Citation for published version:
Urich, A, Maier, RRJ, Mangan, BJ, Renshaw, S, Knight, JC, Hand, DP & Shephard, JD 2012, 'Delivery of high energy Er:YAG pulsed laser light at 2.94ím through a silica hollow core photonic crystal fibre', *Optics Express*, vol. 20, no. 6, pp. 6677-6684. https://doi.org/10.1364/oe.20.006677

DOI:
10.1364/oe.20.006677

Publication date:
2012

Delivery of high energy Er:YAG pulsed laser light at 2.94µm through a silica hollow core photonic crystal fibre

A. Urich,[1,*] R. R. J. Maier,[1] B. J. Mangan,[2] S. Renshaw,[2] J. C. Knight,[2] D. P. Hand,[1] and J. D. Shephard[1]

[1]*Applied Optics and Photonics group, School of Engineering and Physical Sciences, Heriot-Watt University, Edinburgh, EH14 4AS, UK*

[2]*Centre for Photonics and Photonic Materials, Department of Physics, University of Bath, Bath, BA2 7AY, UK*

**au36@hw.ac.uk*

Abstract: In this paper the delivery of high power Er:YAG laser pulses through a silica hollow core photonic crystal fibre is demonstrated. The Er:YAG wavelength of 2.94 µm is well beyond the normal transmittance of bulk silica but the unique hollow core guidance allows silica to guide in this regime. We have demonstrated for the first time the ability to deliver high energy pulses through an all-silica fibre at 2.94 µm. These silica fibres are mechanically and chemically robust, biocompatible and have low sensitivity to bending. A maximum pulse energy of 14 mJ at 2.94 µm was delivered through the fibre. This, to our knowledge, is the first time a silica hollow core photonic crystal fibre has been shown to transmit 2.94 µm laser light at a fluence exceeding the thresholds required for modification (e.g. cutting and drilling) of hard biological tissue. Consequently, laser delivery systems based on these fibres have the potential for the realization of novel, minimally-invasive surgical procedures.

OCIS codes: (060.2270) Fiber characterization; (060.4005) Microstructured fibers; (060.5295) Photonic crystal fibers; (170.1020) Ablation of tissue.

References and links

1. S. Stübinger, B. von Rechenberg, H. F. Zeilhofer, R. Sader, and C. Landes, "Er:YAG laser osteotomy for removal of impacted teeth: clinical comparison of two techniques," Lasers Surg. Med. **39**(7), 583–588 (2007).
2. J. S. Sanghera, L. B. Shaw, and I. D. Aggarwal, "Applications of chalcogenide glass optical fibers," C. R. Chim. **5**(12), 873–883 (2002).
3. N. J. Scott, R. A. Barton, A. L. Casperson, A. Tchapyjnikov, K. Levin, D. Tran, and N. M. Fried, "Mid-IR germanium oxide fibers for contact erbium laser tissue ablation in endoscopic surgery," IEEE J. Sel. Top. Quantum Electron. **13**(6), 1709–1714 (2007).
4. N. M. Fried, Y. B. Yang, C. A. Chaney, and D. Fried, "Transmission of Q-switched erbium:YSGG (lambda=2.79 microm) and erbium:YAG (lambda=2.94 microm) laser radiation through germanium oxide and sapphire optical fibres at high pulse energies," Lasers Med. Sci. **19**(3), 155–160 (2004).
5. A. Hongo, M. Miyagi, Y. Kato, M. Suzumura, S. Kubota, Y. Wang, and T. Shimomura, "Fabrication of dielectric-coated silver hollow glass waveguides for the infrared by liquid-flow coating method," in *Proceedings of Biomedical Fiber Optics,* (Spie - Int Soc Optical Engineering, Bellingham, 1996), pp. 55–63.
6. J. A. Harrington, "A review of IR transmitting, hollow waveguides," Fiber Integr. Opt. **19**(3), 211–227 (2000).
7. B. F. Bowden and J. A. Harrington, "Fabrication and characterization of chalcogenide glass for hollow Bragg fibers," Appl. Opt. **48**(16), 3050–3054 (2009).
8. C. J. Hensley, M. A. Foster, B. Shim, and A. L. Gaeta, "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* (Ieee, New York, 2008), pp. 2010–2011.
9. P. S. J. Russell, "Photonic-crystal fibers," J. Lightwave Technol. **24**(12), 4729–4749 (2006).
10. J. P. Parry, T. J. Stephens, J. D. Shephard, J. D. C. Jones, and D. P. Hand, "Analysis of optical damage mechanisms in hollow-core waveguides delivering nanosecond pulses from a Q-switched Nd:YAG laser," Appl. Opt. **45**(36), 9160–9167 (2006).

