Selective delivery of laser energy to biological tissue based on refractive-index differences

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We report a novel method for selective laser energy delivery into biological tissues based on refractive-index differences. As a specific example, Ho:YAG laser energy is delivered into fat preferentially over other soft tissue, despite the fact that water has a higher absorption coefficient at this wavelength than fat.

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Delivering laser energy selectively into a specific type of tissue and not into other tissues is highly desirable in various medical procedures. This need is particularly acute when one is removing unwanted fatty tissues from specific areas of the body. For example, during liposuction, the most commonly performed cosmetic procedure in plastic surgery, and during laproscopic surgery, a procedure that is becoming increasingly popular, there are times when fatty tissue has to be removed or dissected without affecting adjacent tissues and delicate structures.[1,2] A scheme for selective laser energy delivery into tissue has been suggested in which a feedback system is used that spectroscopically identifies the target issue and determines whether to deliver the laser energy.[3] However, this system is complex and has yet to be demonstrated clinically.

We report here a new method for selective laser energy delivery into tissues, based on refractive-index differences between the target tissue and other tissues. Fats and oils have a substantially higher index in the optical spectrum than other soft tissues and were chosen as an example.[4,5] Optical energy is delivered into tissue through a contacting optical medium at an incident angle selected to be greater than the critical angle for low-index tissues but below the critical angle for high-index tissues, thus allowing optical energy to be coupled efficiently into tissues having a higher refractive index but not into other tissues having a lower refractive index. The selectivity is made possible simply by contact of the optical medium with tissue and does not require any additional complex instrumentation. This method can be used with other biological and nonbiological substances as well.

To demonstrate the feasibility of this method we measured reflectance at a sapphire prism–sample interface. This measurement is a good indication of energy delivery, because the delivered total energy (T) is given by $T = 1 - R$, where R is the reflected energy from the interface. The higher the reflectance R from an optical interface, the lower the energy delivered (T) through this interface and vice versa.

The experimental setup for measuring the reflectance is shown in Fig. 1. A flash-lamp-excited pulsed Ho:YAG laser (Schwartz Electro-Optics, Orlando, Florida) operating at a wavelength of 2.09 μm, with a pulse duration of 250 μs (FWHM) and a repetition rate of 2 Hz, was used. The laser output was s polarized with a stack of 30 glass microscope slides oriented at Brewster's angle. We placed a He–Ne laser aiming beam collinearly with the Ho:YAG laser. The laser beam was then focused by a 25-cm focal-length lens onto a sapphire hemicylindrical prism placed on a rotational stage with a 1/2° scale interval. We eliminated the effects of dynamic changes in light reflection at the interface that were due to cavitation and vaporization of the tissue sample by adjustment of the energy density at the sample's surface to less than the threshold for tissue ablation. The sample was placed in direct contact with the prism, and the reflectance from the prism–sample was measured for various incident angles by adjustment of the rotation stage. We measured the reflected energy from the prism–sample interface and the laser output energy simultaneously to compensate for pulse-to-pulse fluctuations of the laser output.

We calculated the reflectance at the sample–sapphire interface by dividing the value of the energy

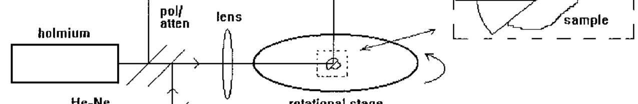

Fig. 1. Experimental setup: pol/atten, polarizer–attenuator.

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reflected from the prism–sample interface at a given incident angle by the value obtained for total internal reflection at the prism–air interface. Figure 2 shows the reflectance measured at sapphire–corn oil, sapphire–water, and sapphire–air interfaces for various angles of incidence. For angles of incidence from 48° to 53°, the reflectance of the water interface is substantially larger than that of the corn oil interface. Figure 3 shows the reflectance measured at sapphire–beef fat and sapphire–muscle interfaces. The reflectance from a sapphire–muscle interface is also substantially larger than the reflectance from a sapphire–beef fat interface for angles of incidence near 50°. These results suggest that, at a particular range of incident angles, optical energy can be selectively delivered to higher-index substances (e.g., fatty tissues) and not to lower-index substances (e.g., water-based soft tissues).

