Fabrication of silica hollow core photonic crystal fibres for Er:YAG surgical applications

Artur Urich*[a], Robert R. J. Maier[a], Jonathan C. Knight[b], Brian J. Mangan[b], Steven Renshaw[b], Duncan P. Hand[a], Jonathan D. Shephard[a]

[a] Applied Optics and Photonics Group, Heriot-Watt University, EH14 1AS Edinburgh, UK ;
[b] Centre for Photonics and Photonic Materials, Department of Physics, University of Bath, Bath, BA2 7AY, UK

ABSTRACT

In this work we present the fabrication of silica hollow core photonic crystal fibres (HC-PCF) with guidance at 2.94µm. As light is confined inside the hollow core with a very small overlap of the guided E-M wave with the fibre material, the high intrinsic loss of silica at these mid-infrared wavelengths can be overcome. The band gap effect is achieved by a periodic structure made out of air and fused silica. As silica is bio-inert, chemically stable and mechanically robust, these fibres have potential advantages over other multi-component, non-silica optical fibres designed to guide in this wavelength regime. These fibres have a relatively small diameter, low bend sensitivity and single-mode like guidance which are ideal conditions for delivering laser light down a highly flexible fibre. Consequently they provide a potential alternative to existing surgical laser delivery methods such as articulated arms and lend themselves to endoscopy and other minimally invasive surgical procedures. In particular, we present the characterisation and performance of these fibres at 2.94 µm, the wavelength of an Er:YAG laser. This laser is widely used in surgery since the wavelength overlaps with an absorption band of water which results in clean, non-cauterised cuts. However, the practical implementation of these types of fibres for surgical applications is a significant challenge. Therefore we also report on progress made in developing hermetically sealed end tips for these hollow core fibres to avoid contamination. This work ultimately prepares the route towards a robust, practical delivery system for this wavelength.

Keywords: Hollow Core-Photonic Crystal Fibre (HC-PCF), Er:YAG, Surgery, Fibre delivery, End-tip, End-cap, Fibre sealing,

1. INTRODUCTION

The Er:YAG laser radiation at 2.94µm is used in medical and surgical applications, due to the high absorption of water at this wavelength (see Figure 1). A result of this high absorption is a small penetration depth and therefore the unique capability of high ablation rates and precise tissue ablation with a minimal heat affected zone. The most common method for surgical laser delivery currently used is an articulated arm [1]. However, articulated arms have several disadvantages, as there are often misalignment issues, the beam can move during use of the arm and as it is a complex system a dedicated and skilled technician is needed for the installation. Also this system restricts the movement of the surgeon and cannot be implemented in endoscopic or minimal invasive procedures. A robust and flexible fibre delivery system would alleviate these problems and radically increase the usefulness of surgical lasers.

However the development of flexible fibre delivery system at this wavelength is challenging. Fibres based on Chalcogenides [2, 3], GeO_2 [3]or Sapphire [4] have all been investigated extensively. They are all large core (multimoded) fibres, in order to reduce the intensity of the energy in the core because of their low optical damage threshold, and hence beam quality will be sensitive to bending [5]. An important requirement for the fibres to be able to