#160528 - $15.00 USD Received 23 Dec 2011; revised 17 Feb 2012; accepted 20 Feb 2012; published 7 Mar 2012

11. J. D. Shephard, J. D. C. Jones, D. P. Hand, G. Bouwmans, J. C. Knight, P. S. Russell, and B. J. Mangan, "High energy nanosecond laser pulses delivered single-mode through hollow-core PBG fibers," Opt. Express **12**(4), 717–723 (2004).
12. J. D. Shephard, W. N. Macpherson, R. R. J. Maier, J. D. C. Jones, D. P. Hand, M. Mohebbi, A. K. George, P. J. Roberts, and J. C. Knight, "Single-mode mid-IR guidance in a hollow-core photonic crystal fiber," Opt. Express **13**(18), 7139–7144 (2005).
13. J. D. Shephard, F. Couny, P. S. Russell, J. D. C. Jones, J. C. Knight, and D. P. Hand, "Improved hollow-core photonic crystal fiber design for delivery of nanosecond pulses in laser micromachining applications," Appl. Opt. **44**(21), 4582–4588 (2005).
14. O. Humbach, H. Fabian, U. Grzesik, U. Haken, and W. Heitmann, "Analysis of OH absorption bands in synthetic silica," J. Non-Cryst. Solids **203**, 19–26 (1996).
15. J. T. Walsh, Jr. and T. F. Deutsch, "Er:YAG laser ablation of tissue: measurement of ablation rates," Lasers Surg. Med. **9**(4), 327–337 (1989).
16. L. J. A. F. Rainer, J. H. Campbell, F. D. DeMarco, M. R. Kozolowski, A. J. Morgan, and M. C. Staggs, "Four-harmonic database of laser-damage testing," Proc. SPIE **1624**, 116–127 (1992).
17. Y. Nishimoto, M. Otsuki, M. Yamauti, T. Eguchi, Y. Sato, R. M. Foxton, and J. Tagami, "Effect of pulse duration of Er: YAG laser on dentin ablation," Dent. Mater. J. **27**(3), 433–439 (2008).
18. A. Urich, T. Delmonte, R. R. J. Maier, D. P. Hand, and J. D. Shephard, "Towards implementation of hollow core fibres for surgical applications," Proc. SPIE **12**, 78940W (2011).
19. M. C. Pierce, M. R. Dickinson, and H. Devlin, "Selective photothermal ablation of tissue with a fibre delivered Er: YAG laser," Proc. SPIE **3601**, 362–368 (1999).
20. U. Hohenleutner, S. Hohenleutner, W. Bäumler, and M. Landthaler, "Fast and effective skin ablation with an Er:YAG laser: determination of ablation rates and thermal damage zones," Lasers Surg. Med. **20**(3), 242–247 (1997).
21. T. Wesendahl, P. Janknecht, B. Ott, and M. Frenz, "Erbium: YAG laser ablation of retinal tissue under perfluorodecaline: determination of laser-tissue interaction in pig eyes," Invest. Ophthalmol. Vis. Sci. **41**(2), 505–512 (2000).
22. M. Contente, F. de Lima, R. Galo, J. Pécora, L. Bachmann, R. Palma-Dibb, and M. Borsatto, "Temperature rise during Er:YAG cavity preparation of primary enamel," Lasers Med. Sci. **1**, 1–5 (2012).

1. Introduction

Er:YAG lasers operating at 2.94 µm are used for medical and surgical applications. Currently, delivery of surgical laser light for use in the operating theatre is achieved using articulated arms [1]. There are a number of shortfalls of these systems, in particular misalignment issues, where the beam is slightly off-axis, resulting in beam wander as the arm is manipulated. These systems are also difficult to maintain. Additionally, articulated arms restrict the movement of the surgeon and reduce the capabilities for endoscopic or minimally 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 reliable fibres for this wavelength is challenging and the fibres which are available (both solid and hollow core) have significant drawbacks.

