# Flexible delivery of Er:YAG radiation at 2.94 µm with novel hollow-core silica glass fibres: Demonstration of tissue ablation

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## ABSTRACT

In this work we present the delivery of high energy Er:YAG laser pulses operating at 2.94 µm through a hollow-core negative curvature fibre (HC-NCF) and a hollow-core photonic crystal fibre (HC-PCF) and their use for the ablation of biological tissue. In HC-NCF fibres, which have been developed recently, the laser radiation is confined in a hollow core and by an anti-resonant or reflection principle (also known as ARROW). Both fibres are made of fused silica which has high mechanical and chemical durability, is bio-inert and results in a fibre with the flexibility that lends itself to easy handling and minimally invasive procedures. The HC-NCF structure consists of only one ring of capillaries around a realtively large core, followed by a protecting outer layer, hence the preform is relatively easy to build compared to traditional HC-PCF. The measured attenuation at 2.94 µm is 0.06 dB/m for the HC-NCF and 1.2 dB/m for the HC-PCF. Both fibres have a single mode output beam profile which can be advantageous for surgical applications as the beam profile is maintained during fibre movement. We demonstrate delivery of high energy pulses through both fibres, well above the thresholds needed for the ablation of biological tissue in non-contact and contact mode. Delivered energy densities reached > 750 J/cm$^{-2}$ after 10 m of HC-NCF and > 3400 J/cm$^{2}$ through a 44 cm HC-PCF.

This flexible high energy delivery system offers an alternative to existing beam delivery systems such as articulated arms and large core multi-mode fibres with enhanced capabilities.

**Keywords:** Hollow Core-Negative Curvature Fibre (HC-NCF) Er:YAG, Surgery, Fibre delivery, End-tip, End-cap, Fibre sealing, Tissue ablation

## 1. INTRODUCTION

The Er:YAG laser radiation at 2.94 µm is used in medical and surgical applications due to the high absorption of optical power in 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. Two of the common methods for surgical laser delivery currently used are articulated arms [1] and optical fibres. However articulated arms restrict the movement of the surgeon and cannot be implemented in endoscopic or minimal invasive procedures. Therefore a robust and flexible fibre delivery system would alleviate these problems and radically increase the usefulness of surgical lasers.

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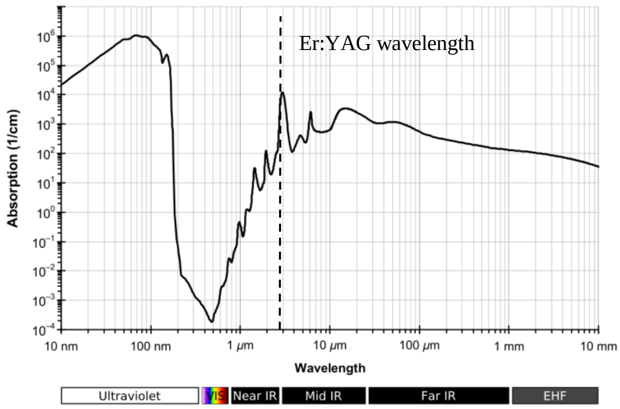


**Figure 1: Absorption of water versus wavelength from [2]**

However, the development of flexible fibre delivery system at this wavelength is challenging. Solid core fibres based on Chalcogenides [3, 4], $GeO_2$ [4] or Sapphire [5] have all been investigated extensively and shown to deliver appreciable energies. To reduce energy densities in the solid fibre core to values below the Laser Induced Damage Threshold (LIDT) of the material a large mode area is normally required. However, these large mode fibres tend to be more bend sensitive which can affect the output beam quality [6]. Also, some of these fibres contain toxic materials which are not ideal for medical applications but this can be overcome by packaging of the fibre.

An alternative to the above fibres is the use of hollow optical fibres, which confine the optical radiation by a number of different mechanisms [7]. The main benefit of this design is the small overlap of the light with the material, thus the LIDT can potentially be much higher than in a traditional solid core fibre. By virtually eliminating end reflections, a high coupling efficiency is possible [8] and they are relatively low cost (compared to sapphire fibres or chalcogenide glass fibres). However, most types of hollow core fibres are highly multimode which leads to high bend losses [9-11]. 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 [12]. This may not be a major issue if used in contact mode but ideally the surgeon needs a constant power output of the fibre and power density on the tissue and therefore these fluctuations are in general undesirable for surgical procedures.

