Thermal imaging of the temporal bone in CO_2 laser surgery: An experimental model

BRIAN J. F. WONG, MD, JON LEE, BS, GEORGE T. HASHISAKI, MD, MICHAEL W. BERNS, PhD, and **JOSEPH NEEV, PhD,** Irvine, California, and Charlottesville, Virginia

The unique properties of lasers create an enormous potential for specific treatment of chronic ear disease. Despite the widespread acceptance and use of the laser, however, a complete understanding of the time- and space-dependent temperature distribution in otic capsule bone immediately after pulsed laser exposure has not been elucidated. Using a liquid nitrogen-cooled mercury-cadmium telluride infrared detector, the temperature distribution in human cadaveric otic capsule bone was determined immediately after pulsed (100 msec) carbon dioxide laser exposure (0.3 to 4.0 W; 200 μm spot diameter). The time- and space-dependent temperature increases and thermal diffusion were determined as a function of the laser power density and were found to vary linearly. (Otolaryngol Head Neck Surg 1997;117:610-5.)

Carbon dioxide (CO_2 [10.6 μm]), argon (488-514 nm), and potassium-titanyl-phosphate (KTP) neodymium: yttrium-aluminum-garnet (Nd:YAG) (532 nm) lasers have been advocated for use in the treatment of chronic ear disease by several authors and have been used clinically to ablate diseased mucosa and remnant cholesteatoma in the mastoid cavity and middle ear space.[1-3] In contrast to mechanical methods of tissue removal or destruction, the laser allows for a "hands-free" treatment of diseased tissue in inaccessible or delicate areas; gross mechanical trauma and vibration are minimized. Difficult to reach areas such as the facial recess or sinus tympani can be accessed with fiber optic delivery systems or a mirror that provides surfaces for laser beam deflection. Typically, laser treatments of diseased tissues have been performed using defocused beams. This defocusing of the beam corresponds to a decrease in the energy density incident on the treated surface.

Although lasers have many theoretical advantages in the treatment of chronic ear disease, including possible percutaneous, transtympanic, and photodynamic treatment, the adverse effects of lasers on the temporal bone have not been fully investigated. The commonly used lasers in middle ear surgery (CO_2, argon, and KTP(Nd:YAG)) work by selective absorption of energy by either tissue chromophores or water. A wide range of laser-tissue interactions may result depending on the tissue absorption coefficient for a given wavelength and the laser pulse duration. Although lasers have been used clinically for treatment of chronic ear disease, the data documenting the potential thermal effects of these devices are insufficient.

The measurement of thermal trauma in middle ear surgery has focused on the analysis of histologic changes or measurement of temperature using thermocouples.[4-10] Although the occurrence of histologic changes is relevant, the absence of any structural change or damage in animal models does not preclude the existence of thermal injury, particularly when laser energy is used to treat tissue surrounding the cochlea and facial nerve. Thermocouples do provide precise information with respect to temperature change, but lack spatial resolution in that these devices typically are large, often as large as the region being ablated. Furthermore, thermocouples by design cannot measure the temperature in the ablation site, so only those tem-

From the Beckman Laser Institute and Medical Clinic, University of California, Irvine (Drs. Wong, Lee, Berns, and Neev), the Department of Otolaryngology–Head and Neck Surgery, University of California, Irvine (Dr. Wong), and the Department of Otolaryngology–Head and Neck Surgery, University of Virginia–Health Sciences Center (Dr. Hashisaki).

Supported by the following grants: ONR N0014-91-C-0134, DOE DE-FG03-91ER61227, and NIH 5P41RR01192. Dr. Wong is supported by the Research Fund of the American Otological Society.

Presented at the Annual Meeting of the American Academy of Otolaryngology–Head and Neck Surgery, San Diego, Calif., Sept. 18-21, 1994.

Reprint requests: Brian J. F. Wong, MD, the Beckman Laser Institute and Medical Clinic, University of California, Irvine, 1002 Health Sciences Rd. East, Irvine, CA 92715.

