# Mid-IR Laser Tissue Ablation with Little Collateral Damage Using a Laser Tunable in the Water Absorption Peak

**D. Chuchumishev[1, 2], E. Nagel[1], A. Nierlich[1], S. Philipov[3], T. Genadiev[3], T. Fiebig[1], I. Buchvarov[1, 2], C.-P. Richter[1, 4, 5]**

*[1]Deptartment of Otolaryngology, Northwestern University, 303 E. Chicago Ave, Searle 12-561, Chicago, IL 60611, USA*
*[2]Department of Physics, Sofia University, 5 James Bourchier Blvd., Sofia, 1164, Bulgaria*
*[3]Department of Medicine, Sofia University, 1 Kozyak Str., Sofia, Bulgaria*
*4Department of Biomedical Engineering, Northwestern University, 2145 Sheridan Road, Tech E310, Evanston, IL 60208, USA*
*[5]The Hugh Knowles Center, Department of Communication Sciences and Disorders, Northwestern University, Evanston, IL 60208, USA*
*Author e-mail address: cri529@northwestern.edu*

**Abstract:** A comprehensive experimental study of mid-IR laser tissue ablation within the water absorption peak is presented. A novel all-solid-state table-top sub-ns mid-IR laser designed for efficient tissue ablation have been used for observation of wavelength-dependent effects on the ablation of hard and soft tissue.

**OCIS codes:** (170.1020) Ablation of tissue; (170.3890) Medical optics instrumentation

## 1. Introduction

An ultimate goal of minimally invasive laser surgery is to incise or remove a defined volume of tissue by laser ablation, while leaving the adjacent tissue biologically viable. Besides, minimizing collateral damage to adjacent tissue structures is vitally important when ablating in close proximity to nerve or other neural structures. The laser ablation characteristics and collateral damage effects depend on the optical absorption properties of the tissue and subsequent energy dissipation on distinct time scale. The optical absorption properties of tissue are governed by the electronic, vibrational, and rotational structures of the constituent biomolecules, thus they are dominated by the absorption of water, lipids and proteins, all of which show absorption bands in the mid-IR range. Tunable mid-IR lasers have an advantage over ultraviolet and visible or near-IR lasers. Water has sharp absorption peak around 3,000 nm, while lipids around. Changing the wavelengths of the laser would allow to fine-tune laser tissue interactions (penetration depth and density of the absorbed energy), minimizing the collateral heating effects in a tissue-specific manner. Hence, intense laser radiation tuned to specific absorption bands in this spectral region has the potential for carrying out selective ablation and is therefore of great relevance for advanced surgical procedures. In addition, tunable mid-IR lasers can optimize the laser tissue interaction for ablation and/or coagulation, i.e. the suppression of bleeding.

Here we demonstrate efficient tissue ablation using a novel all-solid-state table-top mid-IR laser tunable within the water absorption peak (3,000 to 3,500 nm) as a valuable alternative of mid-IR Free Electron Lasers (FEL). Comparative studies of the ablation process in different types of tissues and at different wavelengths across this biologically important spectral region have been carried out for the first time.

## 2. Experimental design

A sub-nanosecond, short cavity, singly resonant optical parametric oscillator (OPO) was constructed and its radiation was amplified in a highly efficient optical parametric amplifier (OPA), which was based on large aperture periodically poled stoichiometric lithium tantalate (PPSLT). The OPO uses a 20 mm long, 10 mm wide, and 3.2 mm thick PPSLT crystal with three poled zones with different domain inversion periods (30.2, 30.3 and 30.4 μm). The OPA stage employed a similar 37 mm long crystal. The pump source for the two frequency conversion stages was a diode-pumped master oscillator power amplifier system, providing 35 mJ pulses with high beam quality ($M_x^2 \times M_y^2 = 1.3 \times 1.1$), 1.4 ns pulse duration at 0.5 kHz repetition rate. The maximum output idler energy from the system was 5.7 mJ with a pulse duration of 1.4 ns and the idler conversion efficiency in the OPA stage was 19%. By changing the temperature of the two PPSLT crystals from 40 °C up to 265 °C and by employing the three domain inversion periods, continuous tunability of the laser from 3 to 3.5 μm was achieved.

