this example, the response of an FPUS interrogated by a 100- μm -Gaussian focus and

a 5-mm-speckled background was simulated, with each component contributing 50% of the beam power.

This interpretation of the directional response of an FPUS is consistent with the results in Sec. III B. At normal incidence, signal amplitude scales with total beam power, producing a relatively uniform maximum intensity projection [Fig. 4(a)]. At oblique angles, only the focused component contributes significantly, and variations in focal power across the field lead to a nonuniform maximum intensity projection [Fig. 4(d)]. This hybrid directivity also explains the weak planar artifacts, as seen in Fig. 5, which result from overestimation of the signal amplitude from normally incident wavefronts. In some applications with complex ultrasonic wavefronts, e.g., *in vivo* photoacoustic tomography, these artifacts may be negligible. In other applications, e.g., ultrasound field mapping for calibration purposes, these artifacts are not negligible and will require mitigation strategies. These could include an improved power ratio (ideally very close to 1) or applying corrections to recorded fields based on the modeled or measured directivity of the system [25].

Second, while the wavefront shaping system enabled practical high-resolution field mapping, there remains room for improvement in both focus power ratio and calibration speed. Two straightforward modifications, i.e., replacing the internal reference with an external reference and controlling the polarization of the light coupled into and out of the fiber, could improve the focus power ratio. The choice of reference and the resulting system trade-offs are discussed in Appendix A. Using these methods, power ratios as high as 0.96 have been reported [26]. In this system, this would translate to a sensitivity improvement of $\sim 2\times$ and would reduce the power contained in the background light by $\sim 15\times$, effectively eliminating any planar artifacts.

In terms of calibration speed, the TM-based approach to wavefront shaping enabled parallelized measurement of all the required wavefronts in ~ 40 min, which represents a reduction in calibration time for practical imaging of $>1000\times$ compared with prior work using a genetic algorithm. The primary remaining limitation on the calibration time was the slow read-out speed of the camera (~ 10 Hz), followed by the refresh rate of the SLM (~ 100 Hz). Substituting a high-speed camera and using a >20 kHz amplitude modulating digital micromirror device in an off-axis configuration [27–29] for wavefront control could reduce the calibration time to under a second. Alternatively, recently developed fast MEMS-based phase modulators may be able to achieve these calibration speeds without the loss of optical efficiency imposed by amplitude-only modulators [30], and compressive sensing methods [31] may allow further acceleration.

The time taken to map US fields was substantially faster than the calibration time, only limited by the SLM refresh rate, with a ~ 5.4 -mm-diameter field of view

mapped in approximately a minute using a step size of $60\text{ }\mu\text{m}$. However, this imaging time could still be significantly improved. Imaging speed could be straightforwardly enhanced by increasing the sampling step size, at the cost of resolution. Using a ~ 30 kHz digital micromirror device for beam control would significantly reduce imaging times, enabling <100 ms mapping of centimeter-scale fields of view for video-rate imaging. Compressive methods [32] could further reduce this imaging time or enable fast mapping of larger fields of view at higher resolution.

Third, three practical features of the prototype imaging system used warrant discussion. First, the detection photodiode was located on the distal side of the fiber. This is suitable for some applications (e.g., handheld imaging) but not for others (e.g., narrow endoscopes). Moving the detection photodiode to the proximal side of the fiber would require suppression of the parasitic interferometer formed by fiber end facets, via angle-polishing or antireflection coatings. Second, the FPUS required uniform thickness due to the need to select a single bias wavelength for TM calibration. This requirement could be relaxed by adopting broadband TM calibration methods [33,34]. Third, the MMF used in this work needed to remain static in a single bending configuration for TM calibration and field mapping. The reason for this, as in other MMF-based systems, is that the TM of the fiber is highly bend-sensitive. For rigid imaging probes (e.g., needle encased endoscopes), this is acceptable but, for a fully flexible imaging system, a method of bending compensation would need to be incorporated. Several promising approaches have been reported in recent years. One class of method relies on the measurement of a library of TMs in different bending states and then determining the appropriate TM to use dynamically during use (e.g., via a virtual guide star or directly from image contrast) [35,36]. Others include methods based on predicting the necessary wavefront corrections using known fiber geometry [37], dynamic recalibration of the fiber TM using guide stars and fiber correlations [38], and the use of graded index fibers more tolerant to bending [39]. Future work could explore which is most practical for FPUS interrogation.

Finally, the use of wavefront shaping to interrogate an FPUS through an MMF introduces the potential for software-defined, reconfigurable sensor read-out, with dynamic control over the interrogation beam. In the current implementation, a copropagating reference is used for transmission matrix measurement, which limits beam shaping to focused beams (or multifocal patterns). This is due to unknown phase offsets introduced by the spatially varying reference, which are constant across each row of the measured transmission matrix but differ between rows. While sufficient for high-resolution mapping, this prevents fully arbitrary beam shaping. Switching to an external reference would remove these offsets, enabling fully reconfigurable sensor interrogation. Straightforward

applications could include switching between linear and matrix array configurations using line beams and point foci, respectively. Structured interrogation could also be used to adjust the trade-off between FPUS sensitivity and directivity [22,40,41] by altering the interrogation spot size or structure, or to eliminate the need for expensive tunable interrogation lasers via angle-based bias control [42].