0194-5998/97/$5.00 + 0 **23/1/75718**

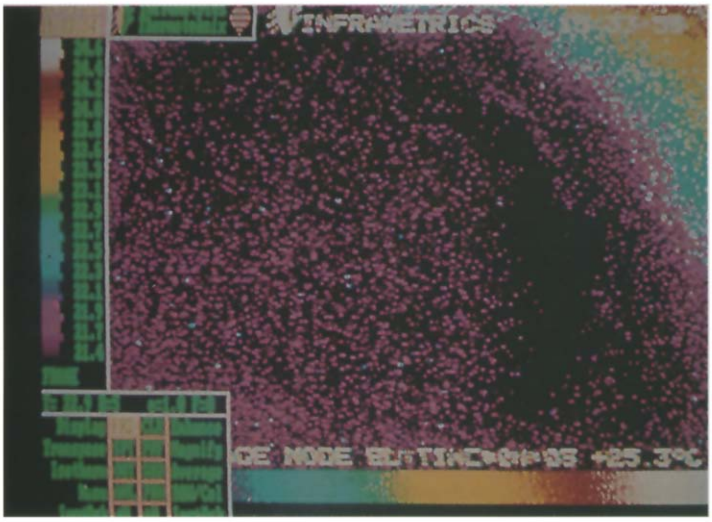

Fig. 1. Thermal camera image of the facial ridge in a cadaveric human temporal bone in the absence of laser exposure.

perature changes that occur adjacent to the area of treatment can be estimated. An additional limitation of thermocouples is that the temporal and thermal responses of these devices vary with the mass, thermal diffusivity, and thermal conductivity of the thermocouple tip.

Temperature changes in tissue treated with a CO_2 laser were determined using a thermal camera device, which provided precise temperature measurements with high spatial resolution and microsecond accuracy. Although analogous to traditional photographic devices, thermal imaging devices differ in several respects.[11] Thermal cameras function by detecting infrared radiation from the object of interest. Objects radiate energy according to their absolute temperature and in proportion to their surface emissivity. A single detector element is used to detect this infrared radiation. An image is obtained by raster scanning the optical lenses of the system with high speed oscillating mirrors.

METHODS AND MATERIAL

Mastoidectomies were performed in five preserved human temporal bones. The experiments were done in accordance with and with approval of the Human Subjects Institutional Review Board of the University of California, Irvine. After surgical exposure, the tissues were irrigated copiously with tap water and then placed in a saline bath at ambient temperatures for a week. The saline solution was changed on a daily basis. This was done to leach away as much of the preservative solution as possible. A thermal camera (detector sensitivity range: 8 to 12 μm) with a ×6 lens, 10.6 μm filter, and close-up lens was used to provide temperature images (Inframetrics Model 600 L, North Billerica, Mass.). The combined optical elements had a focal length of 6 inches. The detector element was made of mercury-cadmium-telluride with a time constant τ of 0.5 to 1.0 μsec. Detected temperature differences could be as small as 0.1° C. Thermal sensitivity was within 0.1° C. The detector measures temperature ranges varying from 5° C to 2000° C. The image was produced by a raster scanning method with two high speed oscillating mirrors for an overall sampling rate of 2.01 MHz. Because the real time images were captured on video film, the overall time resolution of this system was 30 ms, although the system can study phenomena in the order of 100 μsec under select conditions. The images were recorded using a professional quality video camera recorder and subsequently analyzed with a specialized software package (Thermoteknix Systems Ltd., Thermagram v 2.5, Cambridge, UK). The software allowed complex image processing and analysis of the recorded information. An Apple Quadra 840 A/V based video analysis system (Apple Computers, Cupertino, Calif.) also was used for additional image manipulation.

Laser ablation of the exposed otic capsule tissue was performed as previously reported.[1-3] A temporal bone was held in a standard holder and tightly secured. A CO_2 laser (Xanar XA-50, Johnson & Johnson, Colorado Springs, Colo.) was used to perform selective ablation of the otic capsule tissue. Energy was delivered in continuous mode with an exposure time of 100 msec. Power was varied from 300 mW to 4 W with a 200 μm spot size (diameter). This corresponded to a power density varying from 954 to 12,730 W/cm^2. The maximum

