Journal of Biomedical Optics

SPIEDigitalLibrary.org/jbo

Bonding surgical incisions using a temperature-controlled laser system based on a single infrared fiber

Ilan Gabay
Irina Barequet
David Varssano
Mordechai Rosner
Abraham Katzir

SPIE

Bonding surgical incisions using a temperature-controlled laser system based on a single infrared fiber

Ilan Gabay,[a] Irina Barequet,[b] David Varssano,[c] Mordechai Rosner,[b] and Abraham Katzir[a]
[a]Tel Aviv University, School of Physics and Astronomy, Tel Aviv 69978, Israel
[b]Tel Aviv University, Goldschleger Eye Institute, Sheba Medical Center, Sackler School of Medicine, Tel Aviv 52621, Israel
[c]Tel Aviv University, Department of Ophthalmology, Tel Aviv Sourasky Medical Center, Sackler School of Medicine Tel Aviv 64239, Israel

Abstract. Although there has been great interest in laser heating for bonding of surgical incisions in tissues, it has not gained wide acceptance by surgeons. We argue that the main obstacle has been the lack of temperature control, which may lead to a weak bonding. We previously developed a laser bonding system based on two infrared transmitting AgBrCl fibers, one for laser heating and one for temperature control. In view of the inherent limitations of such systems observed in many animal experiments, we developed an improved system based on a single infrared fiber. Besides the decreased dimensions, this system offers many advantages over the two-fiber system. It is less sensitive to accuracy of height and tilt of the fiber distal tip above the tissue, ensuring more accurate heating that can potentially lead to stronger bonding with minimal thermal damage. The system is successfully tested in the soldering of 15 corneal incisions, *ex vivo*. Histopathology shows little thermal damage and good wound apposition. The average burst pressure is 100 ± 30 mm Hg. These findings indicate the usefulness of the system for ophthalmic surgery as well as other surgical procedures, including endoscopic and robotic surgery. [DOI: 10.1117/1.JBO.18.11.111416]

Keywords: infrared fiber; laser soldering; laser welding; temperature control; endoscopic surgery; cornea.

Paper 130227SSRR received Apr. 10, 2013; revised manuscript received Aug. 19, 2013; accepted for publication Aug. 22, 2013; published online Sep. 20, 2013.

1 Introduction

The common approach to the closure of surgical incisions is sutures. However, suturing techniques are often time-consuming, require considerable surgical skill, and do not always provide immediate watertight sealing, resulting in exposure to infection. In addition, the introduction of a foreign material into the body often results in a displeasing aesthetic appearance. Efforts have been made in the past few decades to overcome some of these drawbacks with laser tissue bonding (LTB) as an alternative to suturing. There are two common LTB methods: (1) laser tissue welding, in which the edges of a tissue incision are approximated and heated by a laser beam to a temperature of around 60°C; and (2) laser tissue soldering, in which the edges of an incision are approximated, a layer of some biological material is applied on top, and the area is then heated by a laser beam. LTB is inherently a sterile and nontactile technique that does not involve foreign material, such as a suture. The technique is expected to be easier to master and faster to perform than suturing, and should result in a watertight seal that will prevent infection. Wound healing is expected to be faster and to produce a more aesthetic result.[1]

The first clinical LTB attempts, which were carried out more than three decades ago,[2] involved the primary anastomosis of small to medium-size blood vessels. Since that time, attempts have been made to apply LTB in several clinical areas, including microsurgery,[3] ophthalmology,[4] and neurosurgery.[5] In all these cases, LTB was tested with a variety of lasers and laser irradiation conditions. Spreading a biological solder, such as albumin or chitosan over the incision prior to the laser treatment significantly improved the bonding strength.[6,7] Despite extensive research and numerous clinical trials, widespread clinical acceptance remains elusive. We attribute this lack of acceptance to the fact that the technique does not include temperature control, and any over- or under-heating can result in weak bonding.

