The 9.3-µm CO_2 Dental Laser: Technical Development and Early Clinical Experiences

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SUMMARY

This article summarizes some of the findings of scientific research involving the 9.3-µm laser that led to the development of this latest innovation while referencing its roots in the 10.6-µm soft-tissue-only laser. It describes how use of isotopic carbon dioxide gas as the laser's active medium furnishes the 9.3-µm wavelength with unique hydroxyapatite absorption characteristics for optimum energy transfer into dental hard tissues while it retains the excellent soft tissue surgical abilities of conventional CO_2 at 10.6 µm. A summary of laser device considerations that led to the first practical implementation of this new wavelength for the clinical environment is presented. Selected clinical capabilities of both the 9.3-µm and erbium lasers are then compared. Finally, a report of clinician experience with and patient response to this new laser instrument is provided.

BACKGROUND

While new to clinical dentistry, the 9.3-µm CO_2 laser can be viewed as the culmination of more than 27 years of scientific dental research devoted to this specific wavelength. The 9.3-µm laser brings speed and efficiency to enamel and dentin cutting not experienced heretofore using CO_2, a medium that many considered to be the gold standard of soft tissue surgical lasers. The difference is the new wavelength, 9.3 µm vs. the traditional 10.6 µm that CO_2 is known for.

In 1964, four years after Theodore H. Maiman[1] developed the world's first laser (a ruby laser), C. Kumar N. Patel of Bell Laboratories developed the first 10.6-µm carbon dioxide gas laser.[2] His initial system delivered milliwatts of power, but soon thereafter he was able to achieve several Watts of power with the addition of other gases to the laser tube[3] – nitrogen and helium, used to increase pressure, stabilize the electrical discharge process, cool the laser gas, and store and transfer energy to the laser gas.[4]

Today, carbon dioxide lasers at their native 10.6-µm wavelength are known for their versatility, robustness, and reliability and make them some of the most widely used lasers in industry[5] and also in medicine, with its near-complete absorption by most biologic tissues.[6] Typical industrial applications include cutting of metals and nonmetals, welding, heat treating, cladding, marking, and drilling.[5] Surgical applications are found in gynecology, otolaryngology, pulmonary medicine, neurosurgery, aesthetic surgery, dermatology, plastic surgery, gastroenterology, urology, general surgery, orthopedics, oral surgery and dentistry, ophthalmology, and cardiovascular surgery.[7]

BIOMEDICAL INTEREST IN THE CARBON DIOXIDE LASER

Not long after Patel's achievement, interest in the possible surgical applications of the carbon dioxide developed. As Polanyi noted, "Early in 1966, Yahr, a senior surgical resident at Montefiore Hospital in New York, started surgical experiments with a CO_2 laser and recognized its potential surgical utility."[8] That same year, Yahr and Strully stated: "Using a nitrogen carbon-dioxide laser…clean dry skin incisions which heal as well or better than scalpel controls were fashioned."[9]

Then in 1970, Polanyi and colleagues wrote:

> A renewed interest in laser surgery has taken place since the introduction of the CO_2 laser to the medical field. The major characteristics of this laser that make it interesting for surgical applications are that it is a high power continuously operating laser and that its wavelength of operation is in the infra-red at 10.6 µm, a wavelength that is almost completely absorbed by most biological tissues.[10]

Strong and Jako reported the first clinical use of a carbon dioxide laser in otolaryngology in 1972.[11] From there, numerous accounts of the intraoral use of the 10.6-µm laser were published in the medical literature.[12-24]

CHARACTERISTICS OF THE 9.3-µm CO_2 LASER

Most reports of the surgical use of the carbon dioxide laser focused on the 10.6-µm wavelength. Migliore points out that while CO_2 lasers can produce power at any wavelength between 9 and 11 µm, most units operate at 10.6 µm because that transition has the greatest gain (or degree of amplification). He continues: "There are several ways to force a CO_2 laser to operate at other than 10.6 µm: Some manufacturers use a Littrow grating [a type of diffraction grating or reflective surface with fine lines used to produce optical spectra by diffraction], which allows complete tunability through the CO_2 wavelength range, while others replace the normal $^{12}C^{16}O_2$ molecule with isotopic $^{12}C^{18}O_2$…"[5]