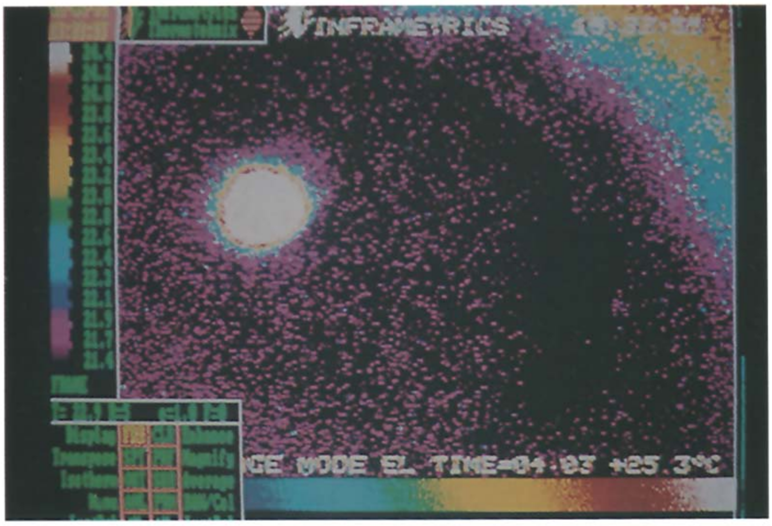

Fig. 2. The temporal bone illustrated here was treated with a CO_2 laser at 4 W power and a 200 μm focused spot size. This image was obtained at the end of a 100 msec laser pulse.

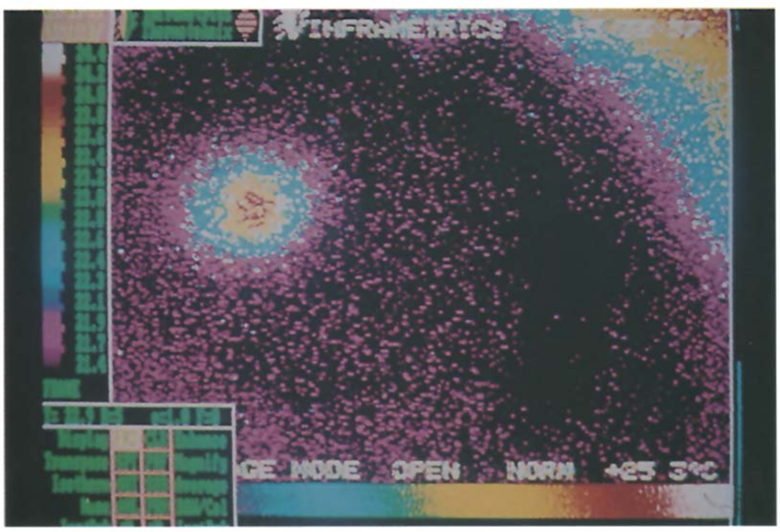

Fig. 3. Approximately 2 seconds have elapsed between this image and the previous image depicted in Fig. 2.

temperature was determined after the laser pulse. The area of the thermal disturbance was recorded. This area was demarcated by the return of the temperature to ambient values. In general, it was circular in shape. The time for the "hot spot" to return to ambient temperature also was recorded.

RESULTS

Figures 1 to 4 illustrate the thermal effect of a CO_2 laser acting on the facial ridge of a single temporal bone. Figure 1 illustrates the temperature distribution in the temporal bone before the laser exposure was applied. False color was used to enhance the visual contrast. Purple in this image represents ambient temperature (25° C). At the base of the image is a color scale, where higher temperatures are toward the right of the image. The information on the left side of the image is from the thermal analysis software. It should be emphasized that all these temperature readings reflect surface temperatures. Interaction effects were obvious to the naked eye at this energy setting. No cracks or fractures were noted in the vicinity of the ablation crater. The crater itself was characterized by brown discoloration, indicating tissue pyrolysis. The lateral extent of charring beyond the crater walls was minimal. The temporal bone illustrated in Fig. 2 was treated with a CO_2

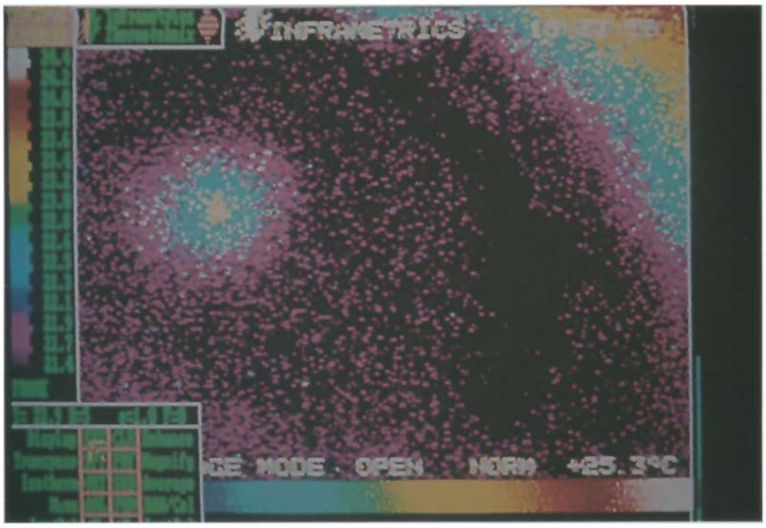

Fig. 4. Another 2 seconds have elapsed from the image in Fig. 3.