The development of temperature-controlled laser welding of tissues, undertaken by the Applied Physics Group at Tel Aviv University[8], began with a two-fiber system. The basic idea was to approximate the edges of a cut and to heat a spot on it using a CO_2 laser (10.6 μm). The heated spot emitted middle infrared (mid-IR) radiation ("blackbody" radiation) whose intensity "I" was proportional to the temperature "T" of the spot. The intensity was measured by an infrared detector, which generated a voltage "V" that was read by a personal computer. A dedicated computer program used the value of V to control the power emitted by the CO_2 laser, so that the temperature T was kept constant for time t. The laser beam was then moved to a neighboring spot and again heated to temperature T for time t. It was found that if each spot was heated to $T \approx 60$°C for $t \approx 10$ s, a strong bonding was obtained.[9] The technique was employed to laser solder incisions in various tissues and animal models.

A major advance in laser soldering was the incorporation of optical fibers in the system. The standard optical fibers used in medical applications and communications are totally opaque in the mid-IR in the spectral range 3 to 20 μm. Consequently, there was a need for specialized optical fibers made of AgClBr, which were transparent in the mid-IR. Two such fibers were developed by us[10] and used in our original LTB system. One transmitted the CO_2 laser radiation for heating a spot, and the other transmitted the "blackbody" IR radiation emitted from the heated

Address all correspondence to: Ilan Gabay, Tel Aviv University, School of Physics and Astronomy, Tel Aviv 69978, Israel. Tel: ++972-3-640-8301; Fax: ++972-3-6415858; E-mail: gabayila@post.tau.ac.il

0091-3286/2013/$25.00 © 2013 SPIE

spot onto a suitable IR detector. These fibers were flexible, insoluble in water and bio-compatible. In a typical system, the two distal tips of the fibers were held in a hand-piece, making it easier for the surgeon to carry out the procedure. This system made it possible to heat a 3-mm diameter spot on the tissue to a desired average temperature T, at an accuracy of roughly 3°C. The system enabled LTB under accurate temperature control.

Despite 20 years of experience applying LTB to different types of tissues *in vitro* and *in vivo*,[11–13] we were not able to determine the optimal conditions for LTB, since each study was carried out with a different hand-piece fit with a different orientation of the fiber tips. During that period the drawbacks of the two-fiber system became apparent. The physical dimensions of the fiber holder prevented it from being inserted through natural body openings, necessitating stopping the experiments in some disciplines.[11] Inaccurate temperature measurements may have led to thermal damage from overheating in other disciplines.[13] As noted above, bonding incisions by laser heating requires heating each spot in the incision area to approximately 60°C. In the case of laser soldering, the solder (e.g., albumin) must be heated to a slightly higher temperature to ensure that both the surface of the incision and the part below the surface are sufficiently heated, creating a stronger bond. On the other hand, vulnerable organs, such as the cornea and small blood vessels, are highly sensitive to heat and can be damaged by overheating. Thermal damage to the cornea may coagulate it and change its curvature, resulting in irregular astigmatism that can potentially affect visual acuity.

It became clear that our two-fiber laser bonding system may have led to thermal damage and weak bonding, and that a system would have to be designed that generates strong bonding while avoiding thermal damage. The following structural and physical properties of the two-fiber laser bonding system may have led to thermal damage and weak bonding.

(i) *The geometry of two fibers*: The system was designed so that the distal tips of the two fibers would have a mutual focus a few millimeters below the distal tip of the holder. Under these conditions, the temperature reading would be correct. However, if the distance between the holder and the tissue changed (if, for example, the surgeon's hand moved), sensing would not take place at the center of the heated area, resulting in an inaccurate temperature reading.[14] This situation would worsen if the surgeon inadvertently tilted the hand-piece with respect to the bonded tissue. Such situations might be avoidable in the laboratory but not in the operating room.

(ii) *Temperature distribution*: If the measured surface has a uniform temperature, then the temperature reading is correct; otherwise, the measured temperature is actually an average of the temperature distribution on the surface. The intensity of a laser beam is bell-shaped (i.e., Gaussian), so when a spot on a surface is heated with a laser beam, the temperature at the center is usually much higher than at the periphery. This means that if one records a temperature of 50°C, which can theoretically cause reversible damage to the tissue, the reading is only an average value, whereas in fact the center of that temperature distribution could be tens of degrees higher than 50°C, and the tissue could be severely damaged at the center of the Gaussian laser beam.