In this work two different designs are presented which can overcome the restrictions and issues described above and provide a potential alternative for developing minimally invasive procedures with Er:YAG lasers. The first design presented here is the Hollow Core-Negative Curvature Fibre (HC-NCF) which is so called due to the central hollow core geometry, as shown in Figure 2a [13]. The laser radiation is confined in a hollow core by an anti-resonant reflection principle (also known as ARROW). This was described, by Litchinitser et. al. [14], as a mechanism where 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.

The second design is the Hollow Core-Photonic Crystal Fibre (HC-PCF) as seen in Figure 2. This fibre type guides single-mode and is relatively bend insensitive, as is shown in references [15, 16]. 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 for a range of angles, and which are therefore confined inside the core. The delivery of high energy nanosecond pulse at 1064 nm has already be demonstrated in such a fibre [15], [17]. As the interaction of light with the fibre material is minimal in this case it has been shown to be possible to extend the practical transmission window of a silica based fibre into the 3 µm region [2].

In this paper the two different designs are compared with regards to the attenuation, bending losses, output beam quality and high energy handling capability. Also a demonstration of tissue ablation using the HC-NCF is shown.

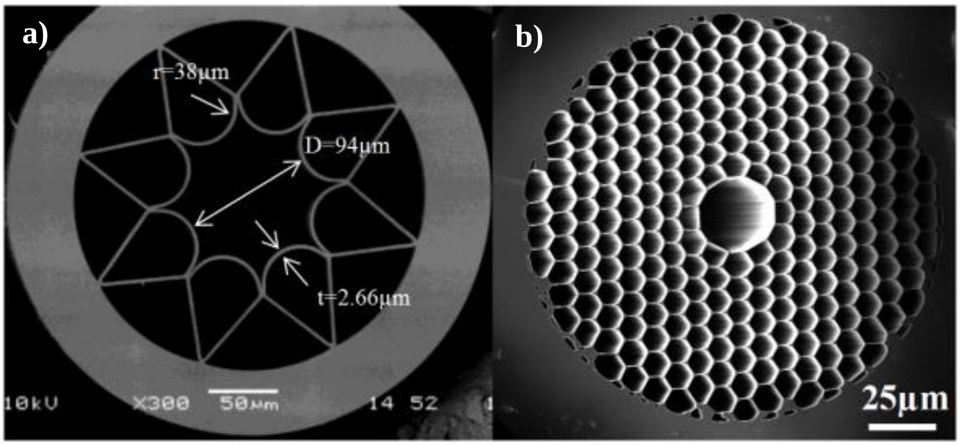


**Figure 2: SEM pictures of the hollow core fibres used for this work.**
**a) Negative Curvature Fibre (NCF) b) Photonic Crystal Fibre (PCF)**

## 2. MATERIALS AND METHODS

The laser used for the characterisation of both of the fibres was an Impex High Tech ERB 15 laser, with an operating wavelength of 2937 nm. The pulse width is 225 µs FWHM (Figure 3a) and the beam profile had a doughnut shape (Figure 3b).