laser at 4 W power and a 200 μm focused spot size. This image was obtained at the end of a 100 msec laser pulse. The temperature recorded in the central white "hot spot" exceeded the temperature scale selected for this particular test. The width of the thermal disturbance for the white spot was approximately 2 mm. In Fig. 3, approximately 2 seconds had elapsed between this image and the previous image depicted in Fig. 2. The blue halo surrounding the central hot region reflects cooling and thermal diffusion in the bone tissue. Figure 4 illustrates further cooling after another 2 seconds had elapsed. For the data presented in these figures, the thermal camera was set with a temperature range of 5° C so that the temperatures that were approximately at ambient levels could be most clearly displayed in the color scale.

Figure 5 illustrates the thermal disturbance in the temporal bone as a function of laser power. The area (mm^2) reflects the region of the bone surface that was heated above ambient temperature by the laser pulse. In this series of investigations, the pulse duration was 100 msec in continuous mode operation for a CO_2 laser. As the power increased, the temperature of the area of bone that was heated to above ambient temperature increased correspondingly. This area varied from 100 mm^2 at high irradiance to 30 mm^2 at low irradiances. Figure 6 depicts the dependence of the surface "hot spot" temperature on laser power. The measurement of the peak temperature in the bone tissue was performed using a 1000° C temperature range on the thermal camera. The temperature was measured in the center of the thermal disturbance exactly 100 msec after the onset of the laser pulse. The temperatures recorded did not saturate the thermal camera, in contrast to the saturation noted in Fig. 2, where a 5° C temperature range was used. Notably, the temperatures exceeded hundreds of degrees celsius for energy settings that have been reported for use clinically. In Fig. 7, the time for the "hot spot" to return to ambient temperature is plotted against the laser power. Because these time measurements correspond to the thermal relaxation properties of the bone tissue under particular CO_2 heating conditions, the data may be used as a guide to determine cooling intervals between laser treatments. Measurement of this return to ambient temperature may be used to determine the interval between laser pulses.

DISCUSSION

Laser stapes surgery gained widespread acceptance in the treatment of disorders of the stapes footplate after Perkins'[12] successful laser stapedotomy using an argon laser. The use of lasers in the treatment of chronic ear disease has had less widespread acceptance, despite several studies demonstrating the use of these devices, because of concerns of thermal events during ear surgery.

In these applications, the temperature of the ablated surface is important for several reasons. The destruction of keratinocytes may be accomplished using an energy density that does not cause obvious ablation (charring in the case of CO_2). Lower energy would theoretically decrease heat conduction to critical underlying neuroepithelial structures in the otic capsule. Even minimal heat transfer to the perilymph may result in thermal damage to the surrounding neuroepithelium and potentially create sensorineural hearing deficits or

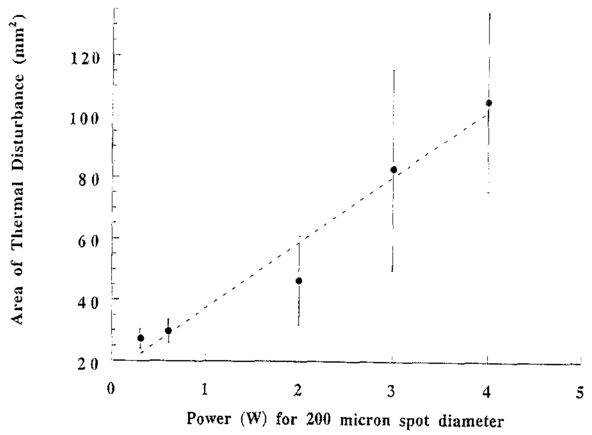

Fig. 5. The thermal perturbation is illustrated as a function of power.