In 1984, Artjushenko et al. published an article titled "Fiber optic device for simultaneous laser power transmission and temperature measuring of irradiated object."[15] Those researchers used an experimental setup that appears hard to align and likely to be sensitive to movement. This is probably the reason why these authors were unable to report successful stabilization of the object's temperature at a constant value. Some of their results showed heating to hundreds of degrees centigrade, while other results exhibited high noise. Artyushenko's later work[16] described a bi-directional mid-IR fiber-optic probe for spectroscopic applications. That paper described several practical optical setups that make it possible to use a single mid-IR fiber for transmitting excitation light and collecting emission light through a single fiber. Each setup was constructed from a simple arrangement of lenses and/or mirrors.

We speculated that similar setups could be used for temperature-controlled laser heating of spots on tissues. To the best of our knowledge, a temperature-controlled LTB using a single mid-IR fiber has not been reported in the literature. We hypothesized that a temperature-controlled laser bonding system based on a single fiber would solve most of the problems posed by the two-fiber setup, and pave the way to its use in surgical fields that have so far been beyond the scope of LTB.

2 Materials and Methods

2.1 *The Single-Fiber Laser Bonding System—The General Idea*

In order to solve the problems associated with the two-fiber system, we developed a temperature-controlled laser tissue soldering system based on a single mid-IR fiber. The same fiber would transmit the CO_2 laser radiation to a spot on the tissue, and measure the blackbody IR radiation emitted from that spot for purposes of monitoring the temperature. It was expected that the sensitivity to accurate positioning of the distal tip of the fiber would be greatly reduced because the temperature would always be measured exactly on the heated spot; raising the fiber tip and/or tilting the hand-piece would no longer de-focus the system since the system would compensate for the reduced fluence by taking the temperature reading on a larger spot; and pinpointing the measurement of the heated spot would enable assessing the potential tissue thermal damage and permit reducing it where possible.

A CO_2 laser was chosen because it emits mid-IR radiation (10.6 μm) that is completely absorbed after penetrating a few tens of microns into the tissue, ensuring that underlying tissues will not be damaged. The plan for the CO_2 laser radiation was to impinge on the proximal end of a single AgClBr mid-IR transmitting fiber, and for the beam emerging from the distal end of the fiber to heat a spot on the tissue. Meanwhile, the mid-IR radiation emitted from the heated spot is transmitted through the same fiber in the opposite direction. The radiation is focused by means of a simple optical setup (similar to that in Artyushenko's later work[16]) onto the IR detector.

While the idea of using a single mid-IR fiber was quite intuitive, its implementation was challenging. As mentioned above, the original system was based on two fibers: a "power fiber" and a "sensing fiber". The power transmitted through the "power fiber" was fairly strong, being totally isolated from the power transmitted through the "sensing fiber". In the single-fiber system, the laser beam and the mid-IR emission from the tissue are transmitted through the same fiber (and the same optical

elements). The laser radiation is slightly absorbed and heats some of the components along the way: the optical elements, the fiber, the surroundings, etc. The mid-IR radiation emitted from the heated components is also measured by the IR detector, interfering with the accurate measurement of the temperature of the tissue. In addition, the mid-IR radiation emitted from the heated tissue is not coupled directly onto the IR detector, but travels through optical elements, which reduces the signal and accuracy of measurement. All these effects had to be eliminated in order to ensure that the power reaching the detector was strong enough and did indeed correspond to the tissue temperature.

2.2 *System Design*

A core-only silver halide fiber of diameter 0.9 mm (NA = 0.23) was extruded from a single crystal of $AgCl_{0.3}Br_{0.7}$. The fiber had very good transmission, with losses ~0.4 dB/m. The proximal tip of the fiber was held stably inside a narrow silver tube. We used a CO_2 laser of wavelength 10.6 μm (Synrad, Washington, Model J48-1), and focused its beam onto a spot 0.3 mm in diameter, on the proximal end of the fiber, using a ZnSe IR lens of focal length 20 cm. The laser power emitted from the fiber distal tip varied between 0 and 2 W.

The blackbody radiation emitted from the heated spot entered the fiber distal tip without any focusing. The radiation propagated through the fiber and emerged from its proximal tip where it was collected by the optical system and focused onto a pyroelectric IR detector (Hamamatsu, Japan, Model P3782-05). To achieve this we tried several optical setups described in Artyushenko's work,[16] all of which worked well. The IR detector generated a signal "V" that was proportional to the tissue surface temperature T. A schematic drawing of the system design is given in Fig. 1.