There are a number of solid core fibres that have been investigated for this application area, based on Chalcogenides [2,3], GeO_2 [3] or Sapphire [4]. Although appreciable power delivery has been demonstrated, in general it is necessary to reduce the energy density carried in the core to overcome the Laser Induced Damage Threshold (LIDT) of the bulk material. Consequently, they tend to be large core (multimoded) fibres, and hence the energy distribution and density at the output of the fibre will be sensitive to bending [5] due to changes in modal patterns. The main drive for fibre delivery is the flexibility it gives the users and it is often important or beneficial to have a small bend radius.

There also exists a range of hollow optical waveguides which confine the optical radiation by different mechanisms, e.g. total internal reflection; Bragg reflection; and internal reflection at a dielectric coated metallic interface in the case of leaky tube waveguides, which guide in the infrared wavelength range up to 20 µm [5–7]. A key benefit of all hollow optical waveguides fibres is that the optical power is mostly contained within a gas and they therefore typically have higher damage thresholds than solid core fibres. Also, since there are no Fresnel reflections at the fibre ends, a high coupling efficiency is possible [8]. However, during practical use an end capsulation of the fibre at the output side is often required, which would introduce loses due to Fresnel reflections at around 6.5% per surface for a sapphire

#160528 - $15.00 USD Received 23 Dec 2011; revised 17 Feb 2012; accepted 20 Feb 2012; published 7 Mar 2012

based endtip. As regards optical damage, these encapsulation surfaces can be placed a short distance from the fibre end to ensure a reduced intensity due to beam divergence (and hence increased damage threshold) compared with the end-face of a solid core fibre. However most types of hollow core fibre support many guided modes which, as with the solid core multimode fibres described earlier, leads to high bend losses, multi-modal behaviour and changes in output beam profile changes as the fibre is bent [6], [9]- [10].

The exception is hollow-core photonic crystal fibres (HC-PCF), which can offer single-mode guidance and which are relatively bend insensitive compared with traditional hollow waveguides [11], [12]. This allows increased flexibility and accessibility in applications such as laser surgery. 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 electro-magnetic waves cannot propagate for a range of angles, confining guided modes inside the core. Delivery of high energy nanosecond pulses at 1064 nm has already been demonstrated with this class of fibre [11], [13] showing the marked increase in single mode power delivery possible with hollow core guidance. 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 [12] with relatively low attenuation. The cross section of a hollow core fused silica fibre guiding in the 2.94 µm region is shown in Fig. 1.

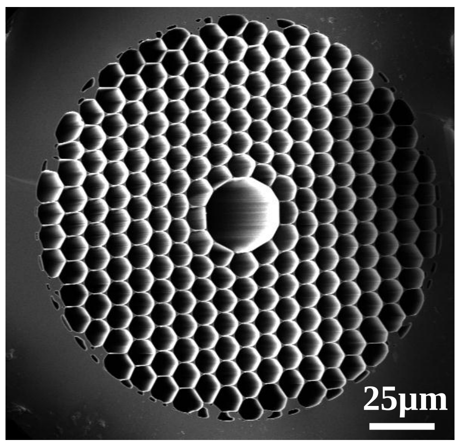

Fig. 1. SEM picture of the HC-PCF used in this paper, the core diameter is 24 µm, the pitch is ~8 µm and the PCF region is ~135 µm across.

Of specific interest for surgical applications is the 2.94 µm wavelength because the water contained in human tissue strongly absorbs light at this wavelength. The high absorption leads to a small penetration depth and therefore the unique capability of high ablation rates and precise tissue ablation with a minimal heat-affected zone. The thresholds for laser ablation at 2.94 µm for different tissue types are shown in Table 1. We present, for the first time to our knowledge, single-mode like pulse delivery through a silica HC-PCF which exceeds these thresholds and therefore has the potential to be used for cutting and drilling of soft and hard tissue, and hence be implemented in surgical applications where a high power laser radiation is needed and/or a single mode profile desired.