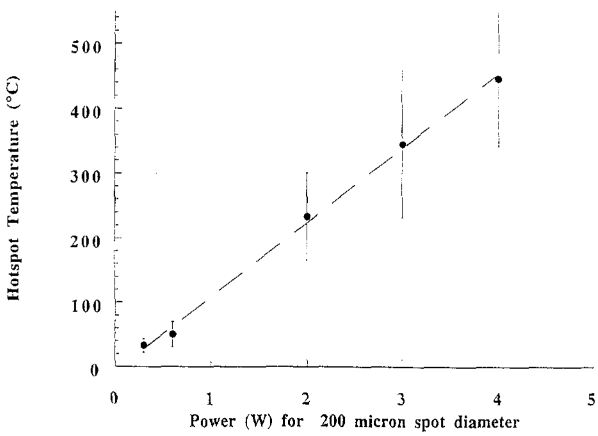

Fig. 6. "Hot spot" temperature is illustrated as a function of power.

dizziness. Moreover, the laser spot size is under the control of the surgeon. This is critically important in fiber optic hand-piece based systems, where the focal size of the spot may vary considerably with the hand motions of the surgeon. The resultant energy density may hence vary considerably. Safe parameters need to be determined.

CO_2, KTP(Nd:YAG), and argon lasers currently dominate clinical practice in laser ear surgery,[1,3,12-19] although excimer and pulsed infrared lasers (such as holmium:YAG and erbium:YAG) are readily available for clinical use and have been used in several models of stapes surgery.[4,6-9,20-22] The investigation of photothermal effects in laser ear surgery has been very limited. Several investigators have addressed this issue in animal models, but they used thermocouples and models that only approximated clinical conditions.[4-6,8-10] This study is the first to provide information on the surface thermal characteristics of lasers used in treating the

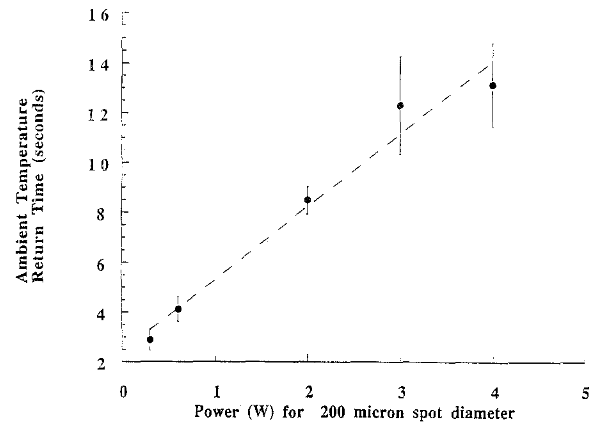

Fig. 7. The time for the "hot spot" to return to ambient temperature is illustrated as a function of laser power.

temporal bone.

Our infrared temperature measurements demonstrate the linear dependence of "hot spot" temperatures on power density. Thermally perturbed areas also depended linearly on power density, as did the time for the "hot spot" temperature to return to ambient levels. These observations suggest that absorption characteristics (and energy deposition) as well as heat removal rates did not change as a function of power density in our study. Thermal modeling as well as additional measurements (now underway) will extract quantitative values for these characteristic parameters (thermal conductivity and diffusivity) over a broader range of time exposures and in other tissues, such as cortical (substantia compacta) and lamellar bone (substantia spongiosa) in addition to otic capsule. Such complete analysis will provide a rational and empirical basis for safe and predictable laser use in the surgical treatment of chronic ear disease.

Even without detailed thermal modeling, however, the fact that we detected no cracks or fractures caused by thermal or mechanical stresses, along with the confinement of carbonization effects to the ablation crater alone, is very encouraging. This suggests that with proper parameter selection, CO_2 laser surgery in hard tissue can be accomplished safely while achieving the desired ablation effects.

In the surgical treatment of chronic diseases of the ear, possible laser applications include the destruction of residual cholesteatoma, fibrotic tissue, and diseased mucosa. The selective destruction of these tissues may be accomplished using an energy density that does not cause obvious ablation (charring in the case of CO_2) and hence less "thermic loading" of the surrounding tissue. The current technology used in clinical practice does not provide the information necessary to monitor and determine temperature adequately. The power-temperature-cooling curves presented in this article have practical application in minimizing the thermal injury during laser surgery of the ear. The need for accurate temperature determination will become more important. As low cost, portable and even hand-held diode laser systems become more readily available, the laser may find greater use in surgical treatment of chronic diseases of the ear. Transtympanic, percutaneous, and even photodynamic methods to treat chronic ear disease may become possible and offer significant cost reductions over current methods that are heavily dependent on drills and extensive mastoid exposure. Infrared imaging devices, as this study demonstrates, may be of use in monitoring and determining temperature during laser surgery of the ear.