The use of isotopic $^{12}C^{18}O_2$ as a laser gas to produce the 9.3 µm wavelength is fundamentally more efficient and less disruptive to the laser manufacturing process than the use of a Littrow grating. CO_2 lasers are molecular lasers, as the lasing medium is carbon dioxide, a molecule. Molecular lasers function differently from atomic lasers, like neodymium or erbium, because they have vibrational and rotational energies as well as electronic energy. These molecular vibrations occur because the relative orientations and positions of the atomic nuclei are not absolutely fixed within the molecule. The molecular rotations occur because the individual molecules are free to rotate and spin in space since they are in a gaseous state. The energies related to molecular vibrations and rotations are quantized just like the electronic energy also in atomic-based lasers. Transitions between vibrational energy levels emit photons with wavelengths in the infrared region, and transitions between rotational energy levels emit photons in the microwave region.

The CO_2 laser is based on the vibrational and rotational transitions of the CO_2 molecule used to lase. This molecule consists of two oxygen atoms covalently bonded to a central carbon atom. In diatomic molecules such as O_2, N_2, or CO, the individual atoms are bound by a molecular binding force, and when excited, the two nuclei will vibrate much like two masses connected by a spring. Although carbon dioxide is a triatomic molecule, it behaves similarly to a simple diatomic molecule because its structure is linear. Such a linear triatomic molecule has three normal modes of vibration, depicted as the asymmetric stretch mode, the bending mode, and the symmetric stretch mode (Figure 1).[25]

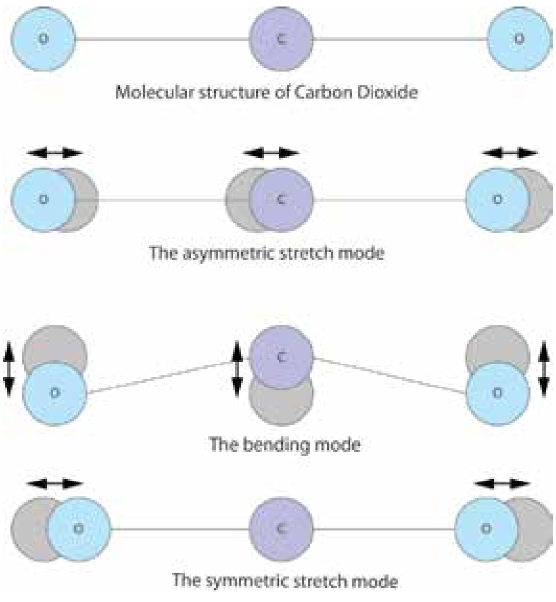


***Figure 1:** Vibrational states of the carbon dioxide molecule. Adapted from Kverno and Nolen[25]*

Each of these normal modes of vibration for the CO_2 molecule is associated with a characteristic frequency of vibration (w) as well as a ladder of allowed energy levels. To manufacture a 9.3-µm CO_2 laser versus a 10.6-µm CO_2 laser, the gas inside the laser is changed from a mixture with $^{12}C^{16}O_2$ molecules to a mixture with $^{12}C^{18}O_2$ molecules. Because ^{16}O is the most abundant stable isotope of oxygen (99.762% natural abundance) it is not usually referred at as an isotope, and because the ^{18}O atom is less abundant, it is commonly referred to as an oxygen isotope, even though both are naturally occurring and stable.[26] Consequently $^{12}C^{18}O_2$ gas used in CO_2 lasers to produce the 9.3-µm energy level is referred to as isotopic CO_2 gas. Having two extra neutrons ^{18}O has a different weight than ^{16}O, and therefore its three normal modes of vibration, described as the asymmetric stretch mode, the bending mode, and the symmetric stretch mode, are distinct from the vibrational modes of ^{16}O. Hence the heavier ^{18}O produces the 9.3-µm wavelength of energy while the ^{16}O produces the 10.6-µm wavelength. Consequently, changing the composition of the CO_2 laser gas inside the laser resonator is all that is required to produce the 9.3-µm wavelength, all other aspects of the laser can remain identical to the standard production CO_2 laser construction, and a "special" customized laser is not required. Standardization in manufacturing on one base platform leads to higher volume, more reliability, and increased repeatability.

The 9.3-µm CO_2 laser wavelength is of particular interest because it better matches the absorption characteristics of hydroxyapatite, "HA" (the principal mineral component of hard tissue), than the 10.6-µm laser, a characteristic that has significance for operative dentistry.[27-29] Because HA is a rather large and complex molecule, $Ca_{10}(PO_4)_6(OH)_2$, its vibrational energy is in the far infrared and is narrow band (Figure 2).

