Published in final edited form as:

Proc SPIE. 2013 March 8; 8565: . doi:10.1117/12.980912.

3 μm CW lasers for myringotomy and microsurgery

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Abstract

This paper describes the development and implementation of 3 μm lasers for myringotomy and microsurgery. Two different lasers were investigated. The first, an Er-doped, CW zirconate glass fiber laser optically pumped by a 970 nm diode laser, emitted > 1 W of CW power at 2.76 μm with concomitant green incoherent emission that served as a convenient visible illumination beam. The second, a 1 W CW Er:YAG solid-state laser also optically pumped by a 970 nm diode laser, emitted > 1 W of CW power at 2.94 μm , coincident with the strongest infrared water absorption peak. Running CW, both lasers are expected to avoid the loud acoustical shocks associated with pulsed lasers. Myringotomies were carried out with the Er:YAG laser on anaesthetized guinea pigs and the effects of the laser were documented. Laser ablated samples of tympanic membrane, soft tissue and bone were histologically examined. Histology results indicated that the CW Er:YAG laser is a potential candidate for a new myringotomy tool and possibly for otologic microsurgery, but deliverable power levels need to be increased to the 2 W (or higher) level. This work was funded under NIH SBIR Grant No. 5R44DC004899.

Keywords

Mid-IR lasers; myringotomy; microsurgery; Er:YAG laser; zirconate glass laser

1. INTRODUCTION

The objective of the work described here was to design, build and test a portable laser for myringotomy applications that overcame the loud pop noise previously observed with pulsed Er and superpulsed CO₂ lasers.¹ Since tissue is characterized by high water content, lasers emitting at a wavelength coincident with strong water absorption peaks were considered to be favorable because of their reduced optical emission power requirement. Furthermore, since the previously observed loud pop noises were associated with pulsed laser operation, lasers capable of operating in a continuous wave (CW) mode were sought. Such CW lasers can be operated with temporally shapeable pulses that can be designed to have gradual rise and fall times so as to eliminate the pop noise thought to be caused by laser ablation shock waves, which have the potential for causing acoustic trauma with sensorineural hearing loss. Specifically, lasers emitting at ~ 3 μm were investigated because, other than the deep-ultraviolet (< 150 nm) spectral region where convenient lasers are not currently available, this is the spectral region where water has its maximum absorption, as illustrated in Figure 1.

The two laser candidates that were investigated were (a) an Er-doped zirconate glass fiber laser emitting near 2.76 μm and (b) an Er:YAG solid state laser emitting near 2.94 μm . In

each of these laser systems the Er ion forms the optically active system whose laser emission is based on the absorption of 970 nm optical pump photons which excite electrons from the Er ion $^4I_{15/2}$ ground state into the $^4I_{11/2}$ excited state from which the electrons, in the presence of a resonant optical cavity, experience stimulated downward transitions into the $^4I_{13/2}$ lower level manifold. The electrons are rapidly depleted from this lower level manifold by transitioning into the ground state, thus enabling the population inversion required for laser action. In the case of the Er-doped zirconate glass fiber lasers, the resulting stimulated emission occurs near 2.76 μm , while in the case of Er:YAG solid-state medium the stimulated emission occurs near 2.94 μm . Details of these two laser systems are summarized in the next two sections, while the results of the laser myringotomy experiments are presented in the section after that, followed by a final summary and conclusions section.

