Scientific
Research

Evaluation of the Bending Loss of the Hollow Optical Fiber for Application of the Carbon Dioxide Laser to Endoscopic Therapy

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ABSTRACT

Since carbon dioxide laser is excellent for incision, hemostasis, coagulation, and vaporization of soft tissues, it has been widely applied in clinical treatments as the laser knife. In these days, flexible thin hollow optical fibers transmitting mid-infrared light have been developed, and the application of carbon dioxide laser to endoscopic therapy has become possible. However, it is expected that the irradiation effect is influenced by the change in the laser power at the tip of the hollow optical fiber due to the change in the transmittance by the bending loss. The purpose of this research is to quantitatively evaluate the change in the output power and therapeutic effect by bending the hollow optical fiber in a gastrointestinal endoscope. The change in the transmittance of the hollow optical fiber due to the insertion of the fiber into the endoscope and bending of the head of the endoscope was measured. Then, the relationship between the irradiated laser power and the incision depth for a porcine stomach was investigated. As the results, the most significant decrease in the transmittance of the hollow optical fiber was caused by the insertion of the fiber into the instrument channel of the endoscope, and bending of the head of the endoscope with the angle of 90° decreased the output laser power and incision depth by 10% and 25%, respectively. Therefore, it was confirmed that the bending loss of the hollow optical fiber due to the bending of the head of the endoscope had no significant influence on the endoscopic therapy using the carbon dioxide laser.

Keywords: Hollow Optical Fiber; Carbon Dioxide Laser; Bending Loss; Endoscopic Therapy

1. Introduction

Since carbon dioxide laser with the wavelength of 10.6 μm is absorbed strongly by water, which is contained about 70% in biological soft tissues, it is excellent in incision, hemostasis, coagulation, and vaporization of the soft tissues [1,2]. In fact, carbon dioxide laser is widely efficient for medical field, e.g., plastic surgery, cosmetic surgery, otolaryngology, and dentistry. Even in the endoscopic surgery, carbon dioxide laser had various promising application, so that many researchers developed new techniques for clinical treatment [3,4]. At the same time, several kinds of optical fibers transmitting mid-infrared light have been developed. These optical fibers are classified broadly into three categories, *i.e.*, glasses composed of fluoride or oxide, special glasses such as chalcogenide, and polycrystalline materials such as metal halide [5]. However, these optical fibers have problems

in terms of the mechanical and chemical properties, toxicity, manufacturing cost, etc. Therefore, they could not be used for medical purposes. Recently, hollow optical fibers with a dielectric coating made of cyclic olefin polymer (COP) inside a cylindrical metal waveguide have been developed as shown in **Figure 1** [6]. Since the core of the hollow optical fibers is air or other gases with a high transmittance at the wavelength of the carbon dioxide laser, extremely low absorption loss could be obtained [7]. Because the reflection loss at the end surface of the fiber and related damage do not occur, the hollow optical fibers are suitable to transmit high-power lasers. In addition, the hollow optical fiber shown in **Figure 1** can simultaneously transmit carbon dioxide laser and visible aiming laser. Although it is possible to enhance the reflectance of the inner surface to nearly 100% by designing the COP layer thickness according to the wave-

length of the transmitting laser light, the leaking loss to the outside of the fiber can not be eliminated in principle. Therefore, by bending the hollow optical fiber, the transmittance decreases due to the increase in the number of reflection per unit length and decrease in the reflectance in the fiber. When the hollow optical fiber is used in a clinical treatment with an endoscope, it is inferred that output laser power and therapeutic effect are changed due to the change in the transmittance of the hollow optical fiber caused by the bending of the endoscope. The purpose of this research is to quantitatively evaluate the change in the output laser power and therapeutic effect caused by bending the hollow optical fiber in a gastrointestinal endoscope.

2. Materials and Methods

2.1. Transmittance of the Hollow Optical Fiber with Bending of the Endoscope

In this research, a carbon dioxide laser system modified from a commercial laser system for dental treatment and so on (COM-2, J. Morita Manufacturing Corp., Kyoto, Japan) was used. The laser was operated in the continuous wave mode, and the maximum output power of the laser oscillator was 30 W. Two types of the hollow optical fiber (J. Morita Manufacturing Corp.) listed in **Table 1** were used to deliver the carbon dioxide laser through a gastrointestinal endoscope (GIF-2T200, Olympus Corp., Tokyo, Japan) shown in **Figure 2**. A laser power meter (30-A-BB-18, Ophir Optronics, Israel) was used to measure the laser power.