*au36@hw.ac.uk; phone: +44 (0)131 451 4245

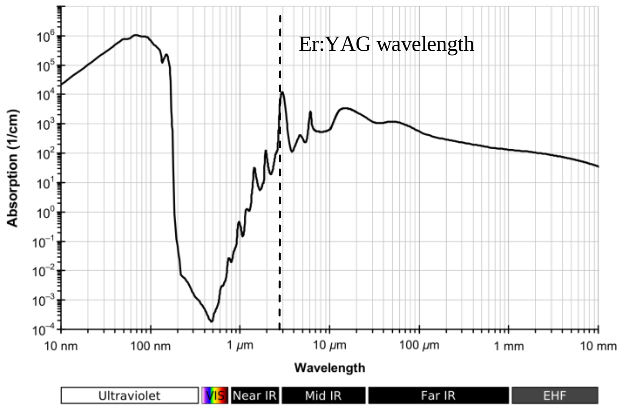

Figure 1: Absorption of water versus wavelength

be used in surgical applications is that they should be non-toxic and biocompatible, this is not always the case for the mid-IR fibres used at the moment. Also these fibres tend to have a low mechanical strength which does not fit with the need of reliability in surgery. There exists a range of hollow optical waveguides, which confine optical radiation by a number of different mechanisms, designed to guide in the infrared wavelength range from 3 µm to 20 µm [6]. One of the main benefits of this fibre type is the guidance of light in a gaseous medium and therefore it is expected that they have high damage thresholds compared to more traditional solid core fibres. Also, by eliminating end reflections, a high coupling efficiency is possible [7] and they are relatively low cost (compared to sapphire fibres or chalcogenide glass fibres). However, most types of hollow core fibre are highly multimode, which leads to high bend losses [8], [9] [10]. The multimode behaviour also has the disadvantage that the output beam profile changes as the fibre is bent which changes the power density available to the surgeon [11]. Obviously this is not ideal when considering the precise nature of surgical procedures, the surgeon needs confidence that his instrument will reliably and controllably cut tissue exactly where they want. One class of hollow waveguide, the *Hollow Core-Photonic Crystal Fibre* (HC-PCF) is of interest for many applications because these fibres can guide single-mode and are relatively bend insensitive compared to more traditional hollow waveguides [12], [13]. The unique guiding mechanism is achieved by a periodic structure surrounding the core. This structure is made of air holes in fused silica leading to a photonic bandgap material within which particular frequencies of E-M waves cannot propagate, and which are therefore confined inside the core. Delivery of high energy nanosecond pulses at 1064 nm has already been demonstrated with this class of fibre [12], [14]. However, with HC-PCFs it is also possible to extend the transmission window of silica significantly beyond that exhibited by the bulk material as the majority of the light is guided in air; indeed fibres have already been demonstrated that guide light of wavelength of ~3 µm [13], with low attenuation. The cross section of a hollow core fused silica fibre guiding in the 2.94 µm region is shown in Figure 2.

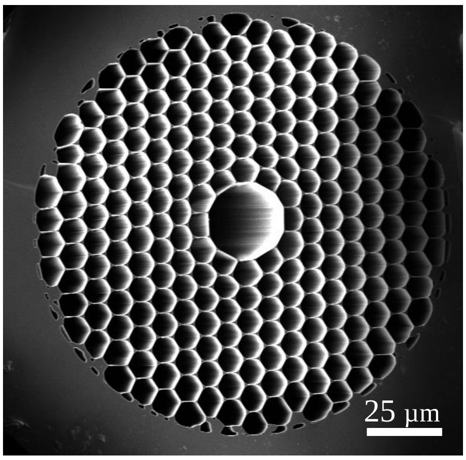

Figure 2: SEM picture of the HC-PCF used in this paper

In this paper we present work on the fabrication of and delivery of high power pulses through a HC-PCF at 2.94 µm and discuss the feasibility of a working system for surgical applications.

2. EXPERIMENTAL MATERIALS, METHODS AND SETUP

2.1 Laser

For the characterisation of high peak power laser beam delivery through the fibre an Impex High Tech ERB 15 laser was used. The operating wavelength is 2.937µm and the pulse length is 225µs FWHM, see Figure 3, at a repetition rate of ~15 Hz. The spatial profile of the laser for high power pulse configuration had a doughnut shape (see Figure 4) and a M^2 of ~2.5.