The blackbody radiation reaching the pyroelectric detector was chopped at 8 Hz, and amplified using a standard lock-in amplifier. The laser beam was also chopped at the same frequency, but at the opposite phase, so that the two devices operated alternately. This synchronization was crucial to avoid crosstalk between the laser beam and the IR detector. Both the laser and detector were connected to a personal computer, which ran a computer program we wrote that provided a

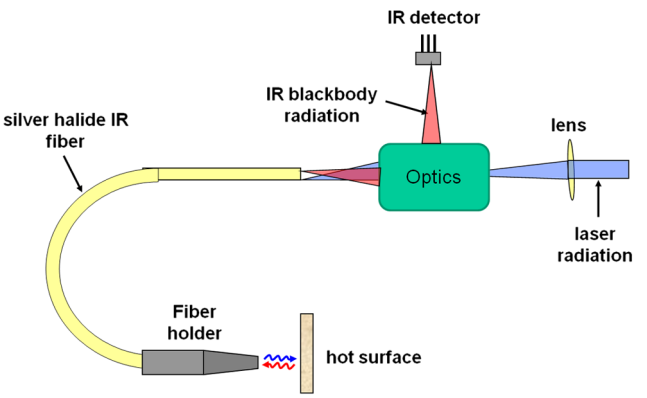

Fig. 1 The single-fiber laser bonding system. A simple optical setup focuses the CO_2 laser beam onto the proximal end of the fiber, and the blackbody infrared (IR) radiation emitted from the heated spot is collected by the same fiber and transmitted back towards the IR detector.

negative feedback loop enabling us to control the surface temperature T and maintain it at any desired average value.

We tested the effect of raising or tilting the fiber tip in an experimental setup comprised of a thin (0.1 mm) slab of corneal tissue phantom (Lucite) whose thermal parameters were: heat capacity $C = 1500\,\mathrm{J\,Kg^{-1}\,°C^{-1}}$, heat conductivity $K = 0.197\,\mathrm{W\,m^{-1}\,°C^{-1}}$, and heat transfer coefficient $h = 8\,\mathrm{W\,m^{-2}\,°C^{-1}}$. It should be noted that all the corneal thermal parameters were of the same order of magnitude. An area on the thin slab was heated in a controlled manner by the laser system until it stabilized at the set value of 60°C, while a thermal camera (FLIR Systems, Oregon, model A40) recorded the temperature profile at the back of the slab. The distal tip of the fiber was gradually elevated from height $H = 2$ to $H = 9$ mm, above and perpendicular to the tissue phantom ($\theta = 0$ deg). For each height increase, several thermal pictures were recorded and averaged. Later, the hand-piece was tilted from $\theta = 0$ deg gradually up to $\theta = 30$ deg (at $H = 5$ mm). Again, thermal pictures were taken and averaged to assess the effect of these changes on the temperature profile at the back of the slab.

2.3 *Bonding of Incisions in the Cornea*

To demonstrate the system's ability to bond incisions in thin delicate organs while causing minimal thermal damage, we performed a set of *ex-vivo* experiments on corneal incisions in 15 bovine eyes. The corneal thickness was 1.0 ± 0.1 mm, prior to the incision, and the epithelial layer was removed using a scalpel. A 3.0 ± 0.1 mm long incision was made at the center of each cornea, perpendicular to the corneal surface, using a 3.0 mm keratome surgical blade. An anterior chamber maintainer was created by inserting a thin plastic tube into the anterior chamber near the limbus and connecting it by a thin hose to an open water tank. The water tank was elevated above the cornea to ensure that water could freely leak through the incision before bonding, and was then lowered to the level of the cornea. The cornea was dried with a surgical Weck-Cell sponge, and the cut edges were approximated. Liquid albumin was prepared by mixing bovine albumin powder with water (44% albumin by weight) and stirring it slowly to a suitable smooth viscous consistency. A thin layer of liquid bovine albumin (0.30 ± 0.05 mm) was then spread over the cut.

