Ultra-miniaturized Endoscopes with Multi-Core Fibers

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Abstract: We take stock of the progress made into developing fiber-optic ultra-thin endoscopes assisted by wave front shaping. We focus on multi-core fiber-based lensless endoscopes intended for multi-photon imaging. We put the work into perspective and outline remaining challenges.

1. Concepts of Lensless Endoscopes

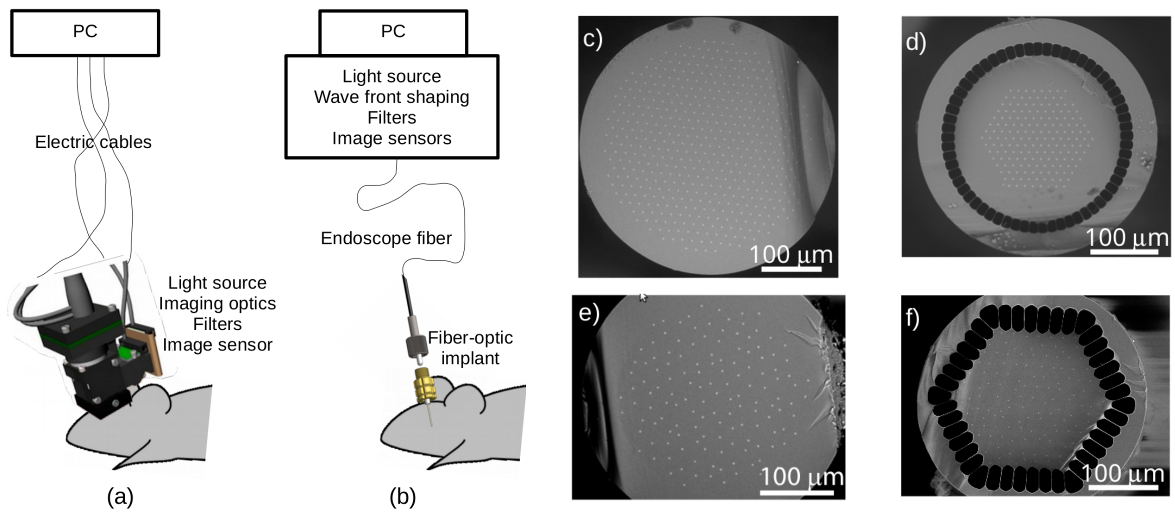

Fig. 1. (a) State-of-the-art: Miniaturized microscope. (b) Concept of the lensless endoscope. (c)-(f) Examples of MCFs developed for lensless endoscopy. (c) MCF with low cross-talk and single-mode cores [1]. (d) MCF with low cross-talk, single-mode cores, and a multi-mode inner cladding [2, 3]. (e) MCF with low cross-talk and aperiodically arranged single-mode cores [4]. (f) MCF with low cross-talk, aperiodically arranged cores, and a multi-mode inner cladding.

Microscopic imaging methods have become essential tools for biomedical research thanks to the fact that they can combine cellular-level spatial resolution, temporal resolution, and – together with genetic markers – cell-type identification, activity measurement, or stimulation. Following cells in their natural environment i.e. inside the tissue of living animals is most often infeasible with bench-top microscopes and require miniaturized imaging tools that can be fixed onto freely-moving animals. The state-of-the-art is the miniaturized microscope which consists of a head-mountable device containing all light sources, imaging optics, filters, and image sensor cf Fig. 1(a). A more recent approach is the lensless endoscope the philosophy of which is to be able to sense the image of an object located at one end of a long optical fiber by only having access to the other end cf Fig. 1(b). Compared to the state-of-the-art this approach should allow for a smaller head-mountable device with diameter down to around 0.1 mm cf Figs. 1(c)-1(f) and allow new imaging methods (like nonlinear or two-photon imaging) because light sources, imaging optics, filters, and image sensor can be remote rather than in the head-mountable device. To succeed in implementing a lensless endoscope the optical fiber must be considered as a part of the imaging system, and wave front shaping methods using for example spatial-light modulators must be employed to compensate the distortion of spatial information content that takes place in the fiber.