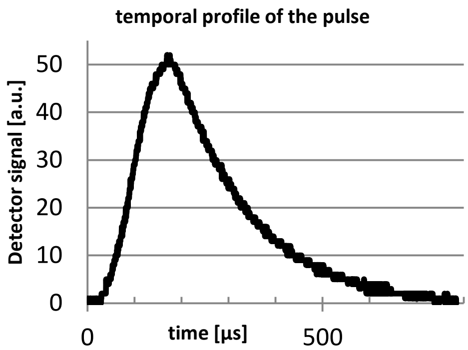

Figure 3: Temporal profile of the laser pulse

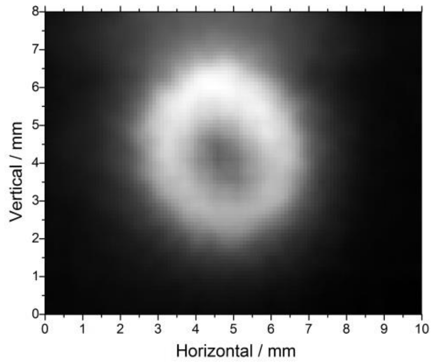

Figure 4: Spatial beam profile of the laser

2.2 Fibre

The production method for the fibre used in this work is a based on standard stack and draw process. A preform is stacked out of capillary tubes with a diameter of around 1-2mm, which has a geometry and periodic structure similar to that of the final desired fibre structure. This preform is then placed in a fibre drawing tower and drawn down to a cane. This cane is then subsequently drawn down to the final fibre dimensions. By controlling the pressure difference between the hollow core and the hollow and the space between cladding and jacketing tube cladding region during the final fibre drawing stage it is possible to control the pitch (distance between two air holes in the cladding) and the core diameter and it is these parameters that dictate the wavelength at which the fibres operate. Without this additional pressure the core and surrounding structure would tend to collapse. Another parameter which has to be taken into account is the drawing speed of the fibre. As the dimensions of the pitch and core dictate the bandgap of the fibre it is of critical importance to control these accurately. A schematic of this procedure is shown in Figure 5. Suprasil silica (F300) was used for the fabrication of the HCPCF. An attenuation measurement is shown in Figure 6. The loss at this wavelength is ~1.2dBm^{-1}. As can be seen in Figure 7 the bulk attenuation for Suprasil F300 at 2.94 µm is 50 dBm^{-1} [15]. No particular care or special procedure was taken during fabrication of the fibre to control water levels. However, following fibre fabrication it is necessary to keep the fibre hermetically sealed to avoid water ingress.

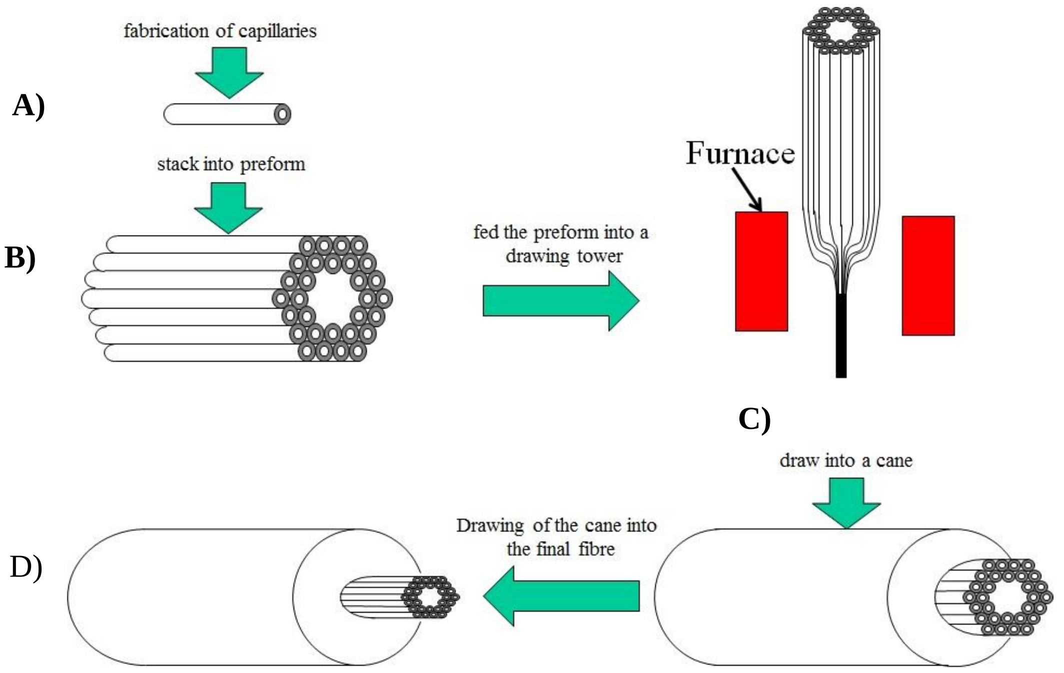

Figure 5: Steps in the fabrication of photonic crystal fibres. A) 1 mm thick capillaries are drawn to precise dimensions and then B) Stacked to form the desired "preform". C) The preform is fused together and drawn down in size to a "cane" (~ 1 mm in diameter). D) In the final drawing step, the cane is drawn down to fibre and encased in a silica outer cladding.