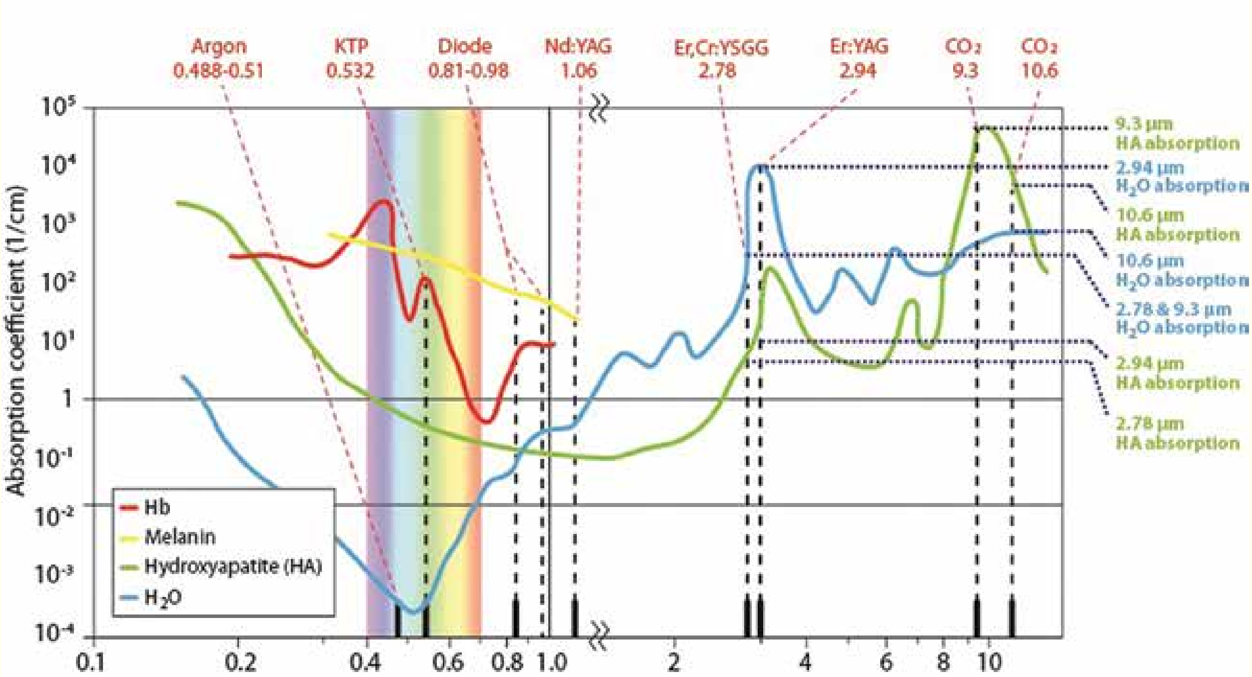


Figure 2: *Absorption curves of various tissue components. Absorption of water is depicted by the blue line, and hydroxyapatite by the green line. Adapted from Featherstone and Fried[30] and Vogel and Venugopalan[31]*

The absorption of HA is shown with the green curve with its peak absorption at 9.6 µm and drops off rapidly between 9.2 µm to 10 µm.[30] Determining the wavelength where peak energy is transmitted into the enamel is not strictly a function of HA's absorption; reflection (scattering) and transmittance also have to be considered. For peak absorption into the enamel, the light energy has to couple into the tooth (not reflect or scatter at the surface), and then be absorbed and not transmitted. HA's absorption peak is at 9.6 µm but HA's reflection also peaks at 9.6 µm. While the absorption of HA is lower at 9.3 µm, the reflection is lower, too. Transmittance at 9.3 to 9.6 µm is negligible. When one considers both reflection and absorption, the total energy transferred into HA is the greatest at 9.3 µm where the reflection coefficient is low and the absorption coefficient is still fairly high. The importance of maximum absorption is understood when trying to achieve the peak energy transfer from the incident laser light into the HA. Maximum energy transfer means the greatest possible opportunity for the rapid ablation and vaporization of enamel and dentin.

