Advances in imaging through a single optical fibre

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ABSTRACT

Traditional endoscopes relay images through a bundle of optical fibres, one fibre for each pixel in the image. However, in principle even one of these fibres carries enough spatial modes to relay an entire image, but modal dispersion means the resulting image is scrambled. As a solution to this, various groups world-wide are using aberration correction techniques to create a scanning spot at the exit of the fibre to raster-scan an object, the backscattered light from this spot is measured to give an image. The current limitation is that the aberration correction depends upon the bend of the fibre, meaning that once imaging is achieved, the fibre cannot be moved. We will present our latest work on high-speed aberration correction and choice of fibre, demonstrating that bends of several 10s of degrees can be achieved without degradation in image quality. Our work opens the route to fibre-based imaging systems being deployed in dynamic situations for various inspection and healthcare applications.

Keywords: Fibre-based imaging, aberration correction, endoscopy

1. INTRODUCTION

Endoscopes and similar instruments are based on bundles of multi-mode optical fibres. A small diameter lens is used to form an image of the object on the entrance facet of the fibres and each fibre relays the light from a single pixel of this image from the entrance facet to exit facet. At the exit facet a further lens relays this pixelated image to a detector array and an image of the object can then be recorded and visualized. However, in principle, even one multi-mode fibre supports enough spatial modes to relay the complete image, the problem being that each spatial mode travels along the fibre with a slightly different velocities such that their relative phase change results in the transmitted image being spatially scrambled.

Over 10 years ago Cizmar and coworkers[1], along with other groups[2], showed that by treating the fibre as a complex aberration and applying corrective beam shaping it was possible to spatially shape an incident laser beam to create a scanning spot at the exit facet, or far-field, of the single multimode fibre. The fibre aberration can be represented as a transmission matrix relating the complex amplitude of each output pixel to the linear, complex sum of each input pixel. The inverse of this matrix gives the recipe for the complex shape of the input beam required to produce a single bright spot at the output of the fibre or the far-field depending in which plane the matrix was measured. The input beam can be continuously re-shaped producing a scanning spot at the output of the fibre to raster scan the scene and measuring the corresponding backscattered light allows an image of the scene to be deduced. In principle this allows the creation of an endoscope type instrument the width of a single human hair. By using a pulsed laser, then in addition to the pixel intensity, it is possible to produce a 3D image of the scene by measuring the arrival time of the backscattered light[3]. The lateral angular resolution of the resulting image is given by the number of spatial modes supported by the fibre and imaging is typically sampled on a 60x60 grid. The depth resolution, typically a few mm, is set by the pulse duration and timing precision. The frame-rate of the imaging system is limited by the update rate of the spatial light modulator used for the beam shaping. Earlier embodiments of this single-fibre imaging approach relied upon the use of spatial light modulators based on liquid crystal to shape the incident beam thereby restricting the pixel rate to 100Hz resulting in single image durations of several seconds at best. More recently rather than using liquid crystal technology, the spatial light modulators have been based upon digital micro mirror devices, giving pixel rates upwards of 20kHz and multi-Hertz frame rates[4].

The main limitation of the single-fibre imaging approach is that the precise values of the transmission matrix depend upon the shape into which the fibre is bent, meaning that although it can be measured for any specific fibre bend, further bending of the fibre changes the matrix and degrades the image. Various approaches have been explored to model the matrix[5], repeatedly infer the matrix including from single ended measurement[6] or developing algorithms to correct for the degradation of the image as the fibre is moved to a different position[7].

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2. THE DEMONSTRATION SINGLE-FIBRE IMAGING SYSTEM

Building on this earlier work we have designed and built a compact technology demonstrator suitable to exploring various end-user applications, see figure 1. This demonstrator is based upon a DFB laser diode producing approximately 150 mW at 780 nm, a digital micro mirror device (DMD) acting as a programmable diffractive optic to shape the input beam to the fibre, relay optics to couple the shaped beam to the fibre. A second multimode fibre of 400 µm diameter to collect the back scattered light and a photomultiplier (PMT) to measure the intensity of the backscattered signal. The illumination and backscatter collection fibres are both embedded in an epoxy-filled, 20 gauge needle, which can be inserted into an object of interest.