We thank J. Stuart Nelson, MD, PhD, and Karen J. Doyle, MD, PhD, for their review of the manuscript, and Sam Tannenbaum, PhD, for his thoughtful discussions. We also thank Glen Profeta, Jeff Andrews, and David Kang for their technical expertise.

REFERENCES

1. Schindler RA, Lanser MJ. The surgical management of cholesteatoma—the contribution of the laser. In: Tos M, Thomsen J, Peitersen E, eds. Third International Conference on Cholesteatoma and Mastoid Surgery. Copenhagen, Denmark June 5-9, 1988: Amsterdam: Kugler and Ghedini; 1988. p. 769-78.
2. Thedinger BS. Applications of the KTP laser in chronic ear surgery. Am J Otol 1990;11(2):79-84.
3. McGee TM. The argon laser in surgery for chronic ear disease and otosclerosis. Laryngoscope 1983;93:1177-82.
4. Bottrill ID, Perrault DF, Pankratov MM, Poe DS. Thulium:YAG laser for stapes surgery: preliminary observations. Proceedings SPIE 1994;2128:23-30.
5. Jovanovic S, Schonfeld U, Berghaus A, Fischer R, Scherer H. CO_2 laser in stapes surgery. Proceedings SPIE 1993;1876:17-27.
6. Lesinski SG, Palmer A. Lasers for otosclerosis: CO_2 vs. argon vs. KTP-532. Laryngoscope 1989;99:1-8.
7. Shah UK, Poe DS, Rebeiz EE, Perrault DF, Pankratov MM, Shapshay SM. Erbium laser in middle ear surgery: in vitro and in vivo animal study. Proceedings SPIE 1994;2128:12-6.
8. Stubig IM, Reder PA, Facer GW, Rylander HG, Welch AJ. Holmium: YAG laser stapedotomy: preliminary evaluation. Proceedings SPIE 1993;1876:10-6.
9. Thoma J, Mrowinski D, Kastenbauer ER. Experimental investigations on the suitability of the carbon dioxide laser for stapedotomy. Ann Otol Rhinol Laryngol 1986;95:126-31.
10. Gantz B. Argon laser stapedotomy. Ann Otolaryngol 1982;91: 25-6.
11. Torres JH, Springer TA, Welch AJ, Pearce JA. Limitations of a thermal camera in measuring surface temperature of laser-irradiated tissues. Lasers Surg Med 1990;10:510-23.
12. Perkins RC. Laser stapedotomy for otosclerosis. Laryngoscope 1980;90:228-41.
13. Bartels LJ. KTP laser stapedotomy: Is it safe? Otolaryngol Head Neck Surg 1990;103(5):685-92.
14. Causse J-B, Gherini S, Horn KL. Surgical treatment of stapes fixation by fiberoptic argon laser stapedotomy with reconstruction of the annular ligament. Otolaryngol Clin North Am 1993;26:395-416.
15. Gherini S, Horn KL, Causse J-B, McArthur GR. Fiberoptic argon laser stapedotomy: Is it safe? Am J Otol 1993;14:283-9.
16. Hodgson RS, Wilson DF. Argon laser stapedotomy. Laryngoscope 1991;101:230-3.
17. Lesinski SG, Newrock R. Carbon dioxide lasers for otosclerosis. Otolaryngol Clin North Am 1993;26(3):417-41.
18. Lesinski SG, Palmer A. CO_2 laser for otosclerosis: safe energy parameters. Laryngoscope 1989;99:9-12.
19. Silverstein H, Rosenberg S, Jones R. Small fenestra stapedotomies with and without KTP laser: a comparison. Laryngoscope 1989;99:485-8.
20. Kautzky M, Trodhan A, Susani M, Schenk P. Infrared laser stapedotomy. Eur Arch Otorhinolaryngol 1991;248:449-51.
21. Schlenk E, Profeta G, Nelson JS, Andrews JJ, Berns MW. Laser assisted fixation of ear prosthesis after stapedectomy. Laser Surg Med 1990;10:444-7.
22. Segas J, Georgiadis A, Christodoulou P, Bizakis J, Helidonis E. Use of the excimer laser in stapes surgery and ossiculoplasty of middle ear ossicles; preliminary report of an experimental approach. Laryngoscope 1991;101:186-91.