SCIENTIFIC INVESTIGATION OF THE 9.3-µm LASER WAVELENGTH IN DENTISTRY

More than 90 published reports have appeared in the dental literature since 1986 that describe the basic science, tissue interactions, mechanisms of action, and operational parameters of the 9.3-µm CO_2 laser wavelength on oral hard and soft tissue. The following summarizes some of the key findings that support the safe and effective use of this laser. While this paper concentrates on the tissue interactions with the 9.3-µm wavelength, selected investigations of other CO_2 wavelengths (including 9.5, 9.6, 10.3, and 10.6 µm) of historical interest are briefly examined to underscore the differences in their tissue interactions and to provide a sense of the scope of research evident in the literature.

HARD TISSUE ABLATION

In 1993 Forrer *et al.* investigated the mechanism of bone ablation using 9.3, 9.6, 10.3, and 10.6 micron CO_2 lasers with different pulse parameters on samples of pig rib bone. Among other findings, they determined that the ablation threshold at 9.3 and 9.6 µm is lower than at 10.3 and 10.6 µm, possibly due to the higher absorption of the mineral component in the bone by the 9.3 and 9.6 µm wavelengths.[32]

Fried and colleagues noted in 1994 that the more efficient absorption of the 9.3 and 9.6 µm laser wavelengths (vs. 10.3 and 10.6 µm) in human and bovine enamel samples "may be more advantageous for fusing enamel surfaces while minimizing the thermal loading of the pulp chamber."[33]

In 1995 Wigdor and associates examined the use of the CO_2 laser for intraoral soft tissue surgery, reviewed the optical properties of hard dental tissues, and surveyed the interaction of different lasers (including the CO_2 laser) with hard tissue.[34]

Also in 1995 a group of researchers from the Eastman Dental Center in Rochester, New York, the University of Rochester, and the University of California – San Francisco (UCSF) reported results of their scanning electron microscopic study of the effects of a tunable CO_2 laser on extracted bovine and human enamel. Specimens were irradiated at various fluences (laser energies per unit area) without water spray. At 10.6-µm, no observable surface melting at low fluences was noted, but "distinct surface cracks associated with the thermal stress created by the heat of the laser beam" were evident; higher fluences were required to melt the enamel and fuse the crystals. In contrast, the 9.3-µm produced uniform surface melting at low fluences; enamel crystals were fused to varying degrees and droplets of recondensed enamel mineral surrounded the area of irradiation. The authors conclude that the commonly available 10.6-µm wavelength "behaves quite differently from the more efficiently absorbed" 9.3-µm wavelength.[35]

Dentin surface and subsurface effects were the focus of a joint research investigation conducted by Showa University School of Dentistry in Tokyo and the University of California, Irvine, California and reported in 2000. Extracted human teeth were irradiated with one pulse of a 9.3-µm CO_2 laser set at different parameters, without water spray. Morphological changes were observed by scanning electron microscopy (SEM) and confocal laser scanning microscopy (CLSM). SEM images revealed some molten and resolidified dentin particles, but no cratering or cracking. Small cracks were seen in the subsurface layer using CLSM. Judging from CLSM photographs, the authors observed that most of the irradiated energy seemed to be concentrated at the surface (less than approximately 20 µm), compared to the approximate 60-µm depth seen in 10.6-µm laser-irradiated dentin surfaces in a previous study conducted by the group.[36]

In 2006, Fan, Bell, and Fried used a 9.3-µm CO_2 laser at high repetition rate (from 10 to 400 Hz) for the ablation of enamel and dentin. Lateral incisions were produced in sections of extracted, unerupted human molars and premolars using a computer-controlled scanning stage, with and without water spray. They found that the laser operating at high repetition rates with a pulse duration of 15 to 20 µs was able to ablate hard dental tissues efficiently and at a practical clinical rate with minimal thermal damage if a low-volume water spray is used. The single pulse ablation rate varied from 9 to 20 µm per pulse for enamel and from 30 to 40 µm per pulse for dentin. The investigators also measured the residual heat remaining in the tooth after each ablative pulse at the optimal irradiation intensities for efficient ablation: 34% and 20% of the energy remains in the tooth for enamel and dentin, respectively. This contrasts with the levels of 50% to 60% observed for conventional 10.6-µm and Er:YAG and Er:YSGG laser pulses; hence, "the residual energy remaining in the tooth for those laser systems is twice as high, which increases the volume of water that is needed to offset heat accumulation."[37]