#160528 - $15.00 USD Received 23 Dec 2011; revised 17 Feb 2012; accepted 20 Feb 2012; published 7 Mar 2012

Table 1. Ablation Thresholds for Different Tissue Types

Rep rate [Hz]	Pulse length [μs]	Tissue type	Threshold [J/cm^2]	Reference
2	250	Human dental enamel	35	[19]
7-10	250	Human skin	1.6	[20]
1.7	250	Pig retina	1	[21]
1	100-5000	Human dentine	2.69-3.66	[17] [10]
5	NA	Pig skin (vitro)	3.6 - 5.6	[22]
2	200	Guinea pig skin	0.6-1.5	[15]
2	200	Guinea pig bone	2.1-3.4	[15]

2 Pulse delivery

2.1 Fibre

The production method for the HC-PCF presented in this paper is a standard stack and draw process with parameters optimized to achieve a bandgap at around 2.94 µm. Fused silica (Suprasil F300) was used for the fabrication of the HC-PCF. An attenuation measurement is shown in Fig. 2. The average loss in the wavelength range 2.9 µm to 3.15 µm is ~1.2 dB/m and the loss at the wavelength of the Er:YAG laser 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. As can be seen in Fig. 3 the bulk attenuation for Suprasil F300 at 2.94 µm is 50 dB/m [14]. No particular care or special procedures were taken during fabrication of the fibre to control or minimise OH levels.

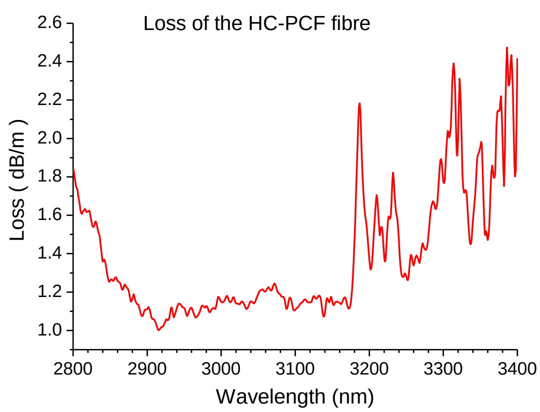

Fig. 2. Loss spectrum for the 2.94µm fibre measured by a cutback technique from an initially 5m long fibre sample in three s

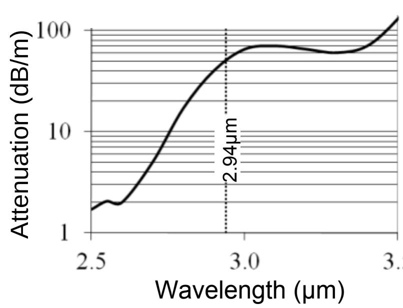

Fig. 3. Bulk attenuation for dry silica (Suprasil F300) used for the fabrication of the HC-PCF, from [14].

#160528 - $15.00 USD Received 23 Dec 2011; revised 17 Feb 2012; accepted 20 Feb 2012; published 7 Mar 2012

2.2 Experimental Setup

2.2.1 Laser

For the pulse delivery investigation an Impex High Tech ERB 15 laser was used. The operating wavelength is 2.937 µm and the pulse-length is 225 µs FWHM (Fig. 4a), with a M^2 of ~2.5 at a repetition rate of ~15 Hz. The spatial profile of the laser has a doughnut shape (Fig. 4b), the low resolution in the image is a result of the relative large pixel size of the IR camera used (each pixel is 50 × 50 µm and in this arrangement imaged an area ~100 × 100 µm).

The laser output was expanded to a diameter of 2.1 cm (4σ beam width) and focused into the 0.4 m long fibre using a CaF_2 lens with f = 100 mm, giving an NA of 0.1 and a corresponding spot size diameter of 46 µm. This is double the size of the core, but due to the relative 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. Previous work with delivery of high energy pulses through HC-PCF has shown that efficiencies around 98% should be achievable [8]. However, for this study a high power laser source at 2.94 µm with high beam quality was not available.

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. For the initial alignment the laser was operated just above threshold in conjunction with one or several glass microscope slides for variable attenuation (T = 47% per slide) [15]. Once the alignment was optimised the slides were removed and the laser output power gradually increased to determine the maximum transmission. The front facet of the fibre was monitored with a camera to detect any damage due to laser radiation. The fibre was bent with a radius of ~10 cm during these experiments.

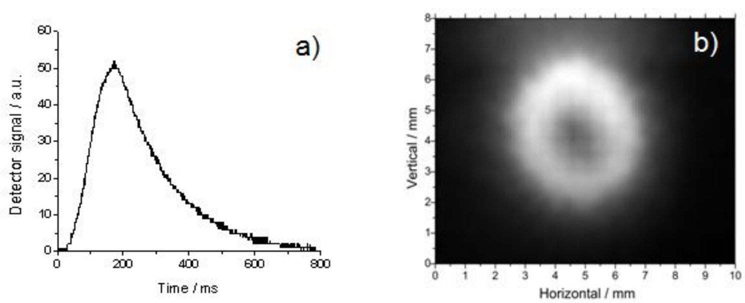

Fig. 4. a) Temporal profile of the laser pulse b) Spatial beam profile of the laser.