The carbon dioxide laser, the visible aiming laser with a wavelength of 650 nm, and air were simultaneously transmitted through the hollow optical fiber. Input and output powers of the hollow optical fibers were measured under four different conditions as shown in **Figure 3**, and each condition was described as follows:
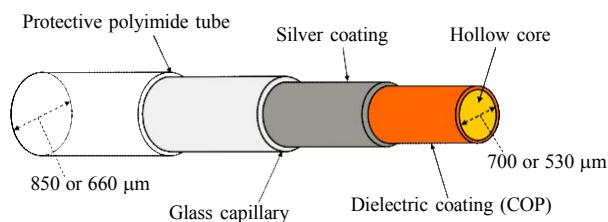

Figure 1. Schematic of the hollow optical fiber used in this research.

Table 1. Specifications of the hollow optical fibers.

	Outer diameter (μm)	Inner diameter (μm)	Length l (cm)
L-1	850	700	250
L-2	850	700	260
S-1	660	530	260

I. The hollow optical fiber was kept straight without insertion to the endoscope.

II. The hollow optical fiber was inserted into the endoscope, and endoscope was kept straight.

III. In addition to the condition II, the middle part of the endoscope was gradually bent 90° with a radius $r_1 = 50$ cm.

IV. In addition to the condition III, the head of the endoscope was steeply bent 90° with a radius $r_2 = 5$ cm.

2.2. Dependence of the Output Laser Power and Incision Depth on the Bending of the Head of the Endoscope

By assuming the case where the endoscope is used in transoral surgery, the middle part of the endoscope was bent 90° with a radius $r_1 = 50$ cm, and the tip of the hollow optical fiber L-1 was bent with various angles and radii r_2 as listed in **Table 2**. To avoid the accidental damage to the endoscope due to the break of the hollow optical fiber during laser irradiation, the fiber was ejected about 15 cm from the head of the endoscope, and only the fiber was bent by using in-house guide plates made of foamed polystyrene in order to simulate the bending conditions of the head of the endoscope.

A segment of a porcine stomach was set on a motorized linear stage (SGSP20-20, SIGMA KOKI Co., Ltd., Tokyo, Japan) and moved at a constant speed of 1.0 mm/s during laser irradiation for a time of 20 s. The car-
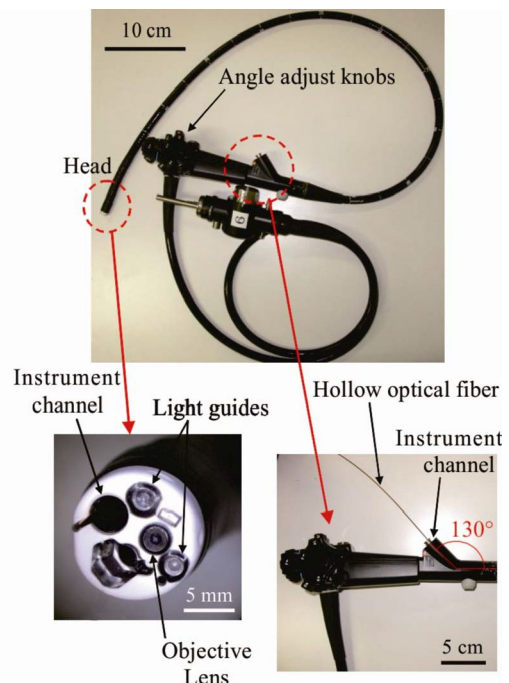

Figure 2. The photographs of the endoscope and its head and instrument channel. The hollow optical fiber is steeply bent 130° just after the insertion to the instrument channel.
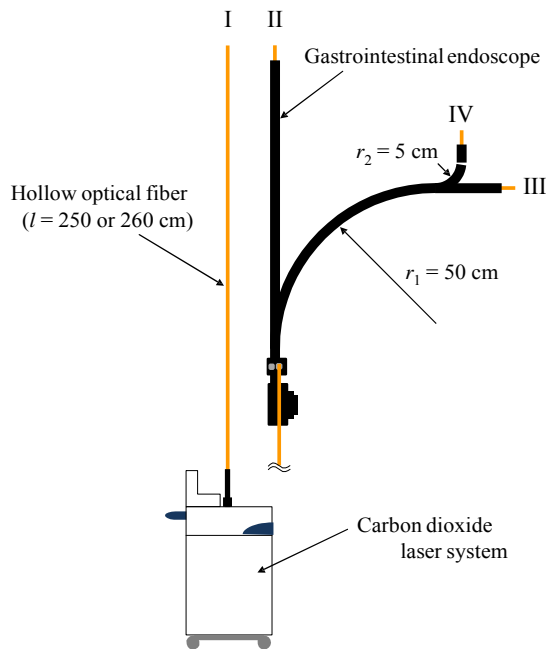
Figure 3. Schematics of the measurement of the transmittance of the hollow optical fibers for various bending conditions.