The experimental results are in excellent agreement with theory, as shown in Fig. 2. The theoretical values for s polarization reflectance were obtained with the Fresnel equation

$$R_s = \left| \frac{n_1 \cos\theta - [(n_2{}^*)^2 - n_1{}^2 \sin^2\theta]^{1/2}}{n_1 \cos\theta + [(n_2{}^*)^2 - n_1{}^2 \sin^2\theta]^{1/2}} \right|^2, \qquad (1)$$

where θ is the incident angle, n_1 is the refractive index of the transparent optical material, and $n_2{}^* = n_2(1 - i\kappa_2)$ is the complex refractive index of the sample. The values of the refractive indices and the extinction coefficients were obtained from Table 1. Theory also predicts that selectivity can be easily obtained for wavelengths that are weakly absorbed by samples (e.g., $n_2\kappa_2 \leq 0.01$). This selectivity is achieved when $n_{2\mathrm{H}} > n_1 \sin\theta > n_{2\mathrm{L}}$, where $n_{2\mathrm{H}}$ and $n_{2\mathrm{L}}$ are samples with higher and lower refractive indices, respectively, is approximately satisfied. However, for wavelengths that are strongly absorbed by samples, selectivity could not always be achieved because of the strong effect of absorption on reflectance. Figure 4 shows the calculated reflectances of water and corn oil at the strongly absorbed wavelength of the Er:YSGG laser, 2.79 μm. Selectivity cannot be easily achieved, even though the refractive index of water is substantially lower than that of corn oil.

We used a hemicylindrical sapphire prism for the experiments because it provided the simplest variable-angle internal reflection element. Practical devices are not limited to this configuration and optical material only. For example, devices could be made from a microprism mounted on the distal end of an optical delivery system at the appropriate angle.

Delivering laser energy into a substance through a well-defined refractive optical interface instead of through air has more practical advantages. Figure 2 shows that the incident angle required for selective energy delivery is higher than the critical angle for total internal reflection from the prism–air interface. Thus no optical energy is emitted from the optical interface when it is not in contact with the desired substance. This is an important safety feature and an advantage over conventional freely propagating laser beams, which can be dangerous to the environment and to adjacent tissues.[8] This method also allows tactile feedback during the surgical procedure. This is another potential advantage for laser surgery that adds more control for the surgeon. Interface-delivery laser surgical tools may be especially compatible with minimally invasive surgery, in which the combination of tissue selectivity and added safety is desirable.

It is well known that selective laser targeting of specific tissues can be achieved by use of wavelengths that are preferentially absorbed by the target tissue.[9] Here we describe a new way to achieve tissue-selective coupling of light based on refractive-index differences. Interestingly, these two methods are largely independent and in some settings can be combined. For example, at 2.09 μm, as shown in Fig. 3, selective coupling into fat can be achieved even though absorption of this wavelength is higher in other tissues. From 480 to 500 nm, human fat absorbs more strongly than many other tissues because of carotenoids[9]; the refractive index of fat is also high at this wavelength. Potentially, therefore, one could design a highly fat-selective laser and delivery system, using the combination of the higher absorption and higher refractive index of fat at these blue visible wavelengths.

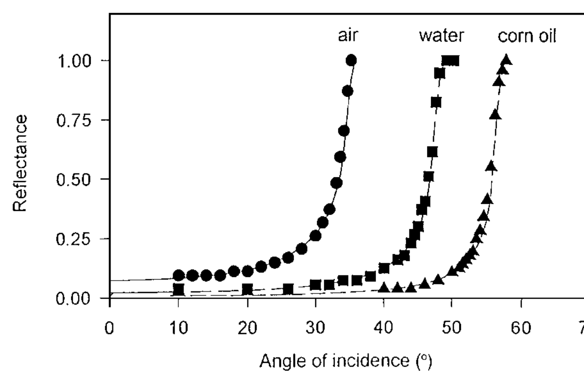

Fig. 2. Reflectance of 2.09-μm laser radiation at (triangles) the sapphire–corn oil, (squares) sapphire–water, and (circles) sapphire–air interfaces. The curves show the theoretical predictions obtained with Eq. (1) and the values in Table 1.