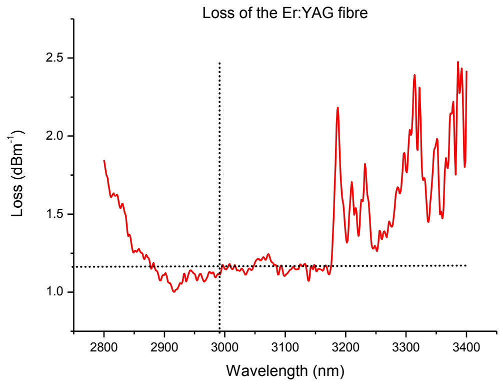

Figure 6: Loss spectrum for the 2.94µm fibre

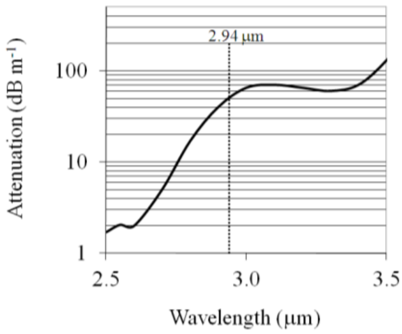

Figure 7: Bulk attenuation for dry silica (Suprasil F300) used for the fabrication of the HCPCF [15]

2.3 High power pulse delivery setup

For the high power pulse delivery the laser was operated with a maximum output power of about 260mJ. The laser output beam was magnified to 21mm (4σ) before using a final focusing lens with f=100mmm this lead to a spot size diameter of 46µm and a NA=0.12. The spot size is about double the size of the core but due to the relatively poor laser beam quality (M^2=2.5) a compromise had to me made between the spot size and the NA. Previous investigations (larger NA=0.2 and smaller NA=0.05) have shown that the match of the NA is more important than the spot size for optimum coupling. However the coupling efficiency was still relatively low around 5%. The power onto the launching side of the fibre was attenuated by using one or several glass microscope slides (T=47%) in conjunction with variation of the flash lamp current to regulate the initial laser output power.

Optimal fibre alignment was achieved by using a 3-axis microblock with additional pitch and yaw control. To prevent damage to the fibre launch side during the alignment the lowest possible power was set and the fibre was under constant monitoring by the means of a camera. The fibre output power was measured with an Ophir detector (PE50-DIF-ER-SH-V2 ROHS). During this experiment the fibre was bent with a radius of around 10cm.

3. RESULTS AND DISCUSSION

3.1 High power pulse delivery results

The maximum delivered power through a 0.4m long fibre was 14mJ. This is to our knowledge the highest power delivery in this fibre type at the 2.94µm wavelength. At higher powers the launching end of the fibre was destroyed as can be seen in Figure 8. However no damage at the output facet could be detected, which implies that with higher coupling efficiency the delivered power could be further increased. Figure 9 shows the false colour near field image of the delivered beam profile and corresponding cross section. The image was taken by imaging the beam reflected on a ceramic surface with a mid IR camera (Electrophysics Cop., PV320 – L2E).