Table 2. Conditions for the measurement of the dependence of the output laser power and incision depth on the bending of the head of the endoscope.

Bending angle (°)	0	30	60	90
Bending radius r_2 (cm)	-	11.0	7.0	5.0
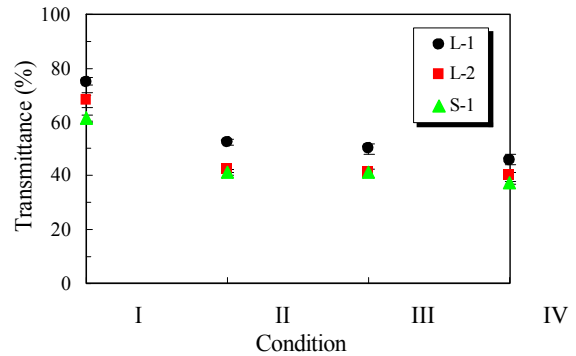
Figure 4. Transmittance of the hollow optical fibers under each condition.

bon dioxide laser was irradiated vertically to the surface of the mucosa from a distance of 2 mm. The setting power of the carbon dioxide laser system was set at 3, 5, and 8 W, and laser power transmitted through the hollow optical fiber was measured under each condition. To simulate the *in vivo* environment, the sample surface was kept wet by pouring saline on the sample at a rate of 120 mL/h using a micro syringe pump (IC3100, AS ONE, Osaka, Japan). After laser irradiation, each sample was stored at −80°C and was sliced to a thickness of 10 μm by using a cryostat microtome (CM1850, Leica Microsystems, Wetzlar, Germany) at a temperature of −20°C. Then, each sample section attached onto a glass slide was stained with hematoxylin and eosin (HE) staining and observed using a high-resolution slide scanner (NanoZoomer 2.0 RS, Hamamatsu Photonics K. K., Shizuoka, Japan).

3. Results and Discussion

3.1. Transmittance of the Hollow Optical Fiber with Bending of the Endoscope

Figure 4 shows the transmittance of the hollow optical fibers under each condition. When the hollow optical fiber was inserted into the instrument channel of the endoscope (from the condition I to II), the transmittance of the fiber L-1, L-2, and S-1 decreased by 29%, 28%, and 24%, respectively. On the other hand, the transmittance was not significantly decreased by the bending of the endo-

scope. When the bending of the endoscope was changed from the condition II to IV, the transmittances of the fibers L-1, L-2, and S-1 decreased by 7, 3, and 4%, respectively. Therefore, it was found that the decrease in the transmittance of the hollow optical fiber due to the insertion into the instrument channel was the most dominant loss in the hollow optical fibers. It is suggested that these results are caused by the steep bending of 130° just after the insertion into the instrument channel as shown in **Figure 2**.

3.2. Dependence of the Output Laser Power and Incision Depth on the Bending of the Head of the Endoscope

Figure 5 shows the relationships between the bending angle of the tip of the hollow optical fiber and the output laser power. The output laser power gradually decreased with the increase in the bending angle of the tip of the hollow optical fiber, and the output laser power at 90° was about 10% lower than that for 0°.