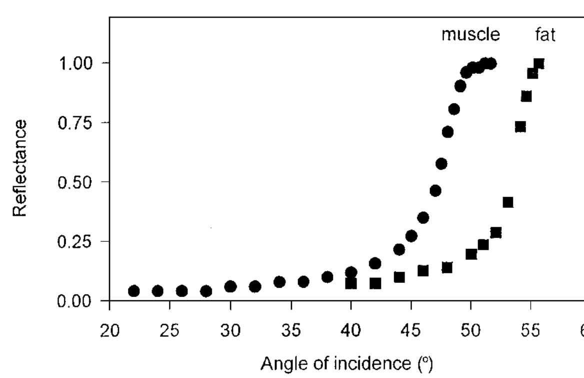

Fig. 3. Reflectance of 2.09-μm laser radiation at the (squares) sapphire–muscle fat and (circles) sapphire–water interfaces.

Table 1. Optical Properties at Two Wavelengths (in micrometers)

	Refractive Index		Extinction Coefficient	
Substance	2.09 μm	2.79 μm	2.09 μm	2.79 μm
Corn oil	1.45[a]	1.45	0.000043[b]	0.00047[b]
Water	1.292[c]	1.091[c]	0.00049[c]	0.14[c]
Sapphire	1.736[d]	1.718[d]		

[a]Refractive index of this substance was experimentally determined from reflectance measurements at 2.09 μm.
[b]Value was determined from transmission through a 1-mm fused-silica cuvette at 2.09 and 2.79 μm.
[c]Ref. 6.
[d]Ref. 7.

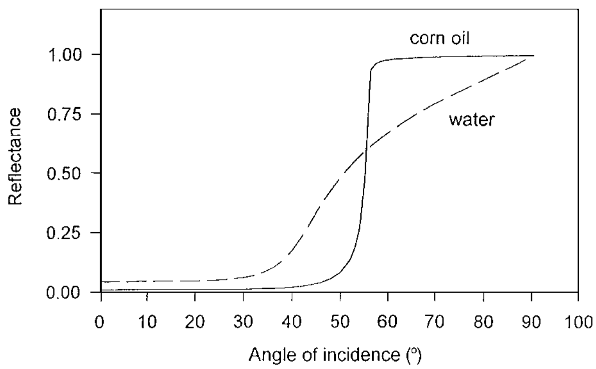

Fig. 4. Calculated reflectance of 2.79-μm laser radiation at the sapphire–corn oil and sapphire–water interfaces.

In conclusion, in this Letter we have shown the feasibility of selective delivery of laser energy into specific higher-index tissues and not into others. Once delivered, optical energy can be used surgically for heating, melting, or vaporizing tissues or potentially to cause specific photochemical reactions, as in the case of photodynamic therapy. Fatty tissue and oil are shown as examples. As we implement this new approach, we hope to devise safer, practical, tissue-selective laser surgical devices.

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References

1. F. M. Grazer and R. H. de Jong, Plast. Reconstr. Surg. **105,** 436 (2000).
2. D. Rattner, Department of Surgery, Massachusetts General Hospital, Boston, Mass. 02114 (personal communication, 1997).
3. R. J. Verbunt, M. A. Fitzmaurice, J. R. Kramer, N. B. Ratliff, C. Kittrell, P. Taroni, R. M. Cothren, J. Baraga, and M. Feld, J. Am. Heart Assoc. **123,** 208 (1992).
4. F. P. Bolin, L. E. Preuss, R. C. Taylor, and R. J. Ference, Appl. Opt. **28,** 2297 (1989).
5. H. Li and S. Xie, Appl. Opt. **35,** 1793 (1996).
6. D. M. Wieliczka, S. Weng, and M. R. Querry, Appl. Opt. **28,** 1714 (1989).
7. *Ealing Company Optical Catalog* (Ealing, Hollistown, Mass., 1995).
8. A. G. Macdonald, Br. J. Anaesth. **73,** 847 (1994).
9. M. R. Prince, G. M. LaMuraglia, P. Teng, T. F. Deutsch, and R. R. Anderson, IEEE J. Quantum Electron. **QE-23,** 1783 (1987).