Figure 8: Destroyed launching end of the fibre

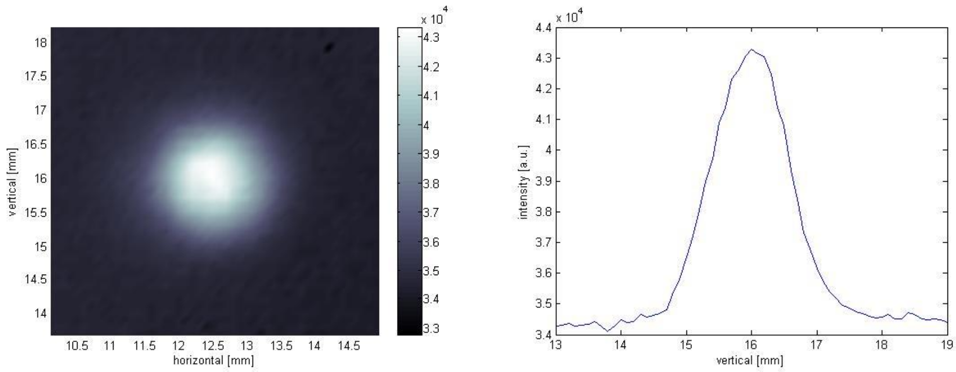

Figure 9: Beam profile of the delivered beam, a) False colour image as a reflection from a ceramic surface b) Cross section of the beam profile

One problem with hollow fibres is the contamination of the core with debris and, specifically in surgical applications such as endoscopy, with blood and other fluids. Therefore sealing of the output end is mandatory, our solution is the use of a sapphire “end-cap” as presented in [16]. A schematic of the end tip is shown in Figure 10, the hermeticity and mechanical stability was shown to be acceptable for practical considerations [16].

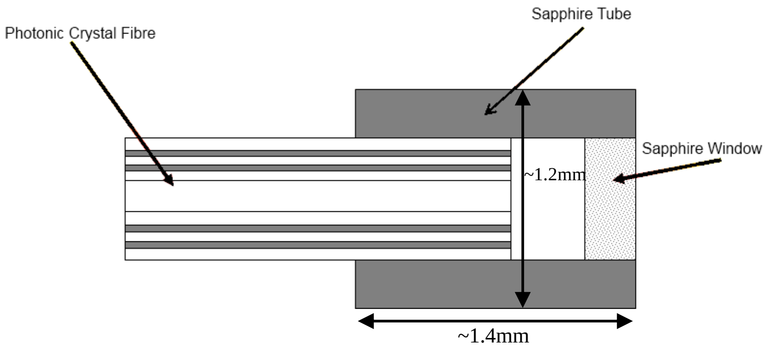

Figure 10: Schematic of the end tip

A typical distance from the fibre end to the machined tissue is considered to be about 500µm due to the thickness of the "window" of the end tip and the distance of fibre to the window. With the NA of the fibre being 0.12 and a core diameter of 24µm this would lead to a spot size of 140µm. Taking into account the losses induced by the endtip (15%) and the fluctuation of the laser, a realistic delivered pulse energy would be 10.2mJ, this leads to an energy density of 66.2 Jcm^{-2}, which considerably exceeds the ablation threshold of biological hard and soft tissue which is in the order of 1.6-6 Jcm^{-2} for skin and bone and up to 35 Jcm^{-2} for human dental enamel [17-22].

4. CONCLUSION

In this work we demonstrated transmission of high power pulses at 2.94µm through an all silica HC-PCF. The maximum pulse energy was 14mJ which is to our knowledge the highest achieved in this type of fibre. Also a practical approach was investigated showing that with an end tipped version power densities up to 66.2 Jcm^{-2} are feasible. Therefore we believe that the overall system could be applicable for surgical applications due to its many advantages as high power pulse delivery, biocompatibility, bend insensitivity, flexibility and good beam quality compared to other fibre delivery systems at this wavelength.

REFERENCES

1. Stubinger, S., et al., *Er : YAG laser osteotomy for removal of impacted teeth: Clinical comparison of two techniques.* Lasers in Surgery and Medicine, 2007. **39**(7): p. 583-588.
2. Sanghera, J.S., L.B. Shaw, and I.D. Aggarwal, *Applications of chalcogenide glass optical fibers.* Comptes Rendus Chimie, 2002. **5**(12): p. 873-883.
3. Scott, N.J., et al., *Mid-IR germanium oxide fibers for contact erbium laser tissue ablation in endoscopic surgery.* Ieee Journal of Selected Topics in Quantum Electronics, 2007. **13**(6): p. 1709-1714.
4. Fried, N.M., et al., *Transmission of Q-switched erbium: YSGG (lambda=2.79 mu m) and erbium: YAG (lambda=2.94 mu m) laser radiation through germanium oxide and sapphire optical fibres at high pulse energies.* Lasers in Medical Science, 2004. **19**(3): p. 155-160.
5. Kuhn, A., et al., *Optical fibre beam delivery of high-energy laser pulses: beam quality preservation and fibre end-preparation.* Optics and Lasers in Engineering, 2000. **34**(4-6): p. 273-288.