Darling and Fried observed that enamel ablation rate and efficiency by a 9.3-µm CO_2 laser can be affected by the incident laser fluence, depth of cut, enamel crater morphology, geometry, and the amount of water present; stalling may occur which in turn leads to increased heat accumulation and thermal damage. Their study showed that the ablation stalls as the crater reaches a certain depth and aspect ratio when the laser beam is scanned along one axis. They report that "the laser needs to be scanned in two dimensions if high repetition rates are used, not only to ensure efficient ablation, but to also avoid excessive heat accumulation that may cause pulpal overheating and injury."[38]

In 2010 Maung, Lee, and Fried compared the mechanical damage to enamel after drilling with a mechanically scanned 9.3-µm laser and a dental high-speed handpiece. Water spray was used to observe the effect on sections prepared from extracted human molars and incisors, 15 samples for each treatment. Fewer cracks were produced by laser irradiation (2 of 15 samples) than by the high-speed handpiece (9 of 15 samples).[39]

FLUENCE AND ABSORPTION DEPTH

The amount of energy delivered to a tissue must be sufficient to have the desired effect, but no more than necessary because extraneous energy can be absorbed by surrounding tissue causing thermal stress, or pulpal damage and death to the tissue. Therefore, for the ablation of enamel or dentin one must choose the proper wavelength, so the absorption is high enough to contain the energy deposition near the surface, the right wavelength to avoid unnecessary scattering, the correct energy to be above the ablation threshold, but also the correct pulse duration so that sufficient energy can be delivered in an optimum period of time to ablate without stalling. For optimum ablation or caries prevention the pulse duration of the appropriate wavelength laser should be matched approximately with the tissue thermal relaxation time. The higher the absorption coefficient, the smaller the absorption depth (Table 1).[30]

Therefore pulse energies should be above the ablation level, with pulse widths approaching the thermal relaxation times and with a high repetition rate pulsing to rapidly remove tissue one absorption depth at a time, to achieve the finest and most precise tissue ablation.

Table 1: Comparison of CO_2 and Erbium Enamel Properties
(Derived from Featherstone and Fried[30])

Wavelength	Absorption Coefficient, μ_a, cm^{-1}	Absorption Depth (1/e*), µm	Thermal Relaxation Time, µs	Scattering Coefficient, μ_s, cm^{-1}
Near-Infrared				
2.78 µm Er:YSGG	480	25	220	~0
2.94 µm Er:YAG	800	12	90	~0
Mid-Infrared				
9.3 µm CO_2	5,500	2	2	~0
9.6 µm CO_2	8,000	1	1	~0
10.3 µm CO_2	1,125	9	40	~0
10.6 µm CO_2	825	12	90	~0

** Absorption depth is where the intensity of the surface light has been reduced 1/e (base of the natural log) = 63% reduction*

CARIES REMOVAL

In 1998 Takahashi, Kimura, and Matsumoto from the Showa University School of Dentistry in Tokyo examined the effects of a 9.3-µm CO_2 laser on extracted human teeth. Their scanning electron microscopic examination of tooth surfaces showed that sound and carious hard tissues, smear layer, and debris could be removed by laser irradiation. Moreover, the effects were dependent on the level of laser energy: "Both sound and carious dentin were shown to be more sensitive to laser irradiation than enamel and the carious enamel and dentin reacted with more sensitivity … than did sound enamel and dentin."[40]

The characteristic that the 9.3-µm wavelength is more efficiently absorbed by dental hard tissue than the conventional 10.6-µm CO_2 wavelength was exploited in a study of successful caries removal from extracted human molars with artificially created caries-like lesions. The researchers, from Okayama University in Japan (Konishi) and UCSF (Fried, Staninec, and Featherstone), used a computer-controlled motion control system to scan the laser beam over the target tooth surface.[41]

The notion of preferential removal of carious lesions was examined by another group of UCSF investigators who used advanced imaging technologies (two-dimension near-infrared imaging and polarization-sensitive optical coherence tomography) to guide 9.3-µm CO_2 laser ablation. Their pilot system selectively removed artificial demineralized lesions on extracted bovine incisors, while sound enamel was conserved with minimal damage to sound tissue. They noted that the caries removal rate varied with the severity of demineralization; their findings "indicate the challenges involved in the selective removal of natural caries including overcoming the highly variable topography of the occlusal pits and fissures, the highly variable organic/mineral ratio in natural lesions, and the more complicated lesion structure."[42]

