Widefield lensless endoscopy with a multicore fiber

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Received 12 August 2016; accepted 11 September 2016; posted 19 September 2016 (Doc. ID 273543); published 11 October 2016

We demonstrate pixelation-free real-time widefield endoscopic imaging through an aperiodic multicore fiber (MCF) without any distal opto-mechanical elements or proximal scanners. Exploiting the memory effect in MCFs, the images in our system are directly obtained without any post-processing using a static wavefront correction obtained from a single calibration procedure. Our approach allows for video-rate 3D widefield imaging of incoherently illuminated objects with imaging speed not limited by the wavefront-shaping device refresh rate.

Fiber-optic microendoscopes allow for minimally invasive high-resolution imaging deep within living organisms. Over the last decades they have continued to become more versatile with the miniaturization of fiber-based devices and multimodal imaging capabilities. A new class of such devices, fiber-based lensless endoscopes, operating without any distal optical or mechanical elements, enabled extreme miniaturization of the probe dimensions down to a few hundred micrometers, permitting minimally invasive imaging [1–6]. Image formation in the lensless endoscopes, capable of producing focal planes at various distances from the fiber tip, relies on either raster scanning or widefield modalities. To reach real-time image acquisition rates, these systems require in the former case ultrafast devices capable of wavefront shaping (typically deformable mirrors) or beam scanning [7], or in the latter case real-time computation [2,8].

Here we show that, using a slow wavefront-shaping device (spatial light modulator, SLM) and an MCF with weakly coupled cores, achieving conventional widefield imaging in real time using a single calibration procedure is straightforward. Building on the framework developed for imaging through scattering media [9] and relying on the practically infinite optical memory effect in such MCFs [10,11], we demonstrate widefield imaging of incoherently illuminated objects.

A conceptual illustration for our technique with the corresponding numerical simulation is depicted in Fig. 1. The schematic in Fig. 1(a) represents a conventional widefield microscope where light, scattered from an incoherently illuminated object, gets collected by a 4f system, thus forming an image on the other side of it. For such an imaging system with a lateral intensity point spread function $PSF(\tilde{r})$, image intensity distribution $I(\tilde{r})$ is related to the object $O(\tilde{r})$ through a convolution operation: $I(\tilde{r}) = O(\tilde{r}) * PSF(\tilde{r})$.

Typically, an MCF acts as an imaging conduit, transmitting object information from one fiber endface to another. In earlier implementations, the individual cores sample the object directly [12,13], giving rise to two important restrictions: (i) pixelation due to inter-core separation and (ii) imaging restricted to a fiber facet itself, putting the probe in contact with the sample. The ability to operate at flexible working distances is highly desirable in endoscopic applications. In our previous works [3,7], we reported MCF devices that met both of the mentioned requirements. The combination of a wavefront-shaping device and an MCF can effectively function analogously to the common 4f system, a major difference being its point spread function (PSF), which exhibits significant side lobes. They arise from the fact that the pupil of an MCF is segmented, and such discontinuities give rise to prominent side lobes [14]. This in turn affects the image, transmitted through the system, as illustrated in Fig. 1(b), resulting in ghost images of the object. Recently we showed [10] that the randomness in the MCF core positions can greatly reduce side-lobe intensity in the PSF of such an imaging system. A simulation of an imaging experiment with such an aperiodic MCF is shown in Fig. 1(b), in which the PSF is calculated given the cores’ distribution in the real fiber used throughout this work. While the images exhibit replicas of the object, their intensity is at least 2.5 times lower compared to the central image. Commercial fiber bundles do exhibit variations in core geometry and spacing in view of decreasing the inter-core coupling [15]; hence, we expect a larger reduction in the side-lobe intensity [11].