Figure 6 shows the photomicrographs of the cross sections of the HE stained porcine stomach samples after laser irradiation with each condition. **Figure 7** shows the relationships between the bending angle of the tip of the hollow optical fiber and the incision depth measured from the photomicrographs of the sample shown in **Figure 6**. The incision depth under each condition decreased by up to 25% compared to that for 0° with an exception for the condition of 8 W at 90°. Therefore, it was confirmed that the bending loss of the hollow optical fiber due to the bending of the head of the endoscope had no significant influence on the endoscopic therapy using the carbon dioxide laser.
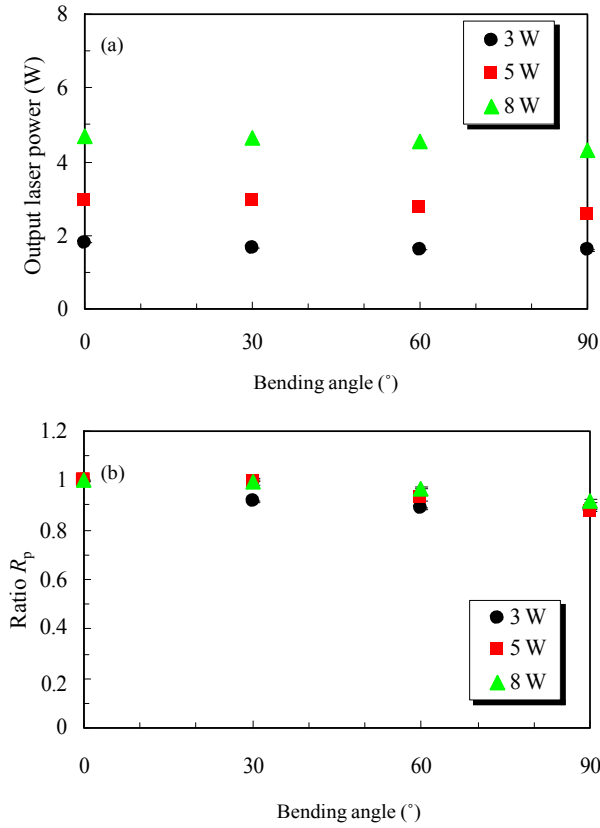

Figure 5. The relationships between (a) the bending angle of the head of the endoscope and the output laser power and (b) the bending angle and the ratio R_p of the output laser power to that for 0°.

Figure 6. Photomicrographs of the HE stained samples after laser irradiation with each condition.

The mechanism of the vaporization of the biological soft tissue by a laser knife is described as follows: When the biological soft tissue absorbs the laser, the temperature of the tissue quickly rises within a time of microseconds. Since the boiling point of water is 100°C, water in tissue is immadiately vaporized and the volume is expanded by continuously making water in the tissue over 100°C. Then, cell membrane and organelle fragment are vaporized. In our experiment, laser was linerly scanned with a speed of 1.0 mm/s, incision was performed using the vaporization. In order to describe the reltionship between the irradiated laser energy density and the incision depth, the steady-state model has been proposed [8,9]. This model was developed for continuous ablation process using a continuous wave or pulsed laser with microseconds or longer pulse width. In the steady-state model, the relational expression between the incision depth δ (mm) and the irradiated laser energy density Φ_0 (J/mm²) is given by

$$\delta = \frac{\Phi_0 - \Phi_{th}}{\rho h_{abl}}, \qquad (1)$$

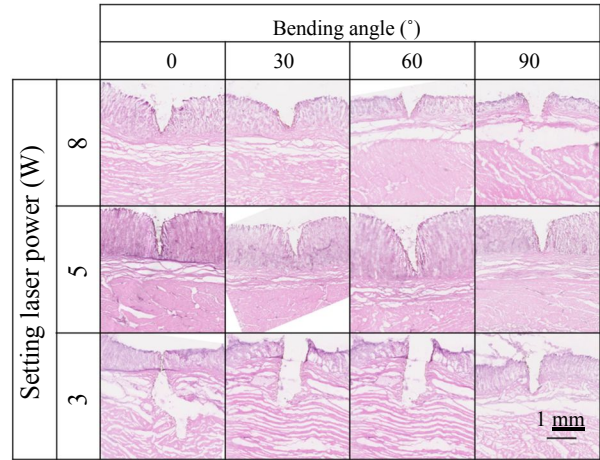where $\rho = 1.02 \times 10^{-3}$ g/mm³ is the density of biological

soft tissue composed of water of 70%, and the ablation enthalpy h_{abl} (J/g) and the ablation threshold energy density Φ_{th} (J/mm²) are defined as follows:

$$h_{abl} = C\Delta T + L,$$

$$\Phi_{th} = \frac{\rho h_{abl}}{\mu_a},$$

where $C = 3.45$ J/(g·K) is the heat capacity, $\Delta T = 63.0$ K is the temperature rise to the boiling point of water, $L = 2.58 \times 10$ J/g is the heat of vaporization, and $\mu_a = 6.15 \times 10^2$ mm⁻¹ is the absorption coefficient of the porcine mucosa at the wavelength of 10.6 μm [10-12]. In **Figure 8**, the dots show the relationship between the irradiated laser energy density and the incision depth obtained from the results in **Figures 5** and **7**, and the solid line represents the theoretical values estimated using Equation (1). The laser beam diameter at the surface of the porcine stomach mucosa was measured for the fiber L-1 by using knife-edge method and was about 710, 712, 772, and 740 μm at the bending angle of 0°, 30°, 60°, and 90°, respectively. The laser beam diameter was increased with the increase in the bending angle, because the beam spread angle from the exit of the hollow optical fiber was increased with the increase in the bending angle due to the conversion of the propagation mode from the low-order modes to the high-order modes in the hollow optical fiber.