2. Merits of Multi-Core Fiber

The majority of initial reports of lensless endoscopes used multi-mode fiber (MMF) and have demonstrated high-resolution fluorescence imaging from within living tissue [5–8]. However multi-core fiber (MCF) can also be used [9], as can any wave guide with spatial degrees of freedom. MCFs have certain merits compared to MMF which we will describe in the following. We have fabricated several MCFs designed to enhance these merits [Figs. 1(c)-1(f)].

2.1 Low Cross-Talk between Cores

The multi-core fibers (MCFs) that we have designed and fabricated for our studies have all been optimized to exhibit minimal cross-talk between cores in order to assure that light injected into one core remains confined to only that core during propagation through the MCF. Examples of cross-talk values are: [Fig. 1(c)] -30 dB/m and [Fig. 1(d)] -20 dB/m.

2.2 Memory Effect

Due to the low cross-talk between cores there is a strong correlation between the wave front injected into the MCF and the one exiting the MCF at the other end. This effect is often referred to as the 'memory effect' and permits very intuitive control over the MCF output wave front; a desired output wave front change is obtained simply by adding the same wave front change to the phase mask displayed on the spatial-light modulator located at the input end.

2.3 Dispersion

Modal dispersion is the phenomenon that different transverse modes of light travel at different velocities leading to a group delay spread. In our MCFs there is no modal dispersion *per se*, since all cores are single-mode. Even so a small group delay spread remains in MCFs because of small intrinsic imperfections. When we characterized the group delay spread of our MCFs we obtained [Fig. 1(c)] 2.0 ps/m (max-min) and [Fig. 1(d)] 2.0 ps/m (max-min). Importantly, it is technically possible to compensate the group delay spread in MCF because light stays confined in the same core along the entire fiber length. As for chromatic dispersion, the phenomenon that different colors travel at different velocities, light in all cores of a MCF experience the same chromatic dispersion which can then be pre-compensated globally for all cores by standard means.

2.4 Double Cladding

In true endoscopic imaging the fiber serves two purposes, first, it transport excitation light to the sample, second, it collects fluorescent signal from the sample and transports it in the reverse direction. This would seem to put MCFs at a disadvantage due to the low core fill factor which is around 3 % in the designs in Figs. 1(c), 1(e). Fortunately this apparent disadvantage can be effectively negated by adding a double cladding to the MCF cf Figs. 1(d), 1(f). In these two designs the single-mode cores are contained in an 'inner cladding' of silica surrounded by an 'outer cladding' of air holes. The inner cladding becomes a high-NA multi-mode structure, very efficient at collecting diffuse fluorescence.