BOND STRENGTH

In 2009 a group of UCSF researchers compared the shear bond strength of dentin to composite after treatment with a 9.3-µm CO_2 laser, with and without post-ablation acid etching to conventionally prepared dentin surfaces. Dentin blocks were prepared from extracted noncarious human molars. The shear bond test was used to determine the adhesive strength of the bonding agent to dentin. The bonding material was Single Bond along with the Z-250 composite (3M, Minneapolis, Minn.). The negative control group was neither irradiated nor etched. The positive control group was only etched with 35% phosphoric acid. One group of laser-treated blocks was etched with 35% phosphoric acid, and the other group of laser-treated blocks was not etched. Bond strengths exceeded 20 MPa for laser-treated surfaces that were acid-etched after ablation; this figure represents a clinically useful bond strength, as noted by the researchers. Moreover, the respective bond strengths of dentin to composite for the etched and water-cooled irradiated groups were significantly higher than the negative control which indicates that etching following ablation is a benefit.[43]

In a subsequent study, Nguyen *et al.* from UCSF measured the adhesion strength of 9.3-µm CO_2 laser-treated enamel and dentin surfaces for various laser scanning parameters with and without post-ablation acid etching using the single-plane shear test. The bonding resin was Single Bond with Z-250 composite (3M-ESPE, Minneapolis, Minn.). Positive control groups were not irradiated by laser but etched with 35% phosphoric acid. Negative control groups were neither irradiated by the laser nor acid-etched. The mechanical strength of laser-ablated dentin surfaces was determined via the four-point bend test and compared to control samples prepared with 320-grit wet sand paper to simulate conventional preparations. Four-point bend tests yielded mean mechanical strengths of 18.2 N (s.d. = 4.6) for ablated dentin and 18.1 N (s.d. = 2.7) for control. Shear tests revealed mean bond strengths approaching 30 MPa for both enamel and dentin under certain irradiation conditions. The researchers noted that respective bond strengths of dentin to composite for the etched and water-cooled irradiated groups were significantly higher than the negative control, which indicates that etching following ablation is beneficial. They concluded that dental hard tissues can be rapidly ablated with a mechanically scanned CO_2 laser at high pulse repetition rates that produce no significant reduction in the tissue's mechanical strength or a major reduction of adhesive strength to restorative materials.[44]

TEMPERATURE, PULPAL SAFETY, LACK OF COLLATERAL DAMAGE

In 1996 a group of investigators from UCSF, the University of Rochester, and the Eastman Dental Center in Rochester, New York collaborated on an examination of the thermal response of hard dental tissues to different CO_2 laser wavelengths. Sections of human and bovine enamel and human dentin were irradiated without water spray. Their *in vitro* results revealed, among other things, that the use of highly absorbed laser light (e.g., 9.3 µm) tends to localize the heating to a thin layer at the sample surface. They report: "Minimizing of the absorption depth significantly decreases the risk of subsurface thermal damage, since less energy is required to heat the surface to the required temperature for a successful treatment" and "The more commonly employed 10.6-µm CO_2 laser wavelength leads to cumulative heating that endangers the pulp."[45]

In 2000 Walsh and colleagues compared the effect of a prototype pulsed CO_2 laser emitting at 9.6 µm with a conventional dental handpiece on the pulps of human teeth scheduled for extraction for orthodontic or periodontal reasons. Teeth were extracted immediately after and 14 days after treatment and then examined histologically; the laser had very limited effect on the odontoblasts and pulpal tissue. The researchers found that the laser had an equal effect on the dental pulpal tissue when compared to the dental handpiece.[46] Two years later, Wigdor and Walsh reported similar findings in beagle dogs.[47]

Nair and associates performed a similar study on human third molars scheduled for extraction using a pulsed 9.6-µm CO_2 laser. Their preliminary histological results reported in 2005 suggested that the laser "induced only minimal response of the dentine-pulp complex when used as a hard-tissue drilling tool…"[48]