2. ZIRCONATE GLASS FIBER LASER DESCRIPTION

The stable heavy metal fluoride glass fiber system $\text{ZrF}_4\text{-BaF}_2\text{-LaF}_3\text{-AlF}_3\text{-NaF}$ (ZBLAN, zirconate glass) has been extensively studied and developed. This glass has been shown to be a good host for rare earth ions which exhibit favorable infrared laser properties. The first watt-level zirconate glass fiber lasers consisted of an Er-doped fiber core surrounded by a double-clad fiber configuration with the inner pump cladding region of rectangular cross section surrounded by a round cross section outer cladding region. In this fiber laser system the inner pump cladding region was optically pumped by a commercially-available 970 nm diode laser.³ The 970 nm pump light leaks into the Er-doped optically active fiber core region by evanescent coupling along the entire length of the fiber. Such distributed optical pumping minimizes localized heating and enables the attainment of watt-level fiber laser output power. More recently zirconate glass fiber lasers were reported to exhibit CW optical output power levels as high as 9 W at a wavelength of 2.78 μm .⁴ This power level was one of the highest reported for this laser system, but its attainment was accompanied by intense power fluctuations followed by severe optical damage at the pump end of the fiber laser. It was clear that such fiber lasers, when used in industrial or medical procedures, would need to be operated at much lower power levels to avoid operational failures. Spire reported zirconate glass fiber lasers emitting at the 1 W optical output power level while simultaneously emitting strong 540 nm green light (incoherent and resulting from electrons exhibiting spontaneous transitions from excited upper-level states to the glass ground state) that is useful as a visible pointing beam.⁵ The fiber laser was optically pumped at one end, while the 2.76 μm laser light together with the green pointing beam light emanated from the other end of the flexible (~ 0.5 mm dia.) fiber, providing a convenient means of laser light delivery. Such simultaneous IR/visible light delivery eliminated the need for the more complex optics (including optically-lossy beam splitters) normally used for visible laser pointing beams in flexible fiber laser delivery systems.

Whereas optical output power levels of 2 W were obtained from this zirconate glass fiber laser, higher optical pump power levels typically resulted in melting of the pump ends of the fibers. The occurrence of these fiber end melting effects corresponded to incident pump power density values of ~ 30 kW/cm². Attempts at coating the fiber ends as well as use of different fiber end polishing techniques did not result in significant improvements. Thus, for reliable operation, it was found necessary to stay well below the 2 W output power level. Based on the water absorption spectrum shown in Figure 1, the optical absorption of water at 2.76 μm is over an order of magnitude less than that at 2.94 μm . Thus the efficacy of an Er:YAG laser operating at the 1 W level is expected to be an order of magnitude better than that from a 1 W zirconate glass fiber laser.

3. Er:YAG SOLID-STATE LASER DESCRIPTION

The Er:YAG laser used in this research is a solid-state laser emitting at 2.94 μm , whose lasing medium consists of crystalline yttrium aluminum garnet ($\text{Y}_3\text{Al}_5\text{O}_{12}$) doped with erbium atoms.⁶ The Er^{3+} ions can be optically excited with 970 nm laser light, similar to that used for pumping the zirconate glass laser. Er:YAG lasers are not as widely used as Nd:YAG laser because of the strong and broad water absorption peak near 3 μm . On the other hand, because of this strong water absorption band, Er:YAG lasers have been used for a number of medical applications including skin resurfacing, wart removal, treatment of acne and melasma, bone and dental cutting and soft tissue treatment.⁷ Bone surgery applications for otolaryngology include stapedectomy, for which argon ion and diode lasers have previously shown promise.⁸ Er:YAG lasers have generally been operated in a pulsed mode because higher output power levels can be obtained under pulsed conditions. The thermal conductivity of the family of garnet crystal materials is such that severe crystal heating occurs at high optical pump power levels, with the thermal conductivity typically decreasing by about 40% as the crystal temperature increases from 25° C to 100° C. This, in turn, can result in thermal lensing effects and other undesirable thermal effects. Under pulsed conditions it is the average power that determines the crystal temperature and with low duty factor operation laser output power levels of 10s of W can be obtained. However, under CW conditions the output power from Er:YAG lasers is limited to the ~ 1–2 W level, due to heating effects.