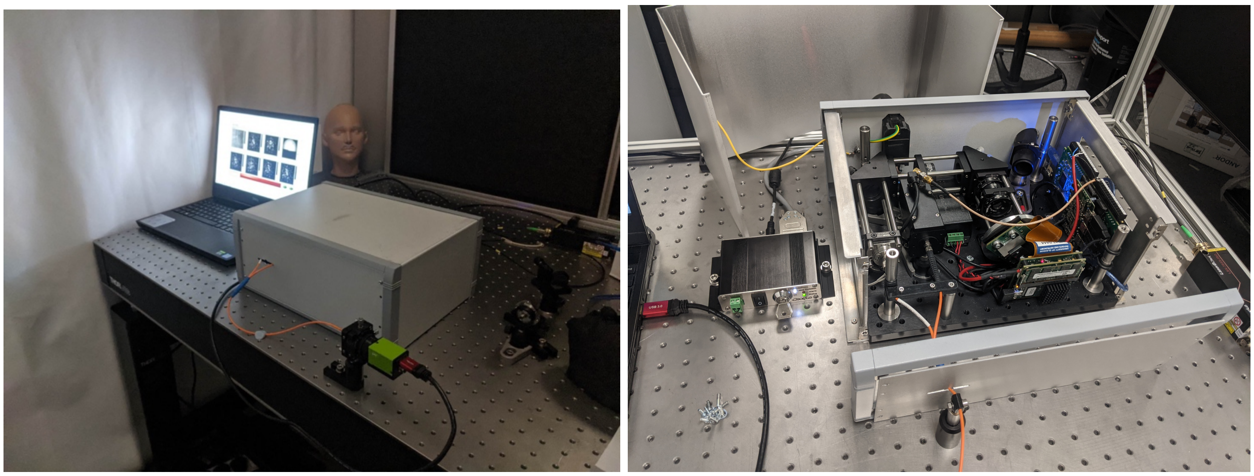

Figure 1. The single-fibre, endoscopic imaging demonstration system showing the main unit, calibration camera and control laptop (left) and with the top removed (right).

Measurement of the transmission matrix is based on recording the resulting interference pattern at the output of the fibre for a complete set of input modes. Most conveniently, this input set of modes are plane waves distributed in k-space covering the full numerical aperture of the multimode fibre. For a graded index fibre with a 62.5µm diameter core and a numerical aperture of 0.275, at a wavelength of 780nm, the number of supported modes is approximately 1200. Rather than the fibre modes themselves, we work in the plane-wave basis described by a regular grid in k-space. For each of these modes measuring the complex field of the resulting interference pattern requires the pattern to be interfered with four phase-stepped reference beams (which in our case comprises an arbitrary superposition of a sub-set of these plane waves). Allowing for bright and dark calibration images this requirement dictates 4 frames per mode, requiring nearly 8,000 images to be recorded. Our demonstrator includes a laptop computer which generates the required diffractive patterns to generate the plane-wave set, and reference beam, reads the resulting interference image from a compact high-speed camera, calculates the transmission matrix and hence the corresponding diffractive patterns required to produce a scanning spot at the end of the fibre. The laptop then reads the output of the PMT to display the live image of the scene.

To minimise the impact of bending the fibre we have focused on the choice of fibre type and the rapid measurement of the full transmission matrix and associated beam shaping algorithms. In the initial demonstrations the multimode fibre of choice had a step-index transverse profile where the resulting spatial modes have different, but in principle known, phase velocities which never-the-less results in a significant loss in contrast for the scanning spot as the fibre is bent. It is well understood that this intermodal dispersion can be dramatically reduced by switching to a graded index fibre, albeit the residual dispersion depends subtly on the manufacturing imperfections. These imperfections notwithstanding, we find that for typical fibre lengths <1 metre the resulting degradation in spot contrast and hence image quality is modest, see figure 2. Furthermore, by optimising the software to speed the data acquisition, calculation of the diffractive patterns we can record a new matrix and produce a scanning spot for any degree of fibre bend in under 100 seconds. The high-speed nature of this process means that we can acquire several transmission matrices for different degrees of fibre bend <10min and subsequently switch between them to account for movement of the fibre into new positions, see figure 3.