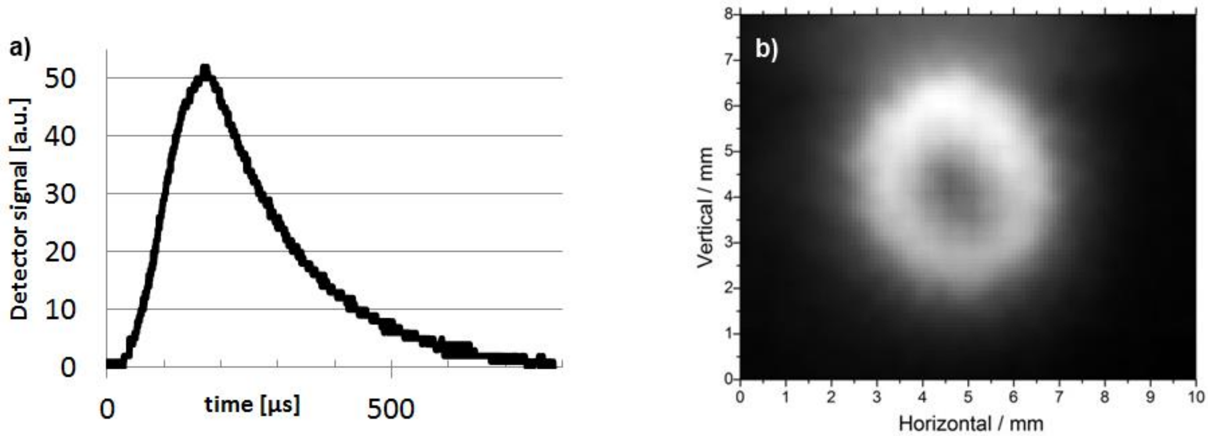


**Figure 3: Temporal (a) and special (b) profile of the Impex laser**

Both fibres were fabricated from fused silica (Suprasil F300) in a traditional stack and draw method. These designs are achieved by careful selection of the pressure difference between the core and the surrounding structure. As the light is mainly guided through air and the overlap of material with the light is small, the high intrinsic losses of the material of ~50 dB/m [2] can be overcome and low loss guidance at 2.94 µm is possible [16].

Both fibres have a low attenuation band around 2.94 µm with losses of 0.06 dB/m and 1.2 dB/m, for the NCF and the PCF, respectively. The NCF has a broadband guidance over two regimes, from 2 µm to 2.5 µm and from 2.8 µm to 3.8 µm and is shown in Figure 4a. This loss measurement was obtained using a cut back method, from 83 m to 3 m. As can be seen from the graph the minimum attenuation is 0.034 dB/m at a wavelength of 3.05 µm. The core size of the fibre is 94 µm. The inner wall has a curvature of 38 µm and a thickness of 2.66 µm (Figure 2a). The NA of the fibre is 0.03.

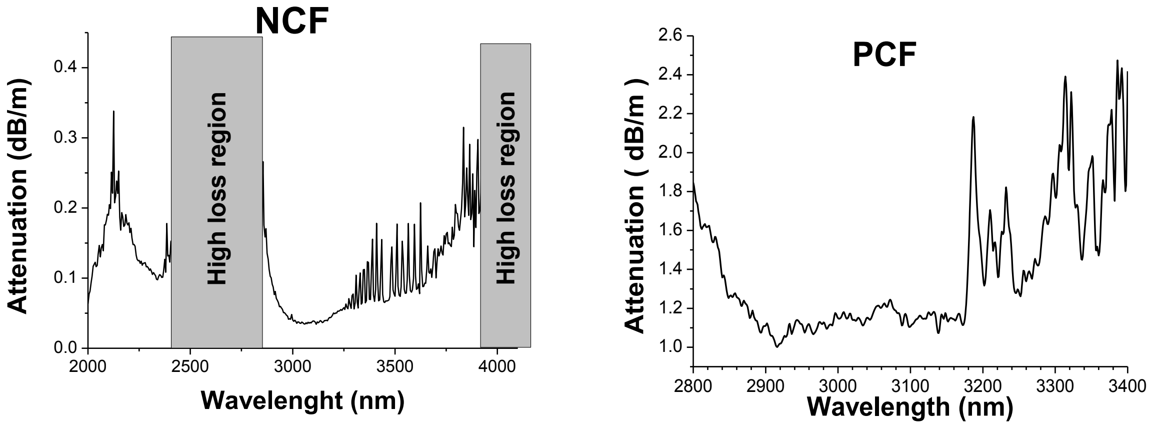


**Figure 4: Attenuation of the a)** ***Negative Curvature fibre*** **and the b)** ***Photonic Crystal Fibre***

The attenuation of the PCF is shown in Figure 4b, and as can be seen the average losses are ~1.2 dB in the regime from 2.85 µm to 3.18 µm. Again attenuation was measured using a cut-back technique. The core size is 24 µm with a pitch (distance between two surrounding holes) of 7 µm. The NA of the fibre was measured as 0.12.

In both cases no particular care was taken during fabrication to minimise the OH level.

## 3. BEND SENSITIVITY

Both fibres were investigated with respect to their bend sensitivity. A 1.23 m long piece of the NCF was bent through 180° with different bend diameters. The additional losses arising are shown in Figure 5 as a function of wavelength.