2.3 *Results*

The maximum output power delivered through the HC-PCF was measured to be 210 mW at ~15 Hz, corresponding to a pulse energy of 14 mJ and a pulse length of 225 µs FWHM. At higher powers the fibre launch end sustained catastrophic damage and the cladding area surrounding the core was completely destroyed, as can be seen in Fig. 5.

#160528 - $15.00 USD Received 23 Dec 2011; revised 17 Feb 2012; accepted 20 Feb 2012; published 7 Mar 2012

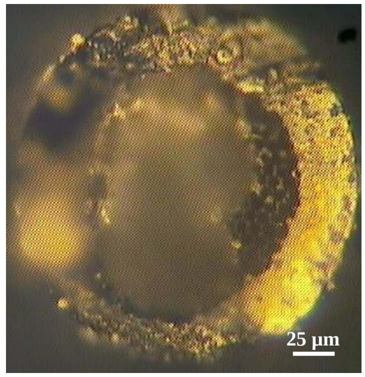

Fig. 5. Optical microscopy image of catastrophic damage at the launch end of the fibre. Most of the periodic structure is destroyed.

However, even with the damaged fibre launch end, it was still possible to guide light through the fibre and a power of 80 mW (5.4 mJ) could still be transmitted. No damage to the output facet was observed during any of the experiments and the output facet had the appearance of a pristine cleaved end (as shown in Fig. 1) which included pulse delivery experiments with the damaged launch facet. To evaluate the LIDT of the HC-PCF, light was deliberately focused onto the cladding structure and the pulse energy required to cause damage was determined. The cladding structure damaged at a fluence of ~30 kJ/cm^2.

Previous work [11] found the LIDT for the cladding structure of a similar silica HC-PCF to be 130 – 140 J/cm^2 for an 8 ns pulse (at 1064 nm). The approximate factor of $t_p^{1/2}$, where t_p is the pulse width, can be used to scale energies [16] from an 8 ns to a 250 µs pulse. Scaling the 8 ns pulse results gives an expected LIDT of 23.5 kJ/cm^2 for a 225 µs pulse which is in close agreement to the value found in these experiments (30 kJ/cm^2). In fact one would expect a slightly different value for the LIDT in these experiments, which were conducted at 2.94 µm compared to 1.064 µm, as materials have be shown to have a different LIDT at different wavelengths [17]. The potential capability of these fibres for pulse delivery can be estimated by considering a Gaussian beam at the input face. Fibre damage will occur if the intensity at the input face incident on the silica glass surrounding the air core exceeds the damage threshold. If the fluence in the silica glass exceeds the 30 kJ/cm^2 measured above, then the fibre will be damaged. The mode field diameter of the fibre is estimated to be the same size as the core i.e. 24 µm. When coupling a Gaussian beam with a 1/e^2 width of 24 µm into the fibre, with a fluence of 111 kJ/cm^2 in the core region, the intensity incident on the silica cladding will be in the order of the damage threshold of 30 kJ/cm^2. At the fluence of 111 kJ/cm^2 in the 24 µm diameter core a pulse of 500 mJ is being delivered. This value could be considered to be the absolute maximum pulse energy deliverable with these fibres using an optimised laser source with sufficiently good beam quality to achieve a Gaussian beam waist of appropriate diameter and divergence.

The output facet of the fibre showed no damage during the experiments, even after the launch end was destroyed. This gives confidence that optimised coupling would lead to improved performance. The grey scale images for the near and far field beam profile and corresponding cross section are shown in Fig. 6 and it can be seen the beam is roughly circular and single-mode like. The pictures were made by capturing the reflection of the laser light from the fibre end on a ceramic surface using a mid-infrared camera (Electrophysics PV320-L2E). A ~15 × magnification telescope was used to take the near field images. Additionally the image was not taken in the exact focal plane however the image plane was still within the Rayleigh length to enhance the resolution. The offset of the pictures is due to background noise.

