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Silica hollow core microstructured fibres for mid-infrared surgical applications

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Abstract

In this paper two silica hollow core microstructured fibres (Negative Curvature Fibre and Photonic Bandgap Fibre) are presented with attenuations of 0.06 dB/m and 1.1 dB/m at 2.94 μm wavelength, respectively. This is an important regime for medical applications, specifically surgery due to the existence of a strong absorption peak for water around 3 μm . The guidance of high energy pulses of the order of 195 mJ and 14.4 mJ respectively has been demonstrated. These energies are sufficient to ablate soft and hard biological tissue. As verification porcine bone was ablated in air and submerged in a water to simulate practical application of a surgical device. These fibres open the way to a new and fully flexible delivery system for high energy Er:YAG laser radiation.

Keywords

Photonic Bandgap Fibre; Negative Curvature Fibre; Fibres for medical applications; Microstructured fibre

Introduction

An Er:YAG laser ($\lambda=2.94 \mu\text{m}$) is particularly suitable for laser surgery because the water contained in human tissue strongly absorbs this radiation ($\alpha\sim 12000 \text{ cm}^{-1}$ [1]). If this laser beam is precisely delivered only to the areas that need cutting or ablating via surgical operation then damage to surrounding tissue can be minimised. Because of this surgical lasers are being increasingly used in medical fields like dentistry and optometry. Additional generic advantages of laser based procedures are that no pressure is applied reducing the pain for the patient e.g. dental drilling [2] and the cut geometry is not limited by the drill/scalpel geometry but is dictated by the focused spot size which generally can be significantly smaller than traditional surgical tools.

Currently the most common method of delivery of surgical lasers, for use in the operating theatre, is using articulated arms. These, although useful for delivering laser light to the patient, is perhaps less user friendly than a surgeon using a blade and there is significant restriction to movement. Therefore the benefits of using laser light for surgery are offset by the restriction to the surgeon's skill that the articulated arms can impose. Consequently, because of this need for truly flexible IR fibre delivery there have been extensive studies carried out with the aim of developing a practical system. A robust delivery fibre, as proposed here, would alleviate these problems and radically increase the usefulness of surgical lasers. Also, the flexibility and small physical size of a fibre system opens new fields for lasers in medicine like endoscopy and minimal invasive surgery.

Many groups have carried out research into fibres delivering laser radiation around 2.94 μm , which are mainly based on Chalcogenides [3, 4], GeO_2 [4], Fluoride [5] or Sapphire [6]. As shown appreciable powers could be delivered with those fibres, however to overcome the Laser Induced Damage Threshold (LIDT) these generally tend to be large core fibres. This leads to a multimode behaviour of the fibre and therefore to a high bend sensitivity, due to bend-induced mode coupling [7]. Also, some of these fibres contain toxic materials and compared to silica, are mechanically and chemically less robust; hence they are not ideal for medical applications.

One approach to overcome the limitations of the above mentioned fibres is to use a hollow-core fibre design. These fibres guide the laser radiation mainly by Bragg reflection, or by internal reflection at a dielectric coated metallic interface in the case of the leaky tube waveguide [7-9]. Due to the nature of the hollow-core the light is mainly guided through air/gas, and therefore the LIDT can be significantly higher than in solid core fibres [10-14].

In this paper two recently developed fibre designs are described based on the principles of hollow core microstructured fibres. Due to the shape of the central hollow core geometry, as shown in Figure 1a [15], the first fibre is referred to as a Negative Curvature Fibre (NCF). The second type is the Photonic Bandgap Fibre (PBF), which is also referred to as hollow-core Photonic Crystal Fibre (HC-PCF) as shown in Figure 1b [14].

Material and methods

The laser used in our experiments was an Impex High Tech ERB 15 laser. The operating wavelength is 2.937 μm and the pulse-length is 225 μs FWHM with an M^2 of ~ 2.5 at a repetition rate of ~ 15 Hz. The spatial profile of the laser has a donut shape.

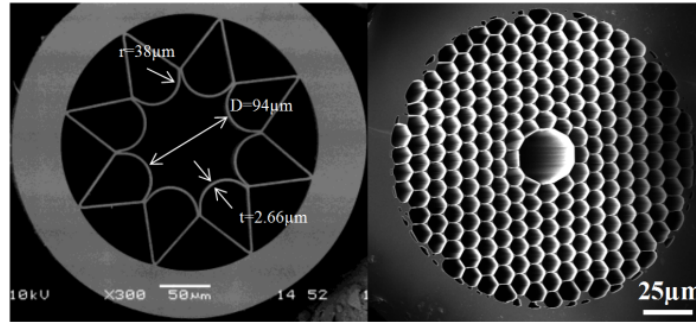


Figure 1: SEM pictures of the fibres used for this paper. a) Negative Curvature Fibre (NCF) b) Photonic Bandgap Fibre (PBF)

Both fibre types are fabricated from fused silica in a conventional stack and draw technique. The structure inside the NCF is formed from 8 stacked capillaries of circular cross section. However, these round capillaries become triangular during drawing as a direct effect of the pressure difference applied to the core and the interstitial ring structure. The material is Suprasil F300 which has a bulk attenuation of ~ 50 dB/m at $2.94 \mu\text{m}$ [16], however as light is mainly confined to the hollow core the influence of this high absorption is significantly reduced allowing low loss fibres to be made in this wavelength region [10].