From the results in **Figure 8**, it was found that both the experimental and theoretical values have similar correlations between the irradiated laser energy density and the incision depth. However, the theoretical values were about 1.5 times higher than the experimental values. It is inferred that the principal factor for this difference is the uncertainty in the physical values used in the model calculations, e.g., the density of biological soft tissue, ablation enthalpy, and the heat capacity. Since these values
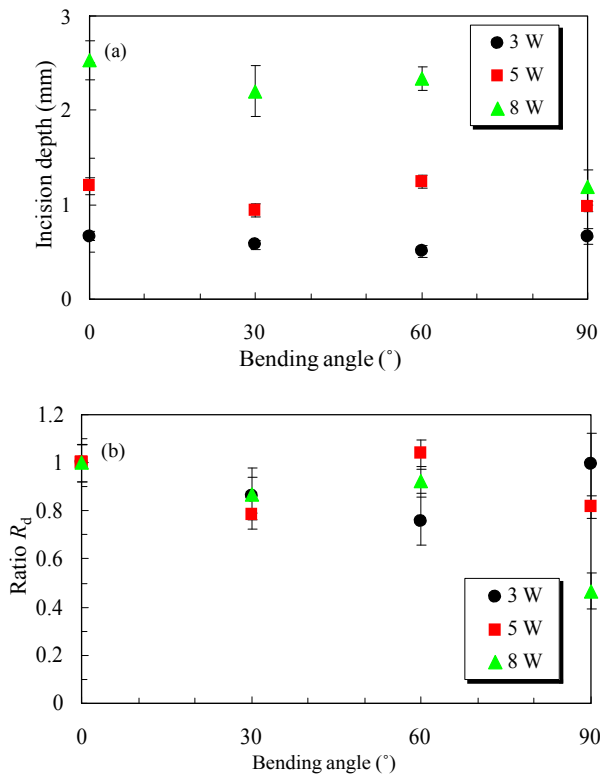
Figure 7. The relationships between (a) the bending angle of the tip of the hollow optical fiber and incision depth and (b) the bending angle and the ratio R_d of the incision depth to that for 0°.

were taken from the literatures, some values might not be suitable for the porcine stomach sample used in this research.

4. Conclusion

The change in the output power of the carbon dioxide laser and therapeutic effect by bending the hollow optical fiber inserted in a gastrointestinal endoscope was quantitatively evaluated. The change in the transmittance of the hollow optical fiber due to the insertion of the fiber to the endoscope and bending of the head of the endoscope was measured. Then, the relationship between the irradiated laser power and the incision depth of a porcine stomach was investigated. As the result, the greatest decrease of the transmittance of the hollow optical fiber was caused by insertion of the fiber into the instrument channel of the endoscope, and bending of the head of the endoscope with the angle of 90° decreased the output laser power and incision depth by no more than 10% and less than 25%, respectively. Therefore, it was confirmed that the bending loss of the hollow optical fiber due to the bending of the head of the endoscope had no significant influence on the endoscopic therapy using the carbon dioxide laser.
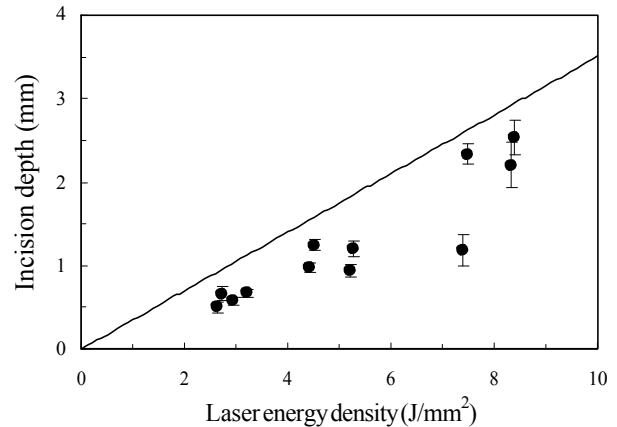

Figure 8. The relationships between the incision depth and irradiated laser energy density, where the solid line shows the theoretical values estimated with the steady-state model.

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