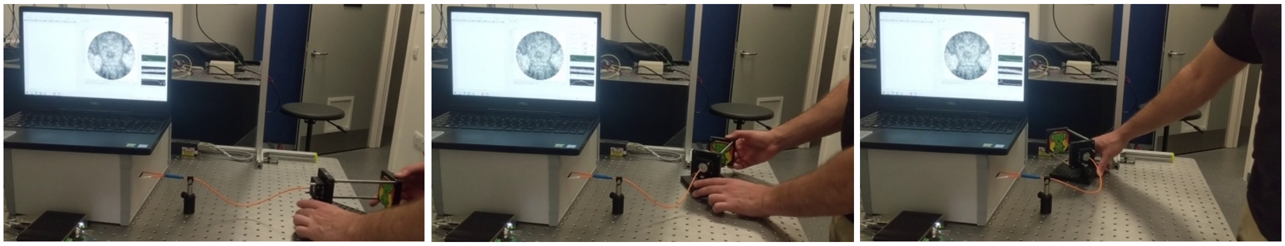

Figure 2. The single-fibre, endoscopic imaging demonstration system showing the degree to which the bend of the fibre can be increased without significant degradation in image quality.

The ability to pre-calibrate the single fibre imaging system to use in situations where the fibre is likely to be moved when imaging is taking place addresses one of the common criticisms of this approach and opens its use to various applications in inspection and healthcare.

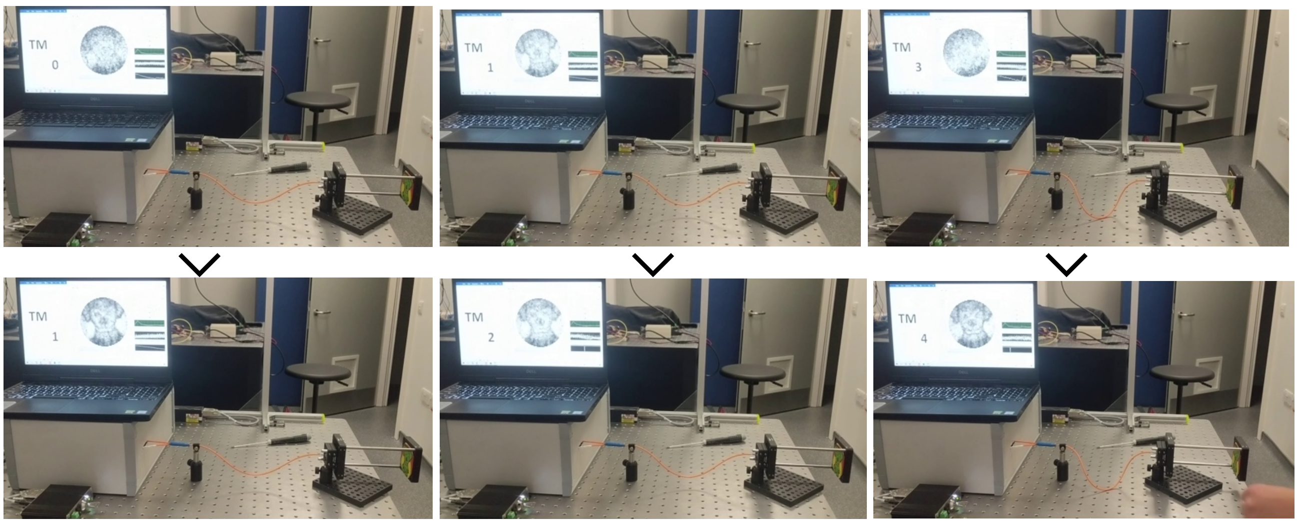

Figure 3. The single-fibre, endoscopic imaging demonstration system showing how the transmission matrix, from which the beam scanning diffractive elements are derived, can be switched (upper to lower) improving image quality for different degrees of fibre bend.

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