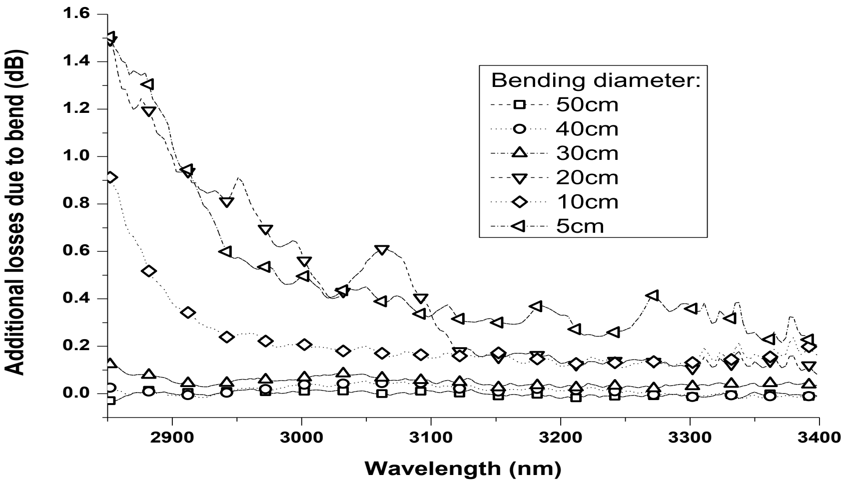


**Figure 5: Additional losses due to a 180° bend in a 1.23 m long NCF.**

These measurements were carried out using a tuneable laser system (M Squared Ltd. Firefly-IR). There are no significant additional losses if the fibre is bent to a diameter of >30 cm. For a bend diameter of 20 cm or less the bend loss significantly increases, in particular at shorter wavelengths. However, for wavelengths shorter than 3.15 µm the loss for a 20 cm bend is higher than for a 10 cm bend. It is assumed that this unexpected behaviour is due to mode coupling from core to cladding modes. However a further investigation is necessary to prove this assumption, which could be done by analysing the output beam profile for shorter wavelengths.

The bend sensitivity of the PCF was only measured at 2.94 µm and no change in output power could be detected down to a bending diameter of 5 mm. The PCF fibre is extremely flexible and the fibre will not physically break until it is bent to a diameter <3 mm.

## 4. OUTPUT BEAM PROFILE

As discussed above a spatial beam profile that does not vary when the fibre is moved is strongly preferred for surgical procedures, particularly if the fibre is not to be used in contact mode. Therefore the output beam profiles of the NCF and PCF were measured.

The PCF was measured by reimaging the fibre end onto a ceramic surface and capturing the reflection using a mid-IR camera (Electrophysics Cop., PV320 – L2E). The greyscale image and a cross section are shown in Figure 6.

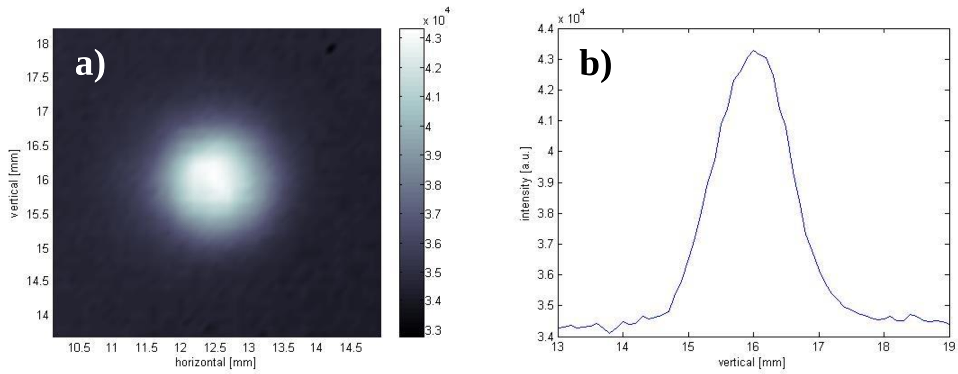


**Figure 6: Output beam profile of the PCF. a) Greyscale as a reflection on a ceramic surface. b) Cross section of the beam profile**

As can be seen the output beam profile is single mode like. The fibre was bent with a diameter of ~50 cm during this test, and the beam profile does not change if the fibre is moved. Also, the output power stays constant when the fibre is bent and moved. The length of the fibre in this instance was approximately 1 m.

The NCF output beam profile was measured by transversally moving the PCF at the end of the NCF. This way the average power of a ~ 450 µm$^2$ area (core size of the PCF is 24 µm) could be measured. The PCF was moved in 5 µm steps along the central line of the NCF. The output power of the PCF was measured using an Ophir detector (PE50-DIF-ER-SH-V2 ROHS). The cross section of the NCF output beam profile is shown in Figure 7.