The portable CW Er:YAG laser used in the present experiments was manufactured by Sheumann Laser, Inc. The laser could be configured to either emit through a hollow metallized glass tube which was inserted into a stainless steel sheath optical fiber delivery tip⁹ or via a lensed direct light wand delivery output. The laser was optically pumped with a 970 nm diode laser and nominally rated for ~ 1 W CW output power at 2.94 μm . A photograph of the laser with the fiber delivery tip attached to the end of an otoscope is shown in the left side of Figure 2. In this configuration ~ 30–40% of the laser power was lost in the fiber delivery tip, resulting in a useful output power of ~ 600 – 700 mW. A photograph of the laser fitted with a lens for direct light wand delivery at ~ 1 W of optical output power is shown on the right side of Figure 2 as pre-positioned for use on one of the anaesthetized guinea pigs.

The 2.94 μm , direct light wand delivery Er:YAG laser as shown in Figure 2, right, was operated at a CW output power level of 1 W for durations of 1 s per CW interval from a distance of 20 mm from each tissue target. The pulse turn-ons were controlled by a foot pedal switch. A diagram illustrating the ~ 2-cm long Gaussian beam profile zone emanating from the laser delivery wand is shown in Figure 3. The laser beam diameter was ~ 0.45 mm in this zone.

Prior to histological analysis, a macro analysis of the visible laser tissue impact damage was performed by measuring its 2-dimensional extent on a micrometer scale by use of a Leica dissecting microscope at 5–10 x magnification. A description of the myringotomy procedure and results is given in the next section.

4. MYRINGOTOMY DESCRIPTION AND RESULTS

The animal experiments were carried out at the Boston Children's Hospital (BCH). The appropriate animal (guinea pig) protocol was reviewed by veterinarians and by the animal care committee at the Children's Hospital and received official approval prior to commencement of the experimental work described here.

4.1 Experimental Methodology

The animal models included 5 guinea pigs ranging in weight from 660 g to 725 g. The animals were anaesthetized with Ketamine (0.25 ml, 100 mg/ml), Xylazine (0.06 ml, 20 mg/ml), and Isoflurane inhaled (1–1.5%). On anaesthetized animals, 3 adjacent laser burns were carried out on the following tissues: skin, muscle (m. temporalis), bone (skull bone), and a single burn was made through the ear canal to impact the tympanic membrane (TM). The CW Er:YAG laser at the 1 W power level, 1s exposure, 20 mm distance from the target tissue was used for each laser exposure. Exposures were first made on the skin, then a post-auricular incision was made on the right side, exposing the temporalis muscle and squamous portion of the temporal bone. Laser burns were made on these tissues. The ear canal was then transected at the level of the bony meatus and with guidance from the operating microscope, the final laser exposure was made onto the TM. After euthanizing the animals, the treated tissues were harvested for fixation and histological assessment. Temporal bones were harvested whole and the cochleas perfused with Bouin's solution. The histology samples included skin of the neck, temporalis muscle, squamous portion of the temporal bone, and the temporal bone including the TM and inner ear and corresponding untreated tissues from the contralateral side (control).

The histology was performed by BCH-Histology Core as follows:

- a. Fixation in Bouin's Fixative
- b. Removal of picric acid in ddH₂O and washes of graded ETOH (50%)
- c. The samples were stored for further processing in 70% ETOH
- d. Bone samples and Temporal bones were decalcified using hydrochloric acid/formic acid solution, neutralized in ammonia solution and chemically tested (ammonium hydroxide/ammonium oxalate solution) for complete decalcification.
- e. Samples initially were prepared for Paraffin embedding, and sectioned under a microtome. Since the tissue quality did not allow proper sectioning due to overly hard tissue, it was deparaffinized, rehydrated and embedded in OCT (Tissue Tek) for sectioning under a cryotome. Tissue sections of 20 µm thickness were achieved. Section quality was not optimal, but substantially better than Paraffin sections.
- f. Slides were stained with Hemotoxylin and Eosin (H&E) and studied under light microscopy.

4.2 Experimental Findings from Histology

This section deals with the macroscopic examination of the laser irradiated bone, skin and muscle tissue. The 2-dimensional diameter of the visible laser tissue evaporation zone in all samples (bone, skin and muscle) was nearly round. Slight elliptical measurements resulted from movement artifacts which are due to hand holding the laser wand, as well as breathing of the live test animal. The following three sub-sections describe the surface burn zones for bone, skin and muscle as observed with Leica dissecting microscope at 5–10 x magnification.