V. CONCLUSIONS

In summary, an ultrasound field mapping system based on interrogating an FPUS through MMF was presented, in which a focus beam is scanned across the sensor using wavefront shaping-calibrated wavefronts.

Experiments mapping known ultrasound sources demonstrated that this method is capable of high-resolution ultrasound field mapping through an optical fiber, with only a small sensitivity penalty relative to conventional FPUS-based systems (due to imperfect optical focusing). To interpret the system's directional response, we introduced an explanation based on spatial averaging [22], which offers a framework for understanding and correcting observed artifacts. Validating and refining this model could form an avenue for future work.

By enabling high-resolution ultrasound field mapping through a single, narrow multimode optical fiber, this work lays the foundation for a new class of high-performance, compact, and versatile imaging devices. These systems would combine the advantages of high-performance FPUS-based platforms, such as high sensitivity, broad bandwidth, fine spatial sampling, and large-scale detection arrays, with the compactness and flexibility of piezoelectric devices. In addition, the ability to dynamically adjust the sensing configuration could allow a single device to meet the requirements of several imaging applications. Modified versions of the system presented here, with the FPUS mounted directly on the fiber tip, could also achieve extraordinary sampling resolution ($<1\ \mu\text{m}$) for applications at high acoustic frequency ($>100\ \text{MHz}$). As such, devices based on this work could unlock a wide range of new applications, from industrial inspection in confined spaces to minimally invasive medical imaging in the gastrointestinal or vascular systems or handheld, high-resolution and MRI-compatible clinical photoacoustic and ultrasonic imaging.

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DATA AVAILABILITY

The data are available from the authors upon reasonable request.

APPENDIX A: WAVEFRONT SHAPING PHASE REFERENCE

As described in [Sec. III A](#), the quality of optical focusing through the multimode fiber varied across the sensor surface, with implications for ultrasound sensitivity. The expectation was that the copropagating reference was at least partially responsible for this variability: since the reference beam is speckled, the poor signal-to-noise ratio in dark regions degrades phase recovery.

To explore this fact, experiments were performed in which the fiber TM was measured using different choices of copropagating reference, and spatial variation in focusing performance was measured. This entailed repeating the TM measurement procedure from [Sec. II B](#) with reference beams generated by coupling a beam into different positions on the proximal fiber facet. Focusing performance with each TM was then evaluated by scanning a focus across a grid of positions on a camera positioned on the distal side of the fiber (as in [Fig. 2](#)) and recording images. To ensure a full sampling of the positions on the distal facet of the fiber, the grid pitch was $30\ \mu\text{m}$ for the camera, equivalent to $1.2\ \mu\text{m}$ for the distal fiber facet (less than the diffraction limit).

Three illustrative maximum intensity projections, from the images recorded via three different TMs, are shown in [Fig. 7](#). The relevant TMs were recorded with references generated by beams focused near the fiber core, at approximately two-thirds of the distance between fiber core and cladding and near the boundary of the fiber cladding in the projections shown in (a), (b), and (c). Each pixel in these maximum intensity projections provides the highest intensity that can be achieved when attempting to focus at the corresponding position, essentially giving a map of focusing performance.

The focusing performance map in each case has a structure similar to that of a speckle pattern. Different reference positions provide various performance maps, with the structure of each performance map matching the expected structure of the speckle produced by a focused beam coupled into the fiber at the relevant position. For example, the map shown in [Fig. 7\(a\)](#) shows the larger speckle grain size and radial structure expected for a speckle pattern produced by coupling a focused beam into the center of the fiber.

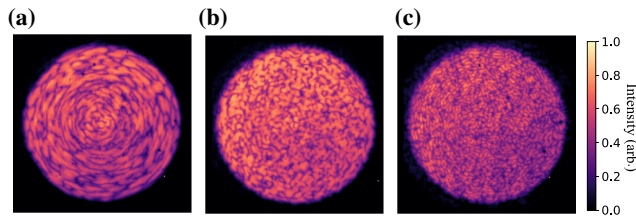


FIG. 7. Spatial maps of the maximum focusing performance achieved when using different copropagating references for the TM measurement. Reference beams were generated by coupling a focused beam into the fiber at positions (a) at the center of the fiber core, (b) two-thirds of the way between the center and the edge of the fiber core, and (c) near the edge of the fiber core.

These results confirm that the spatial variation in the intensity of the copropagating reference did substantially contribute to variable wavefront shaping performance. This result points to a trade-off in system design: compared with an external phase reference, a copropagating reference significantly simplifies the optical setup, at the cost of more variable focusing performance and the inability to create arbitrary structured beams. An alternative would be to make multiple TM measurements using different copropagating references, improving the focusing performance and maintaining the simpler optical setup at the cost of increased numbers of measurements [43]. These results also show that it is important to appropriately choose the copropagating reference. For example, the reference beam generated by coupling light close to the center of the fiber gave notably more regions of poor focusing. Finally, it may be possible to mitigate the effect of variable focusing performance on ultrasound field mapping without increasing optical complexity by using a nonuniform array of sensing positions chosen to avoid areas of poor focusing performance.

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