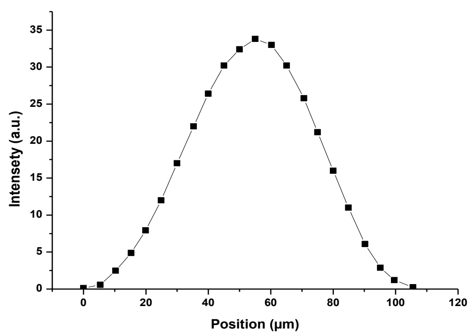


**Figure 7: NCF output beam profile for a bent (d~50 cm) fibre measured by moving a PCF transversally relative to the NCF.**

The output beam profile of the NCF, which was also bent to a diameter of around 50 cm, is single mode like. The output beam profile does not change if the fibre is moved, however the output power fluctuated which is to be expected due to the bend dependent loss of the fibre. The length of the fibre in this instance was 80 cm.

## 5. HIGH ENERGY DELIVERY AND TISSUE ABLATION

The energy handling capabilities of the two fibres was tested using the Impex Er:YAG laser described in section 2. The coupling efficiency into both fibres was relatively low with 35% for the NCF and just 5% for the PCF, respectively. It is assumed that this is a direct result of the relatively poor mode overlap of the laser beam profile (doughnut shape) and the two fibre modes (Gaussian like).

The maximum pulse energy delivered through the NCF was 195±1 mJ for a 33 cm long fibre and 54±4 mJ for a 988 cm long fibre. The 33 cm fibre was held straight, whereas the 988 cm fibre was bent with a diameter of around 50 cm. Considering the core size of 94 µm these pulse energies translate to energy densities directly at the output of the fibre, of 2300 J/cm$^2$ for the shorter length of fibre and 764 J/cm$^2$ for the longer length, respectively. It should be noted that the maximal output was limited by the source; the fibre was undamaged at both the launch and output ends in all experiments.

The maximum output energy delivered through a 44 cm long PCF was 14 mJ. Using the core diameter of 24 µm this translates into an energy density of 3465 J/cm$^2$ at the fibre output. At higher energies the launch end of the fibre was catastrophically damaged, with the core surrounding structure destroyed. However, the output end facet of the fibre was undamaged and looked identical to a freshly cleaved end as seen in Figure 2b. This gives confidence that if improved coupling can be achieved the performance of the fibre could be increased.

It is important to point out that the maximal energy density needed for human tissue ablation is 35 J/cm$^2$ [18]. This is the threshold needed to ablate human dental enamel. As the results above demonstrate this threshold has been exceeded by a minimum factor of >21.

A demonstration of tissue ablation was done using porcine bone. The fibre was sealed by a sapphire endtip, as was reported in [19]. This is necessary to protect the fibre from debris and liquids the tests were carried out in air and with the fibre submerged in water. Also the fibre was used both in contact and non-contact mode. Two examples of tissue ablation can be seen in Figure 8. In both case a single shot was used.

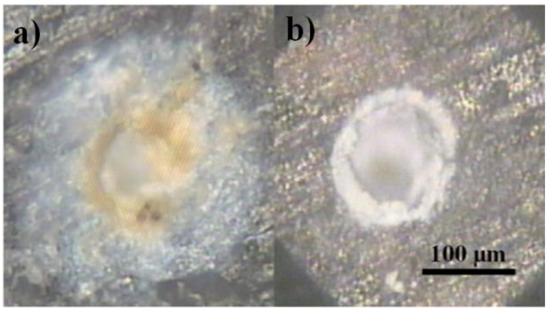


**Figure 8: Demonstration of porcine bone ablation. a) Single shot in air. b) Single shot under water**.

## 6. CONCLUSION

Two different fibre designs were presented, the Hollow Core Negative Curvature Fibre and a Hollow Core Photonic Crystal Fibre. Both fibres are fabricated from fused silica and guide around the surgically important wavelength of 2.94 µm, with attenuations of 0.06 dB/m (NCF) and 1.2 dB/m (PCF). The fibres both guide light in a single mode and are relatively bend insensitive. Both fibres exceed the threshold needed for human tissue ablation by a factor of at least 21. The viability of using the output of these fibres for surgery was demonstrated with the ablation of porcine bone in different environments. Ultimately these fibres provide an alternative to established delivery systems, and could pave the way for novel minimal invasive surgical procedures using Er:YAG laser systems.

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