Bone (skull):	Sample 1:	Lost due to tissue processing	
	Sample 2:	Laser burn surface diameter:	x-axis: 0.83 mm, y-axis: 0.67 mm
	Sample 3:	Laser burn surface diameter:	x-axis: 0.42 mm, y-axis: 0.87 mm
	Sample 4:	Laser burn surface diameter:	x-axis: 0.66 mm, y-axis: 0.50 mm

Skin:	Sample 5:	Laser burn surface diameter:	x-axis: 0.73 mm, y-axis: 0.69 mm
	Sample 1:	Lost due to histological processing	
	Sample 2:	Laser evaporation diameter:	x-axis: 0.67 mm, y-axis: 1.13 mm
	Sample 3:	Laser evaporation diameter:	x-axis: 1.2 mm, y-axis: 1.1 mm
	Sample 4:	Laser evaporation diameter:	x-axis: 1.24 mm, y-axis: 1.38 mm
Muscle (M. temporalis)	Sample 5:	Laser evaporation diameter:	x-axis: 0.73 mm, y-axis: 0.72 mm
	Sample 1:	Lost due to Paraffin processing issues	
	Sample 2:	Laser burn surface diameter:	x-axis: 1.22 mm, y-axis: 1.29 mm
	Sample 3:	Laser burn surface diameter:	x-axis: 1.08 mm, y-axis: 1.26 mm
	Sample 4:	Laser burn surface diameter:	x-axis: 1.1 mm, y-axis: 0.76 mm
	Sample 5:	Laser burn surface diameter:	x-axis: 1.1 mm, y-axis: 0.99 mm

Photographs of representative samples from the above 3 groups of tissue laser burns are shown in Figure 4.

More complete 3-dimensional examinations of the above-described laser ablated tissue were carried out. The laser tissue impact histology and microscopy data were obtained. The terminology used to describe the histology findings is illustrated in Figure 5, where the immediate laser tissue impact zone is the evaporation zone which is followed by the char zone (a thin layer of tissue char) and the secondary laser impact zone (tissue necrosis) followed by the usually wider outer thermal damage zone.

The histologic samples of bone, skin and muscle showed an immediate tissue evaporation crater upon H&E staining which was usually lined by tissue char (char zone). The necrosis zone results from the indirect yet severe impact of the laser on cellular and extracellular structures leading to severe tissue damage such as tissue necrosis and coagulation. The secondary laser-tissue impact zone causes thermal damage beyond the necrosis zone and can reach far into the tissue depending on the tissue type and laser conditions. In general, denser tissues show less laser impact damage. In descending order of the tissue used, bone is densest, followed by skin and then skeletal muscle.

The following tables and associated microscopic images describe the observations resulting from the above-described histology examinations. Out of 5 tissue samples for each tissue type, only 3 could be analyzed histologically.

Bone (skull bone) showed the least amount of tissue evaporation damage. This is most likely due to the density of the tissue as well as the water content of bone in general. The laser radiation associated necrosis was extensive. Heat damage appeared to be less compared to softer tissues. Table 1 summarizes the measured dimensions of the laser impact crater regions of skull bone tissue for laser exposures of 1W at 1s.

Fig. 6 shows the histological microscopic cross section of the bone tissue from which the Table 1 was obtained.

Table 2 summarizes the dimensions of the skin tissue laser impact craters, CW, 1 W, 1 s exposures.

The histological microscopic cross section of the skin tissue from which the Table 2 data were obtained is shown in Figure 7.

The thermal damage zone reached well into the dermis and hair follicle area. The heat damage associated with the laser radiation causes immediate structural damage to the cellular and extracellular matrix proteins. At this stage additional Eosin gets incorporated into the collagen fibers of the dermis. Cellular nuclei are reduced. This indicates that the laser damage reached at least about 150 μm further than the immediate burn depth, thereby heat damaging tissue structures beyond the visible burn crater. The extracellular and cellular structure of the dermis is denser than that of muscle, but less than that of bone. This is reflected in the overall laser damage to the tissue.