Implementations of the MCF with wavefront shaping, reported so far, have all used raster scanning for imaging [3,7,11,16]. Unlike in endoscopes based on multimode fibers...
(MMFs), this becomes trivial in MCFs due to their very large memory effect [10]. As there is little or no cross-talk between fiber cores, the transmission matrix of an MCF is practically diagonal. Hence, any phase gradient at the proximal end is preserved during light propagation through the fiber. This phenomenon has been used to remotely scan the beam by applying a global tip-tilt on the composite wavefront entering the MCF. Since these are relatively simple patterns, the SLM can now be decoupled from having to perform the scanning, and conventional galvanometer-based scanners were used allowing imaging rates of 11 fps [7].

In this Letter, we propose and demonstrate the extension of the memory effect of MCFs to the widefield imaging framework. The reasoning is relatively straightforward, since a translation of a point source in the object plane would result in a phase gradient, which is preserved through the fiber (linear in case of the transverse plane and quadratic for the axial plane). This results in an accurate mapping of any object shift to the image. These concepts are reminiscent of earlier experiments in scattering media [9,17], and the additional key advantage of the MCF is that the memory effect does not limit the imaging process.

We focus on the experimental evaluation of such widefield incoherent imaging using the memory effect to enhance speed and simplicity. The design and the original fabrication approach of this fiber were previously reported in [10]. Figure 2(c) depicts the fiber fabricated with the following parameters: individual core diameter \( d_0 = 3.2 \, \mu m \), its numerical aperture \( NA = 0.18 \), master triangular lattice pitch \( \Lambda = 20 \, \mu m \), and the randomness parameter \( PR \approx 0.22 \) (see [10] for details). The outer diameter of the probe is about \( 360 \, \mu m \) [Fig. 2(c)], and the length of the fiber used in this experiment is 40 cm, ensuring no considerable inter-core group delay dispersion [18].

A simplified scheme of the experimental setup, used for the widefield imaging with this MCF, is shown in Figs. 2(a) and 2(b).

The laser beam from a femtosecond (fs) source (Amplitude Systèmes t-Pulse, \( \lambda = 1030 \, nm \), 180 fs, 50 MHz) is extended with a telescope (\( L_1, L_2 \)) to overfill the aperture of the 2D liquid crystal SLM (Hamamatsu X10468-07). The latter is used to shape the segmented wavefront entering the MCF via its proximal endface. During the initial system calibration step, a micro-lens segment inscribed on the SLM for each fiber core produces a focal spot at the focal length \( f_{ml} \) from the SLM face and is spatially scanned around its initial position to optimize the coupling into the corresponding core. The array of the optimized focal spots is then imaged onto the MCF proximal end via a system of lenses (\( L_3, MO_1 \)). A focal plane is created at a distance \( z = 600 \, \mu m \) away from the MCF distal end and imaged via another telescope system (\( MO_2, L_4 \)) onto a camera (CCD) for calibration and testing. Output polarization from different cores of such non-polarization-maintaining MCF is arbitrary [19]; therefore, we employ a linear polarizer (P, Thorlabs LPNIR100) to discard any concomitant effects. After the initial system calibration and compensation of the distal wavefront for the intrinsic MCF phase distortion, one obtains a characteristic PSF [Fig. 2(d)] comprising a central spot surrounded by six dimmer replicas distributed on a circumference with \( r \approx 37 \, \mu m \). In the linear imaging regime the brightest speckle contrast relative to the central peak \( I_{BS}/I_C \) is 0.4, measured in the field of view (FoV) center [Fig. 2(d)]. We verify the PSF variation across the FoV, and in Fig. 2(e) we summarize such measurements for off-axial points, showing the variation of PSF FWHM less than 1 \( \mu m \) and \( I_{BS} \approx 0.6I_C \) on the FoV edge.
should not drastically decrease the imaging performance of the widefield technique.