One of the potential problems with implementing a hollow core fibre in surgical applications is possible contamination due to ingress of blood and other fluids during *in-vivo* use. In order to protect and seal the 'open' end of the fibre from liquid ingress we previously developed a sapphire "end cap" [18]. With such an end-capped fibre, a typical distance of up to 500 µm could be anticipated between fibre end facet and the tissue due to the thickness of the window in the endtip. The NA of the fibre was measured as 0.12, which gives a spot size of 140 µm diameter (core diameter is 24 µm) at a distance of 500 µm from the fibre end. If pulses of energy 12 mJ (180 mW) are delivered through the fibre and assuming a total loss due to the endtip of around 15% the pulse energy incident on the tissue would be 10.2 mJ. Hence the energy density in a 140 µm spot would be 66.2 J/cm^2. A value of 12 mJ (180 mW) for the fibre output was chosen as tests were done at this level to check the stability. The output power was measured for 5 minutes (4425 pulses) at 180 mW with no observable damage to the launch facet. During this time the output power changed by ± 30 mW, which was attributed to the instability of the laser and slight movements of the fibre at the input end. The fluctuations in output power had no effect on the pulse delivery capability of the fibre. However for a clinical system a more stable laser is required to ensure a constant output power. The experiments were carried out over a number of weeks, and during this period no degradation of the fibre was detected. The demonstrated power density of 66.2 J/cm^2 exceeds the thresholds needed for cutting and drilling of biological tissue, as shown in Table 1.

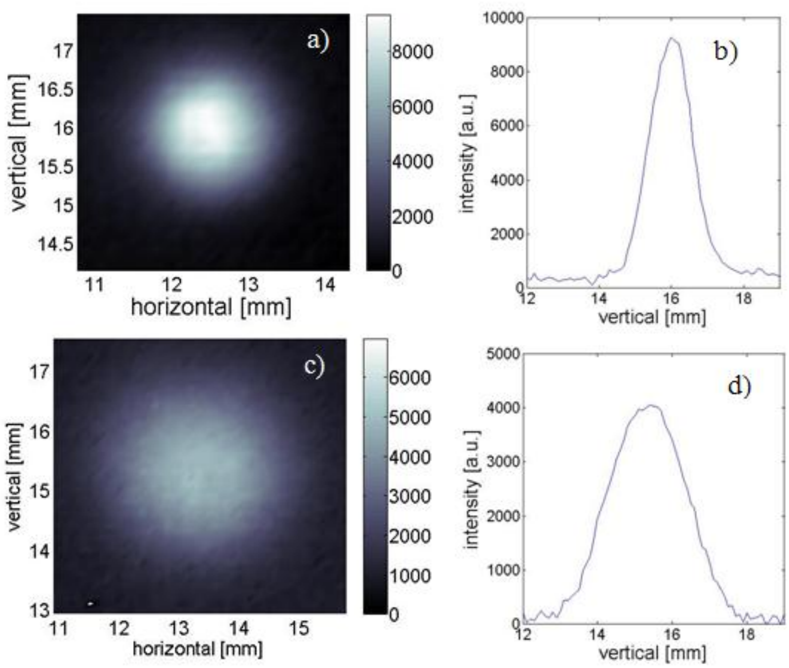

Fig. 6. Near- and far-field beam profiles. a): near-field profile at the fibre end with b) associated line scan (images are magnified); c) far-field profile at a distance of 4 cm from the fibre output end with d) associated line scan.

3 Conclusion

In this paper we demonstrate the ability to deliver, for the first time, high energy microsecond pulsed Er:YAG laser light at a wavelength of 2.94 µm through a silica HCPCF. We have shown that despite the non-optimised launch conditions we are able to deliver pulse energies up to 14 mJ at a pulse length of 225 µs (FWHM). Taking into account practical considerations for realising a surgical delivery system, such as end capping, an estimated power density of

66.2 J/cm^2 is achievable which exceeds previously reported data for the ablation of biological tissue. Additionally, silica HC-PCF's are more mechanically and chemically robust, less bend sensitive and more biocompatible compared to current fibres fabricated from alternative materials for delivery at this wavelength. Our results therefore demonstrate that silica HC-PCF have great potential to be developed into surgical laser delivery systems and offers a promising new alternative to the established articulated arms or large core fibres providing improved flexibility and enabling and enhancing minimally invasive surgical procedures.

Acknowledgments

This work was funded by the UK Engineering and Physical Sciences Council under EP/G039097/1.