The repetition rate of the laser pulses was 500 Hz and the selected radiant energy per pulse was 3.0, 2.9, and 2.3 mJ at 3, 3.32, and 3.47 μm radiation wavelength, respectively. The focused beam spot has and elliptic shape with dimensions of 350 × 250 μm and had a near Gaussian beam profile. By advancing the stage by 2 mm/s continuous groves were created in the tissue. During the irradiation, the sample surface was kept at the focal point of the laser beam.

Following standard histological protocols, the specimen were decalcified if necessary, dehydrated, and embedded in Araldite Resin. The embedded specimens were serial sectioned at 5 μm, mounted on glass slides, and stained with hematoxylin and eosin. Bright field images of selected sections were examined with a light microscope (Leica DMRB, Buffalo Grove, IL) after they were captured with a Spot Insight Color camera (SPOT™ Imaging Solutions, Sterling Heights, MI). Images were analyzed for tissue carbonization or loss of birefringent activity when examined under a pair of cross-polarizing filters.

## 3. Results

Tissue ablation was achieved at 3, 3.32, and 3.47 μm with single pulses and trains of pulses. For skin the depth of the ablation crater increased from 113 μm (Fig. 1A) to 250 μm when the radiation wavelength of the single pulses was changed from 3 to 3.47 μm. In contrast to skin, for cartilage the depth of the ablation crater decreased from 95 μm (Fig. 1B) to no ablation for the same changes in radiation wavelength.

Possible damages resulting from the irradiation were examined in bright field images (Fig. 1 *top row*) and with cross-polarizing filters (Fig. 1 *bottom row*). No carbonization was seen and the loss in birefringent activity was less than 20 μm. Little additional signs for thermal or mechanical damage could be detected. Note, the current examination and judgment is based on fixed cadaveric material, which has been examined with light microscopy only.

When the tissue samples were exposed to trains of pulses, different results were obtained. With increasing of the number of pulses delivered, the depth of the groove increased (data not shown). Also along the ablation grove a thin black line can be seen in the bright field image, indicating the carbonization of the organic tissue (Fig. 1C). Next to the carbonization line, a small tissue layer, about 10-20 μm thick, shows a small layer of darker staining. This is the layer, for which thermal damage could be detected. The damage could be verified by the loss of optical activity as judged from the loss of birefringence seen under cross-polarizing imaging conditions.

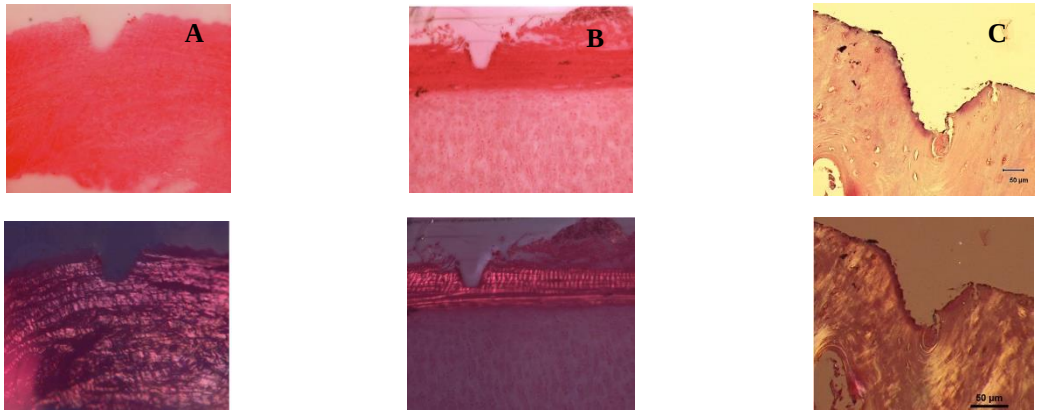


Fig. 1 Histological sections of tissue irradiated at 3μm: A – skin; B – cartilage of the pig pinna; C - bone. ***top row***: bright field images; ***bottom row***: imaging of the tissue with a pair of cross polarizing filters

## 4. Conclusion

A high energy and high average power laser, tunable in the mid-infrared region between 3,000 and 3,500 nm, has been designed and constructed. Hard and soft tissue ablation is achieved and at penetration depths below 20 μm the ablation was possible with little collateral damage. Our initial evaluations using hematoxylin staining of the exposed tissue sections and images with a cross polarizer confirmed the findings.

## 5. Acknowledgements

This work was funded by a grant from the NIH, 8R21EB015899 and in part by the Bulgarian National Science Fund under grants number DDVU 02/105/2010.