The temperature-controlled single-fiber laser bonding system was then activated. The desired temperature was set to 60°C (T_{set}), and the distal tip of the fiber was placed approximately $H = 4$ mm above the corneal surface (at $\theta = 0$ deg). We continuously observed the cornea through a stereo microscope. The spot temperature was initially gradually stabilized at the set value, T_{set}, and kept there for roughly 10 s. The surgeon then moved the tip to the neighboring spot that slightly overlapped the previous spot, and stabilized the temperature there for roughly 10 s, and the process was repeated until the entire incision was bonded (Fig. 2). Immediately after completing the procedure, the water tank was slowly elevated (at roughly 10 cm/s) to produce an increasing hydrostatic pressure, until the bonded incision started leaking. That pressure was considered the burst pressure P_B.

The same experiment performed on 5 additional bovine eyes but without the burst test served for histopathologic evaluation of the bond formed by the laser. After performing the bond, the eyes were fixed in formaldehyde, and a piece of the cornea, including the bonded cut, was processed and embedded in paraffin. Four sections, 4 μm thick, from the middle area of each

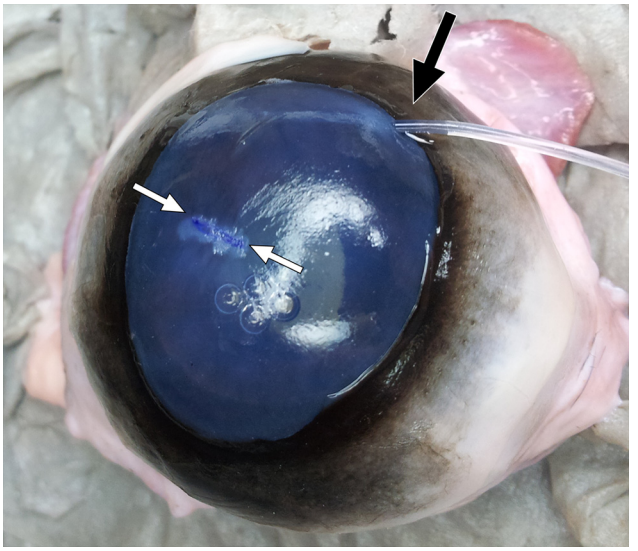

Fig. 2 The cornea after the laser soldering treatment. Prior to the treatment, the cut was marked with a surgical pen. The white material around the marked area is the coagulated albumin on top of the cut. A thin hose was used to create the burst pressure. The edges of the cut are indicated by white arrows. The black arrow shows the anterior chamber maintainer.

bonded cut, were stained by hematoxylin and eosin and examined by light microscopy.

3 Results

The system performed well following the alignment. It was first tested by heating a spot on the tissue phantom (e.g., Lucite). The system was able to maintain a set temperature of a spot on the Lucite slab with fluctuations of 0.5°C. The thermal images of the thin slab recorded by the thermal camera showed very good results. When the height H was increased in steps of 1 mm, from $H = 2$ to $H = 9$ mm, for each increase of 1 mm, the central temperature increased by only ~1°C. The temperature profile was slightly broadened (in proportion to H), but this was a less important parameter compared with the maximal temperature.

When the fiber tip was tilted from $\theta = 0$ deg to $\theta = 30$ deg, surprisingly, the temperature at the center of the profile remained constant. The shape of the temperature profile lost its circular symmetry, as expected; but, more importantly, the maximal temperature did not change.

When the system was operated on biological materials, it again easily stabilized the temperature at the set value T_{set}. However, there was a slight increase in the amplitude of the fluctuations to ~2°C. The negative feedback loop, with the correct parameters, quickly (~5 s) and accurately (no overshoot) achieved the stabilization (Fig. 3).

The entire procedure of laser soldering the corneal incisions with the single-fiber system in each of the 15 eyes was very simple and fast. The temperature stabilization was also fast and accurate, conditional upon the surgeon slowly moving from spot to spot along the incision, with a slight overlap between neighboring spots. The single fiber system was very flexible and the target area was easily observed, as opposed to the older two-fiber systems, in which the field of view was partially blocked.