The broad band guidance of the NCF has been previously described, with low attenuation ranges from $2 \mu\text{m}$ to $2.5 \mu\text{m}$ and from $2.8 \mu\text{m}$ to $3.8 \mu\text{m}$ [15] (Figure 2). The lowest attenuation achieved was 0.034 dB/m at $3.05 \mu\text{m}$ and 0.06 ± 0.01 dB/m at $2.94 \mu\text{m}$. A cutback measurement was carried out from 83 m to 3 m. The core size is $94 \mu\text{m}$, the core wall thickness $2.66 \mu\text{m}$ and the radius of curvature for the core wall is $38 \mu\text{m}$ (Figure 1a). The NA of the fibre was measured to be 0.03 .

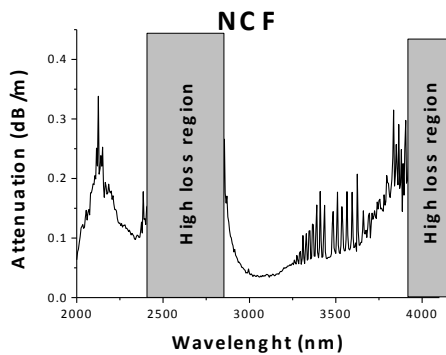


Figure 2: Attenuation of the Negative Curvature Fibre

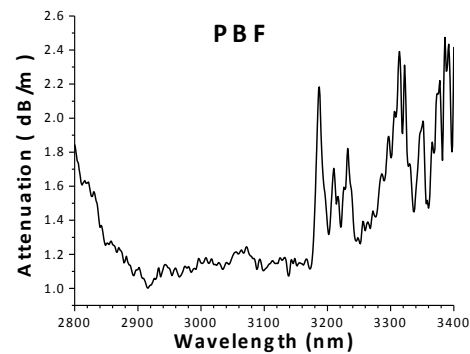


Figure 3: Attenuation of the Photonic Bandgap Fibre

The attenuation measurement for the PBF is shown in Figure 3. The average loss in the wavelength range $2.9 \mu\text{m}$ to $3.15 \mu\text{m}$ is ~ 1.2 dB/m and the loss at the wavelength of $2.94 \mu\text{m}$ is 1.1 dB/m. Attenuation was measured using a Bentham TM300 monochromator with a spectral resolution of ~ 20 nm using a cut-back technique. A tungsten halogen bulb was used as a broadband light source. The core size is $24 \mu\text{m}$ with a pitch of $7 \mu\text{m}$ and a NA of 0.12 . The pitch is the distance between two neighboured air holes. It should be noted that no particular care or special procedures were taken during fabrication of the fibres to control or minimise OH levels. The fibre was stored in a desiccator when it was not in use and no OH grow could be detected.

Theory

The guidance of the NCF is achieved by the Anti-resonant reflecting Optical Waveguide (ARROW) principle [17]. As described by Litchinitser et. al. [17] wavelengths which are in resonance with the core wall cannot be confined in the core but leak away through the wall, resulting in a high attenuation. However, frequencies that are anti-resonant with the wall cannot propagate within it and will be more confined inside the core. The two interfaces of wall and air can be described as a Fabry-Perot-like resonator. Anti-resonant wavelengths experience a low leakage through the wall and hence a lower attenuation as a result of destructive interference in the Fabry-Perot resonator. As the wavelength which is confined inside the core is dependent on the wall thickness it can be selected by adjusting this parameter. The losses can be explained by a coupling of core modes into modes supported by the nodes where two cladding elements meet [18]. The numerical simulation and first experimental presentation of the feasibility of guidance in similar fibres was shown by Pryamikov et. al. [19].

The guidance of the PBF on the other hand is described by Russell et. al. [20]. The periodic structure surrounding the fibre core leads to a photonic bandgap effect, so that no states are available and the light is

guided along the fibre. The wavelength which is confined inside the core is dependent on the pitch of the surrounding structure, where the pitch is the distance between the air holes. By changing the pitch of the fibre structure selection of the guided wavelength is possible.