3. Key Results

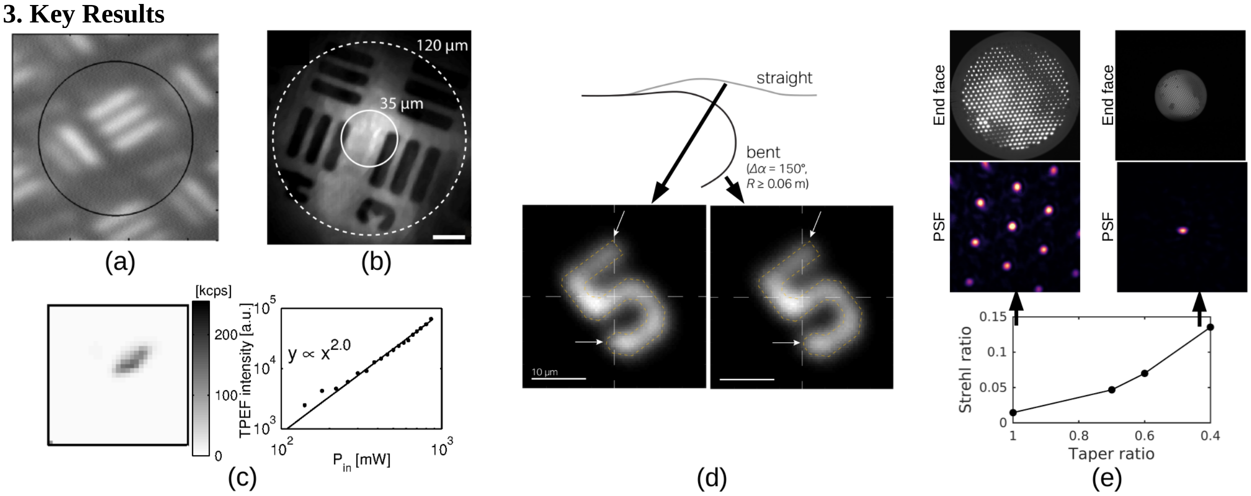

Fig. 2. Key results on lensless endoscopes. (a) Frame from a 12 fps frame rate video. Circle denotes the effective 45 µm-diameter effective field-of-view [1]. (b) Acquired image with extended field-of-view thanks to an aperiodically-tiled MCF [4]. (c) Two-photon point-scanning image of Rh6G crystal and verification of squared dependence of two-photon signal upon excitation power [2]. (d) Images acquired with conformationally-invariant MCF in straight and bent conformations, showing that the acquired image remains unaltered with changes in MCF conformation [10]. (e) Mode densification in down-tapered MCF. The Strehl ratio (fraction of total energy in central lobe of PSF) increases with decreasing taper ratio. Example images of end face and PSF for taper ratios 1 and 0.4. Taper ratio is defined as tapered to untapered diameter [unpublished].

3.1 Video-Rate Point-Scanning Imaging

The combination of spatial-light modulator and two-axis scan mirror may be employed in order to reach frame rates way beyond what can be reached when employing only a spatial-light modulator to perform the wave front shaping. Thanks to the memory effect of the MCF the phase mask on the (slow) spatial-light modulator can be chosen to be a static mask that focuses the MCF output to a point; the two-axis scan mirror then scans the point across the sample. By doing so and using the MCF from Fig. 1(c) we were able to acquire images at 12 fps cf Fig. 2(a).

3.2 Mode Scrambling

A downside of using MCFs with cores on a periodic, triangular lattice like Fig. 1(c) is the appearance of replica images cf Fig. 2(a). These can be avoided by resorting to various types of mode scrambling. One of these is to tile the cores of the MCF aperiodically as in Figs. 1(e), 1(f). Using such an aperiodic MCF in a lensless endoscope effectively eliminates replica images and extends the effective field-of-view as can be seen in Fig. 2(b) demonstrating imaging over a 120 μm-diameter field-of-view whereas the field-of-view would have been limited to 35 μm with periodically-tiled MCF.

3.3 Two-Photon Imaging

Acquisition of two-photon images is a challenge in lensless endoscopes. In Ref. [2] the required delivery of ultra-short pulses through the MCF to the sample was possible thanks to the low group delay spread, and the required collection of the weak, diffuse two-photon signal was possible thanks to the high-NA double cladding of the MCF in Fig. 1(d). Figure 2(c) presents a two-photon image of a single Rh6G crystal acquired in this way as well as a measurement of two-photon signal yield as the laser power was turned up, revealing a squared dependency that confirms the two-photon origin of the signal.

3.4 Impact of Fiber Bends and their Mitigation

Bends of the fiber have a detrimental impact on the function of lensless endoscopes if they occur while an image is being acquired, especially so in MMF where a bend impacts the guiding properties in a seemingly random way. In MCF – thanks to the low cross-talk between cores – the detrimental impact is limited to a translation of the image proportional to the relative angle of the MCF input and output ends [11]. We have demonstrated [10] how this effect may be mitigated by a helically-twisted MCF cf Fig. 2(d) where it can be seen that the acquired image is virtually static even as the MCF is bent from straight to a 150° bend.

3.5 Mode Densification

A downside to MCF as compared to MMF in the context of lensless endoscopes is their inferior mode density which limits the achievable spatial resolution in the acquired images. This can be remedied by different types of micro-structuring of a few cm of the MCF tip aimed at mode densification. One such example is given in Fig. 2(e) where the micro-structuring is a down-taper of the tip of the MCF (upper part of the Figure). This physically brings the modes closer together resulting in mode densification. Positive consequences are an increased fraction of total energy in the focus (Strehl ratio), increased spacing of replica images, as seen from the middle and lower part of Fig. 2(e).

4. Conclusion

MCFs possess a list of demonstrable advantages as the wave guide in a lensless endoscope, particularly so in lensless endoscopes aimed at two-photon imaging, or any other kind of nonlinear imaging. Future work should try to realize lensless endoscopes that simultaneously combine all the advantages.

5. References

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