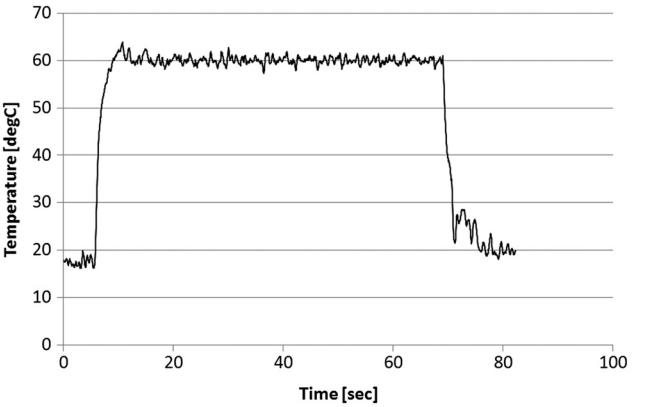

Fig. 3 Temperature monitoring and control of a heated spot on a bovine cornea. The set value was reached within ~5 s. There is hardly any overshoot.

The average burst pressure was $P_B = 100 \pm 30$ mm Hg ($N = 15$), which is more than three-fold higher than the normal intra-ocular pressure. When the albumin layer was peeled off, a macroscopic examination of the incision area showed no signs of thermal damage other than a slight retraction of the cut edges close to the surface. Clinical evaluation of the eyes did not turn up any significant distortion of the corneas, and their appearance was less distorted than in the case of a sutured corneal incision.

Histopathological analysis of 5 eyes revealed the corneal cuts to be well bonded (Fig. 4). The superficial layers exhibited coagulated albumin and coagulated collagen, and both stained dark pink and dark purple by the Hematoxilin and Eosin stain. The coagulated collagen area was approximately 1 to 1.5 mm in width and 0.1 to 0.15 mm in depth. The deeper layers in the cornea stroma appeared to be undamaged and the wound edges were well apposed. The posterior layers of the cornea, the Descemet's membrane and endothelium, were well apposed, with mild gapping of the inner wound.

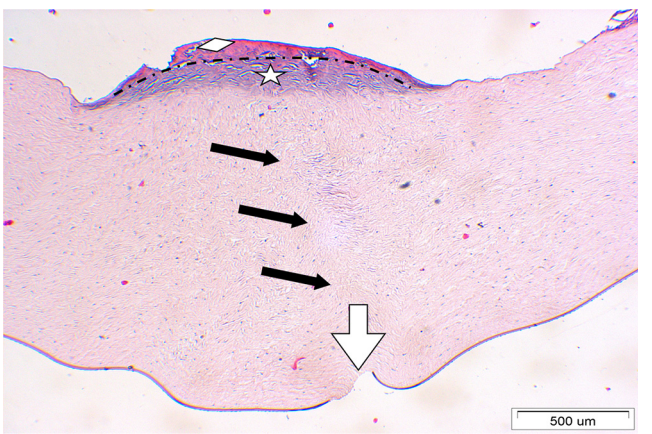

Fig. 4 Histology of the cross-section of the bonded area of the cornea stained with Hematoxylin and Eosin. The coagulated albumin is indicated by a white diamond. The coagulated superficial corneal area is indicated by a white star (the dashed line is the approximated borderline between them). The white arrow indicates the gap in Descemet's membrane. The black arrows point to the successful bonding of the stroma.

4 Discussion

Although laser bonding of incisions in tissues is promising, it has failed to make an impact on clinical practice. We attribute this to the fact that temperature control is not used in most of the current research programs, and bonding strength and thermal damage remain major problems. The single mid-IR fiber system we designed and describe here proved capable of bonding corneal incisions in bovine corneas, *ex vivo*, with greatly reduced thermal damage and more precise control of temperature both on and below the surface of the tissue. The burst pressure measurements indicated that the bonding was sufficiently strong. The system demonstrated the basic capabilities of heating, sensing, and controlling spot temperature, and exhibited several advantages over the two-fiber system. It is more suitable for endoscopic procedures; easier to manipulate; is less sensitive to the height and tilt of the fiber tip; has a symmetry that allows more precise, quantitative estimation of the issue temperature profile, resulting in less thermal damage; and permits a larger field of view of the treated area thanks to the smaller diameter of the hand-piece. Light microscopic examination of the bonded wound revealed no retraction of the corneal surface. Some superficial thermal damage was observed (100 to 150 μm), which could potentially be cleared and healed under *in-vivo* conditions. Simulations and experiments are underway aimed at modifying the single-fiber system to further reduce the sensitivity to height and tilt, and to eliminate the potential for thermal damage.