6. Hongo, A., et al., *Fabrication of dielectric-coated silver hollow glass waveguides for the infrared by liquid-flow coating method*, in *Biomedical Fiber Optics, Proceedings Of*1996, Spie - Int Soc Optical Engineering: Bellingham. p. 55-63.
7. Hensley, C.J., et al., *Extremely High Coupling and Transmission of High-Powered-Femtosecond Pulses in Hollow-Core Photonic Band-Gap Fiber*, in *2008 Conference on Lasers and Electro-Optics & Quantum Electronics and Laser Science Conference, Vols 1-9*2008, Ieee: New York. p. 2010-2011.
8. Russell, P.S.J., *Photonic-crystal fibers.* Journal of Lightwave Technology, 2006. **24**(12): p. 4729-4749.
9. Ferrarini, D., et al., *Leakage properties of photonic crystal fibers.* Optics Express, 2002. **10**(23): p. 1314-1319.
10. Parry, J.P., et al., *Analysis of optical damage mechanisms in hollow-core waveguides delivering nanosecond pulses from a Q-switched Nd : YAG laser.* Applied Optics, 2006. **45**(36): p. 9160-9167.
11. Harrington, J.A., *A review of IR transmitting, hollow waveguides.* Fiber and Integrated Optics, 2000. **19**(3): p. 211-227.
12. Shephard, J.D., et al., *High energy nanosecond laser pulses delivered single-mode through hollow-core PBG fibers.* Optics Express, 2004. **12**(4): p. 717-723.
13. Shephard, J.D., et al., *Single-mode mid-IR guidance in a hollow-core photonic crystal fiber.* Optics Express, 2005. **13**(18): p. 7139-7144.
14. Shephard, J.D., et al., *Improved hollow-core photonic crystal fiber design for delivery of nanosecond pulses in laser micromachining applications.* Applied Optics, 2005. **44**(21): p. 4582-4588.
15. Humbach, O., et al., *Analysis of OH absorption bands in synthetic silica.* Journal of Non-Crystalline Solids, 1996. **203**: p. 19-26.
16. Urich, A., et al. *Towards implementation of hollow core fibres for surgical applications*. 2011. SPIE.
17. Walsh, J.T. and T.F. Deutsch, *Er Yag Laser Ablation of Tissue - Measurement of Ablation Rates.* Lasers in Surgery and Medicine, 1989. **9**(4): p. 327-337.
18. Pierce, M.C., M.R. Dickinson, and H. Devlin, *Selective photothermal ablation of tissue with a fibre delivered Er : YAG laser.* Laser-Tissue Interaction X: Photochemical, Photothermal, and Photomechanical, Proceedings Of, 1999. **3601**: p. 362-368.
19. Hohenleutner, U., et al., *Fast and effective skin ablation with an Er:YAG laser: Determination of ablation rates and thermal damage zones.* Lasers in Surgery and Medicine, 1997. **20**(3): p. 242-247.
20. Wesendahl, T., et al., *Erbium:YAG Laser Ablation of Retinal Tissue under Perfluorodecaline: Determination of Laserâ€“Tissue Interaction in Pig Eyes.* Investigative Ophthalmology & Visual Science, 2000. **41**(2): p. 505-512.
21. Nishimoto, Y., et al., *Effect of pulse duration of Er : YAG laser on dentin ablation.* Dental Materials Journal, 2008. **27**(3): p. 433-439.
22. Contente, M., et al., *Temperature rise during Er:YAG cavity preparation of primary enamel.* Lasers in Medical Science: p. 1-5.