Experimental

The laser light was coupled into both of the fibres using a CaF₂ lens of focal length $f=100$ mm. The beam diameter onto this lens was adjusted to fit the focus spot size to the different core sizes of the fibres, 24 μ m and 94 μ m for the PBF and NCF, respectively. For the NCF a focus spot size diameter of 67 μ m and a NA of 0.03 was used resulting in a coupling efficiency of $\sim 35\%$.

In the case of the PBF the spot size was set to be 46 μ m with an NA of 0.1, this is larger than the core of the fibre, however due to the relatively poor laser beam quality a compromise had to be made between the optimal spot size and NA. Tests with other NA values (0.2 and 0.05) resulted in reduced coupling efficiencies. However, it was apparent that there was still a significant mismatch between the laser and fibre fundamental mode which led to a very low coupling efficiency of around 5%. We would expect this coupling efficiency to be significantly improved by using a laser with improved beam quality to allow optimised matching into the fundamental mode of the fibre.

Optimal fibre alignment conditions were achieved by using a 3-axis micro block with the additional alignment of pitch and yaw, to maximise the output power. The initial alignment was performed with the laser running just above threshold and additional attenuation with microscope slides ($T=47\%$ per slide) [21]. This was done to protect the fibres from unwanted damage around the core. After the alignment was optimized the slides were removed and the laser power increased until the maximum output power could be determined. The input facet of the fibres was monitored with a camera to detect any damage due to laser radiation.

Results

The maximum output energy delivered through the NCF was 195 ± 1 mJ for a 33 cm length of fibre and 54 ± 4 mJ for a 988 cm length. The attenuation for these measurements appears higher than the value in Figure 2 however this is due to the fact that for practical reasons the 988 cm length was bent whereas the 33cm length was held straight. These pulse energies translate to energy densities of 2300 J/cm² for the short length and 764 J/cm² for the long length immediately at the end of the fibre with a core diameter of 94 μ m. It should also be noted that the limiting factor for the power delivery capability of the fibre was the laser source so in this case the maximum threshold of the fibre could not be determined. Both the input and output facets of the fibre were undamaged during the transmission experiments. It is likely that given a laser with higher output energy and/or better beam quality higher pulse energies could be delivered.

The maximum output energy delivered through the PBF was measured to be 14 mJ for a 44 cm long fibre. This translates to an energy density of 3465 J/cm² at the output end of the fibre with a core diameter of 24 μ m. At higher powers the fibre launch end sustained catastrophic damage and the cladding area surrounding the core was completely destroyed. The output facet was undamaged during any of the experiments and it had the appearance of a pristine cleaved end (as shown in Figure 1b). This gives confidence that optimised coupling would lead to improved performance.

A key parameter for a surgical laser delivery system is that the power delivered should not vary as it is manipulated. In endoscopy bend radii in the order of 15 cm are required [1] however for future minimally invasive procedures bend diameters in the order of 10's mm are envisaged hence it is desirable to develop fibres that show little or no bend sensitivity at these curvatures.

Both fibres guide in a single mode and are relatively bend insensitive. The NCF shows no appreciable loss of power at bend radii of around 15 cm whereas the PBF can be bent to diameters of less than 5 mm with no effect.

To demonstrate that tissue ablation is possible with this system a porcine bone was used. The fibre was sealed with a sapphire tip as was reported in [22], to protect the fibre core from debris and other contaminations. Successful cutting/drilling of the bone was performed using the fibre in contact and non-contact mode in air and under water. Figure 2 shows ablation using single shot pulses in contact mode.

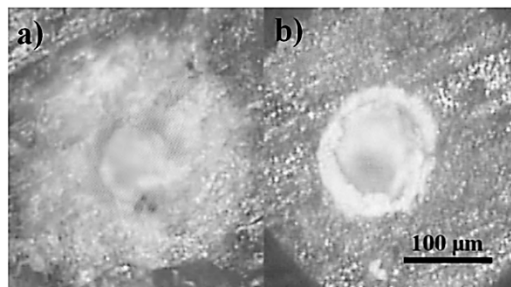


Figure 2: Demonstration of porcine bone ablation. a) Single shot in air. b) Single shot under water.

Discussion and conclusion

The delivered power energies for both types exceed the energies needed for biological tissue ablation by far. As reported by Pierce et. al. [23], human dental enamel has the highest ablation threshold with 35 J/cm^2 . For a 10m long NCF fibre the energy density delivered is a factor of >21 higher than that required. To verify this statement ablation of porcine bone was demonstrated in different environments. Both fibres demonstrate that they would be suitable components in a surgical laser delivery system and hence a promising alternative to the existing delivery systems already used in medicine and other high power applications at $2.94 \mu\text{m}$. Ultimately it is believed that these fibres may pave the way for novel minimally invasive surgical procedures.

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