5 Summary

A laser-bonding system based on a single AgBrCl mid-IR transmitting fiber was successfully designed and built. When tested for the controlled heating of spots on inorganic samples, the system exhibited excellent results. It was then tested for laser soldering of incisions in bovine corneas, *in vitro*. The burst pressure results obtained were three-fold (or more) higher than the normal intraocular pressure. The heating appeared to have caused little, and only superficial, thermal damage. We plan to use the system in the near future for the bonding of incisions in corneas of rabbits *in-vivo*. The new system was based on a single mid-IR fiber with a diameter of 0.9 mm, which can be reduced to make it suitable for laser bonding of small blood vessels in microsurgery, for endoscopic surgery, and even for robotic surgery.

Acknowledgments

The authors thank the Esther and Chaim Koppel Trust and the Chief Scientist Office, Ministry of Health, Israel for partial support of this project. We also thank the Elizabeth and Nicholas Slezak Super Center for CardiacResearch and Medical Engineering for their support.

References

1. L. S. Bass and M. R. Treat, "Laser tissue welding: a comprehensive review of current and future clinical applications," *Laser Surg. Med.* **17**(4), 315–349 (1995).
2. K. K. Jain and W. Gorisch, "Repair of small blood vessels with the neodymium-YAG laser: a preliminary report," *Surgery* **85**(6), 684–688 (1979).
3. F. M. P. Leclère et al., "Outcomes after 1.9-microm diode laser-assisted anastomosis in reconstructive microsurgery: results in 27 patients," *Plast. Reconst. Surg.* **125**(4), 1167–1175 (2010).
4. P. Garcia et al., "Robotic laser tissue welding of sclera using chitosan films," *Laser Surg. Med.* **41**(1), 60–67 (2009).
5. T. Menovsky, J. F. Beek, and M. J. C. van Gemert, "CO_2 laser nerve welding: optimal laser parameters and the use of solders in vitro," *Microsurgery* **15**(1), 44–51 (1994).
6. J. M. Massicotte, R. B. Stewart, and D. P. Poppas, "Effects of endogenous absorption in human albumin solder for acute laser wound closure," *Laser Surg. Med.* **23**(1), 18–24 (1998).
7. A. Lauto et al., "In vitro and in vivo tissue repair with laser-activated chitosan adhesive," *Laser Surg. Med.* **39**(1), 19–27 (2007).
8. O. Shenfeld et al., "Temperature monitoring and control of CO_2 laser tissue welding in the urinary tract using a silver halide fiber optic," *Proc. SPIE* **1876**, 203–214 (1993).
9. A. Barak et al., "Temperature-controlled CO_2 laser tissue welding of ocular tissues," *Surv. Ophthalmol.* **42**(Suppl 1), 77–81 (1997).
10. A. Zur and A. Katzir, "The use of infrared fibers for low temperature measurement," *Appl. Phys. Lett.* **48**(7), 499–500 (1986).
11. B. Forer et al., "Repair of pig dura in vivo using temperature-controlled CO_2 laser soldering," *Laser Surg. Med.* **37**(4), 286–292 (2005).
12. D. Simhon et al., "End-to-end small bowel anastomosis by temperature-controlled CO_2 laser soldering and an albumin stent—a feasibility study," *Proc. SPIE* **5312**, 176–185 (2004).
13. E. Strassman et al., "Laser soldering of the cornea in a rabbit model using a controlled-temperature CO_2 laser system," *Proc. SPIE* **4244**, 253–265 (2001).
14. G. Noguera et al., "Novel laser-activated solder for sealing corneal wounds," *Invest. Ophthalmol. Visual Sci.* **48**(3), 1038–1042 (2007).
15. V. G. Artjushenko et al., "Fiber optic device for simultaneous laser power transmission and temperature measuring of irradiated object," *Electron. Lett.* **20**(23), 983–984 (1984).
16. V. Artyushenko, "Fiber optic probe," U.S. patent No. 7956317 (2011).