Using this fiber, we perform a series of proof of concept experiments, described in the following. The experimental setup, used for the fiber calibration [Fig. 2(a)], features two conjugate planes: object plane (OP) and image plane (IP), where a particular distance \( z \) for the IP can be flexibly chosen during the calibration step. Unlike the calibration step, where it is required to have a spatially and temporally coherent source, the following imaging experiments are performed with spatially incoherent illumination of the object. (Nevertheless, the bandwidth of the illumination source has to be smaller than the speckle spectral correlation width.) We now perform an incoherent projection of an amplitude mask from the OP to the IP; we denote such operation as forward projection. The related experiment is performed with a reflective object from the United States Air Force (USAF) resolution chart [Fig. 3(a)]. The phase mask on the 2D-SLM is the same used to correct for the intrinsic MCF phase distortion and does not change throughout the experiment, unless one switches the projection to another working plane (different \( z \)). A number 5 target (object height \( \approx 39 \mu m \)) was placed in the OP and illuminated incoherently by placing a rotating diffuser \([D \text{ in Fig. 2(a)}]\) between the object plane and the laser source. The position of the diffuser along the beam propagation direction is chosen to create in the OP a sufficiently large illumination area, slightly exceeding the dimensions of the used mask. The measured projection in the IP is shown in Fig. 3(b); it is clear that the setup performs like a conventional 4\( f \) imaging system with only a single calibration aided by the memory effect. As the SLM does not need to be updated further, we can easily perform high-speed image acquisition. (Exposure time for the presented example was 30 ms.) For a real-time imaging experiment of a moving target, see Visualization 1.

Next, we perform a widefield imaging experiment in the epi direction, using the modified setup shown in Fig. 2(b). In this configuration the spatially incoherent illumination (the same as for the forward projection experiment) reaches the sample plane (OP) after passing the MO\(_2\), and the IP is matched directly to the CCD\(_2\) camera plane. (Note that the physical locations of IP and OP are switched compared to those in the forward projection experiment). A linear polarizer (P) is used after the MCF proximal end in the same scope as in the forward projection setup, as we measure only the scalar transmission matrix [20]. We use two types of USAF targets—positive and negative—to compare the operation of our imaging system in different sample configurations, particularly for the influence of the side lobes. In the first case we use a positive target consisting of the number 6 [Fig. 3(c)]; object height is \( \approx 23.5 \mu m \), which is almost 3 times less compared to the distance between the side lobes of the system PSF. This results in an image on a relatively homogeneous background [Fig. 3(d)] without any overlying ghost replicas. In the case of a negative target [Fig. 3(e)], given the object height \( \approx 39 \mu m \) and the distance between replicas, the object convolution with the system PSF results in ghost images of the object that begin to overlap with the central (brightest) 2. Unlike in two-photon excited fluorescence (TPEF) microscopy [10] where this weak background is screened by the inherent nonlinearity and does not contribute to image formation, these now result in ghost images albeit with reduced intensity. As it can be seen from Fig. 3(f), the image associated with the central lobe is still the brightest one and can be easily distinguished from the background generated by the side lobes and any weak speckle. We note that earlier results from our group on the same MCF achieved a FoV of 120 \( \mu m \) (0.064\( \pi \)), whereas in the present work the illumination area was restricted to cover a FoV of \( \approx 50 \mu m \) (0.03\( \pi \)) due to lower experimental SNR.

Another advantage of the direct imaging approach with an SLM over speckle-correlation-based techniques [8] for widefield imaging is that our approach offers a degree of optical sectioning due to its 3D transfer function [21]. We performed an experimental measurement of the depth of field (DoF) in the following manner. After distortion compensation, the calculated phase pattern is displayed on the SLM to obtain a focal plane at \( z = 600 \mu m \) from the MCF distal end. Next, a diffraction-limited point source is placed at different \( z \) and the DoF is evaluated from the stack of images measured at CCD\(_2\), cf. Fig. 2(b). This results in a Gaussian distribution [Fig. 4(f)] with FWHM = 46 \( \mu m \), which is in qualitative agreement with the expected DoF (32 \( \mu m \)). Then, using the same experimental layout as in Fig. 2(b) and the same test object as in Fig. 3(e), placed into the OP at \( z = 600 \mu m \) away from the MCF distal tip, we show in- and out-of-focus images of the target [Fig. 4(a)] by switching the focal planes through displaying on the SLM the respective differential phase patterns with no mechanical translation of the sample or the fiber. As expected, the object appears in focus and then is defocused on the CCD\(_2\) plane (IP). For the full stack of 12 imaged focal planes between \( z = 500 \) and 700 \( \mu m \), see Visualization 2.