Table 3 summarizes the measured dimensions of the laser impact crater regions of M. temporalis muscle tissue for CW laser exposures of 1W and 1s exposure times.

The histological microscopic cross section of the muscle tissue from which the Table 3 data were obtained is shown in Figure 8.

In the skeletal muscle tissue the laser radiation damage was extensive. Skeletal muscle containing a high concentration of water and an overall less dense matrix showed a deep and wide impact crater, as well as far reaching thermal damage.

4.3 Microscopic Examination of Tympanic Membrane, Middle Ear and Inner Ear

The laser effects on the temporal bone inner ear wall and tympanic membranes were noted macroscopically with a TM perforation resulting in each case and the burn usually impacting the promontory of the middle ear. Laser ablated samples of tympanic membrane and tympanic bone were examined histologically. The opposite side untreated controls all showed normal macroscopic and histological exams with no evidence of TM perforation and no TM, middle ear or inner ear thermal injury. Five laser treated tissue samples with external auditory canal, middle ear and inner ear intact were studied.

Figure 9, left shows the intact tympanic membrane with malleus, while the right figure shows an intact tympanic membrane and intact cochlea.

Figure 10 shows the tympanic membrane impacted by the laser and shows the impact of the laser beam on the promontory/tensor tympani muscle below.

The heavy red stain on the left side of Figure 10 is blood collected into the tympanic cavity, most likely due to harvesting artifact. The laser burned through the tympanic membrane largely with vaporization and there was some substantial secondary damage (stasis zone) at the perforation periphery, as shown in Figure 11 left and right.

Based on histological examination there appeared to be no evidence for structural damage to the cochlea in any of the specimens. However, a definitive assessment for the possibility of functional injury would require electrophysiological studies that were not included in the present work. The laser tissue impact leads to major thermal damage of soft tissue, including the tensor tympani muscle when it was impacted. However, the thermal damage to the promontory of the cochlea appeared to be limited to the superficial layers and never reached the endosteum of the cochlear lumen. No structural injury to the cochlea was observed.

5. SUMMARY AND CONCLUSIONS

Continuous wave (CW) double-clad zirconate glass fiber lasers with Er-doped fiber cores were fabricated and produced nominal 1 W CW output power at 2.76 μm . While CW power levels of up to 2 W were obtained, the fiber lasers exhibited burnout at the pump ends of the fibers and the myringotomy experiments were carried out with a more reliable 1 W Er:YAG solid state laser produced by Sheumann Laser, Inc. At a wavelength of 2.94 μm , the

Er:YAG laser light absorption by water-containing tissue is expected to be an order of magnitude stronger than at 2.76 μm . Myringotomy experiments were carried out with the CW Er:YAG laser on anaesthetized guinea pigs, and the effects of this laser light delivered in the form of a free-space 0.45 mm diameter laser beam were measured. Laser ablated samples of tympanic membrane and tympanic bone were histologically examined. The 1 W power level was high enough to achieve repeatable tympanic membrane ablation, but there was evidence of some thermal damage to middle ear and inner ear structures. No evidence for cochlear injury was observed other than superficial thermal damage to the promontory. The 1 W level laser light was also shown to create craters in bone, suggesting potential applications in microsurgery involving bone tissue. Whereas final determination of laser ablation efficacy requires additional study, the preliminary results obtained here suggest that the CW Er:YAG laser is a potential candidate for a new myringotomy tool without the undesirable “pop” noise associated with pulsed lasers, and possibly as a new tool for bone microsurgery.

Acknowledgments

This work was funded under NIH SBIR Grant No. 5R44DC004899.

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Figure 1. Spectral absorption of liquid water as a function of wavelength.² Other than the deep ultraviolet spectral region at < 150 nm, the strongest water absorption peak occurs at ~ 3 μm (3000 nm).