Considering the measured DoF, we further demonstrate proof of concept optical sectioning experiments [Fig. 4(a)] with a 3D phantom, schematically depicted in Fig. 4(e). It consists of a 50-nm-thick layer of gold flakes, deposited onto a 130 \( \mu m \)-thick glass cover slip (number 1). From the bottom side of the latter, we stack a positive USAF target (Thorlabs R1DS1P) such that the separation between two metal layers equals the cover slip thickness. Therefore, when introduced into widefield imaging setup, the gold flakes layer is at \( z_1 = 600 \mu m \) away from the MCF distal end, whereas the USAF target layer is at \( z_2 = 730 \mu m \) [Fig. 4(e)]. By adjusting the phase mask on the SLM to translate the endoscope focal plane along the \( z \) axis, we could
acquire clear images at several \( z \) without moving either the fiber probe or the sample. Images of the two planes of interest, with the gold flake [Fig. 4(b)] and with USAF target group 7, are shown with the overlaying white-dashed lines indicating the outline and the position of the gold flake. Remarkably, we can focus on both these planes remotely using the SLM, demonstrating pixelation-free and 3D-resolved imaging. For the full stack of eight focal planes, imaged between \( z \approx 600 \) and \( 730 \mu m \), see Visualization 3. We believe this is the first demonstration of widefield 3D-resolved imaging at multiple depths in lensless endoscopy using linear contrast, and we expect this to be valuable in imaging 3D structures.

In the current implementation, the presented widefield lensless endoscope is not resilient to fiber bending and thus would change the phase dispersion within the MCF. However, solutions under development may eventually permit real-time compensation for these effects, see, e.g., [22] and references therein.

We have demonstrated real-time pixelation-free widefield imaging through an aperiodic MCF using the optical memory effect without any distal opto-mechanical elements and requiring no further post-processing. Both the forward projection of an amplitude mask from the proximal to distal side of the endoscopic system and epi widefield imaging were shown. Using wavefront control at the fiber proximal end, we showed that our system is capable of optical sectioning—producing clear images of different focal planes within a quasi-3D sample and doing so without physically displacing either the MCF or the sample. In future, we expect that widefield imaging could be combined with nonlinear imaging, e.g., TPEF imaging in a lensless endoscope [7]. Fluorescence sources or another broadband illumination whose bandwidth is narrower than the MCF’s speckle spectral correlation bandwidth could also be used without significantly affecting the imaging performance (see [8] for details). Considering the robustness and fabrication simplicity of the presented aperiodic MCF with low inter-core coupling as well as facilitated ultrashort pulse delivery and the small effective probe diameter (an order of magnitude smaller compared to the existing scanning endoscope solutions), such a fiber emerges as a promising candidate for miniaturized imaging systems.

**Funding.** Agence Nationale de la Recherche (ANR) (ANR-11-INSB-0006, ANR-10-INSB-04-01, ANR-14-CE17-0004-01); Aix-Marseille Université (AMU) (ANR-11-IDEX-0001-02); University Lille 1—Sciences et Technologies (USTL) (ANR-11-LABX-0007, ANR-11-EQPX-0017, CPER P4S Région Nord Pas-de-Calais); Institut National de la Santé et de la Recherche Médicale (Inserm) (PC201508); SATT Sud-Est GDC Lensless Endoscope; European Research Council (ERC) (677909); Azrieli Foundation; CNRS/Weizmann NaBi European Associated Laboratory.

**Acknowledgment.** The authors thank Juan de Torres for providing the sample of gold flakes.

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