Figure 2.

Left: photograph of the Er:YAG laser fiber delivery tip attached to an otoscope Right: The Er:YAG laser with light wand near an anaesthetized guinea pig

Figure 3. Schematic of the Er:YAG laser light delivery wand illustrating the ~ Gaussian beam profile of the exiting $2.94\ \mu\text{m}$ laser light.

Figure 4.

Microscope photographs of representative samples of laser-burned bone tissue (left) with the blue line pointing to the laser impact crater, skin tissue (middle) with the blue line pointing to the laser impact crater, and muscle tissue (right). Laser power ~ 1 W, exposure times 1 s.

Figure 5. Diagram illustrating the terminology used to define the four zones of laser ablation effects. The laser beam is incident on the tissue in a vertically downwards direction.

Figure 6. Histology cross section of guinea pig skull bone exposed to a 1W CW Er:YAG laser beam of approximate diameter 0.45 mm for 1s.

Figure 7.
Histology cross section of guinea pig skin exposed to a 1W CW Er:YAG laser beam of approximate diameter 0.45 mm for 1s.

Figure 8.
Histology cross section of guinea pig M. temporalis muscle tissue exposed to a 1W CW Er:YAG laser beam of approximate diameter 0.45 mm for 1s.

Figure 9.
Left: Cross-section photograph of an intact tympanic membrane with incus.
Right: Cross section of an intact tympanic membrane and intact cochlea.

Figure 10.

Tympanic membrane impacted by the laser showing the impact of the laser beam on promontory/tensor tympani muscle.

Figure 11.

Left: Cross section of middle ear and promontory of cochlea histo-photograph showing char and necrosis – secondary impact zone at the margin of the TM vaporization site. There is no injury seen on the promontory.

Right: Cross section of middle ear and promontory of cochlea showing vaporization (evaporation) and necrosis zone and thermal damage zone in both the TM and a zone of thermal injury in the superficial surface of the promontory. No thermal injury is seen extending into the lumen of the cochlea.

Table 1Measurements of laser impact on skull bone tissue (dimensions in μm)

	Bone Sample 1	Bone Sample 2	Bone Sample3
Evaporation zone diameter (Mean)	820.4	854.2	833.9
SD (Standard Deviation)	11.9	13.1	5.7
Evaporation zone depth (Mean)	30.7	28.9	22.5
SD	4.2	5.4	5.1
Char	7.9	5.5	9.52
SD	1.3	0.9	1.9
Necrosis zone	47.9	57.9	57.1
SD	3.9	6.7	3.9
Thermal damage zone	18.1	19.1	12.6
SD	4.7	2.7	0.8

Table 2

Measurements of laser impact on skin tissue (dimensions in μm).

	Skin Sample 1	Skin Sample 2	Skin Sample 3
Evaporation zone diameter (Mean)	861.4	917.2	874.7
SD (Standard Deviation)	7.7	10.5	9.9
Evaporation zone depth (Mean)	341.4	300	334.7
SD	12.1	26.6	25.4
Char	25.2	24.2	24.3
SD	3.4	5.2	7.2
Necrosis zone	45.5	39.7	50.1
SD	12.3	4.7	7.9
Thermal damage zone	148.3	150.1	156.4
SD	19.1	30.1	14.9

Table 3

Measurements of laser impact on muscle tissue (dimensions in μm).

	Skeletal Muscle Sample 1	Skeletal Muscle Sample 2	Skeletal Muscle Sample 3
Evaporation zone diameter (Mean)	1003	1052.9	1134.7
SD (Standard Deviation)	39.8	36.15	67.8
Evaporation zone depth (Mean)	689.1	653.6	702.5
SD	57.2	67.2	145.4
Char	38.2	44.5	49.3
SD	9.8	10.4	11.9
Necrosis zone	120.7	101.1	140.9
SD	19.5	13.8	23.5
Thermal damage zone	371.2	364.2	362.7
SD	59.1	47.9	60.1