In 2006 another group of UCSF investigators used a 9.3-µm CO_2 laser with and without water spray on sections of dentin from human unerupted third molars to determine extent of thermal damage after irradiation at various laser fluences and repetition rates. They found that high repetition rates were feasible with very little thermal damage in both wet and dry conditions: "Polarized Light Microscopy showed thermal damage zones of less than 22-µm for dry dentin and less than 16-µm for wet dentin for all repetition rates less than 400-Hz. These experiments were performed with a fluence of 80-J/cm^2 and the results were compared to 40-J/cm^2 and 20-J/cm^2. The thermal damage zones at 40-J/cm^2 and 20-J/cm^2 were both smaller than those found at 80-J/cm^2 at 400-Hz." The authors indicate there was negligible charring of dentin for laser pulses of less than 10-µs duration, and no mechanical damage was observed for any of the laser irradiation parameters.[49]

In 2008, Assa, Meyer, and Fried reported their use of a pulsed 9.3-µm carbon dioxide laser with an integrated/programmable optical scanner to scan the laser beam over the targeted area of human and bovine samples to remove enamel and dentin tissue "without excessive peripheral thermal damage or heat accumulation."[50]

In the temperature portion of their 2011 study of the effects of a pulsed 9.3-µm CO_2 laser on noncarious extracted teeth, a team of UCSF researchers used computer-controlled XY galvanometers to scan the laser beam over sample surfaces. Microthermocouples placed within the pulp chamber measured the temperature rise during a 2-minute irradiation period with the laser set at a pulse repetition rate of 300 Hz and two laser fluences (20 J/cm^2 and 30 J/cm^2). The scanner permitted the ablation of a 5.0-mm diameter cylindrical pattern over the occlusal surface of the tooth, under water spray. Heat accumulation measurements indicated that the mean temperature change at 20 J/cm^2 was 2.0 $\pm$ 0.6°C, and at 30 J/cm^2 was 3.2 $\pm$ 0.8°C. The authors note that a temperature rise of 5.5°C is indicative of excessive heat accumulation and can lead to pulpal necrosis. For the peripheral thermal damage investigation portion of the study, they used polarized light microscopy (considered the most effective way to visualize thermal damage) to look for changes in birefringence of the collagen. No thermal changes in the dentin samples were seen in the polarized light micrograph, indicating the zone of thermal damage is less than 10 µm. The investigators conclude: "These results suggest that dental hard tissues can be rapidly ablated with a mechanically scanned CO_2 laser at high pulse repetition rates without excessive heat accumulation in the tooth or peripheral thermal damage."[44]

Another team of UCSF investigators conducted a clinical study involving 29 patients requiring removal of third molars. A pulsed 9.3-µm CO_2 laser was used to irradiate the occlusal surfaces of the teeth prior to extraction for 1 minute at 50 Hz and 2 minutes at 25 Hz with a fluence of 20 J/cm^2 and 12-15 mJ/pulse, with water spray, for a total number of 3,000 pulses per tooth. Seventeen teeth treated at 50 Hz were extracted within 72 hours, and 11 were extracted after 3 months. At 25 Hz, 7 teeth were extracted at 72 hours and 10 teeth after 3 months. These time periods were selected to detect reversible pulpal changes such as inflammation (after 72 hours) and permanent changes to the pulp (after 3 months). No significant adverse events occurred; no analgesic was used and 2 patients reported discomfort which was attributed to cold sensitivity from the air-water spray (the laser was turned off to confirm that the water spray alone caused the discomfort). Histologically, all teeth showed a lack of pulpal response, regardless of treatment category and time extracted post-treatment. Limitations of the study included laser treatment of enamel only (no dentin ablation was involved) and restricted repetition rate (25 and 50 Hz). The authors state that the 9.3-µm CO_2 laser should be electronically scanned to expose a new area for each pulse for most effective tissue ablation.[51]

INTRAORAL SOFT TISSUE SURGERY

Because the 10.6-μm laser is well recognized as an effective device for soft tissue surgery, researchers were interested in investigating the soft-tissue capabilities afforded by its 9.3-μm wavelength counterpart. In 1995, Wilder-Smith and colleagues from the University of California – Irvine reported on a comparison of continuous-wave 9.3- and 10.6-μm CO_2 lasers used to make standardized incisions in the oral mucosa of pig mandibles. Thermal analysis revealed that the danger of thermal damage to adjacent structures during laser incision of soft tissues at either wavelength is minimal unless the adjacent structures are impacted by the laser beam. Histologic examination showed little difference between the 9.3- and 10.6-μm lasers equipped with coherent delivery systems. They concluded that thermal and histologic effects were related to the parameters used and beam characteristics rather than wavelength.[52]

Two years later, Wilder-Smith, Dang, and Kurosaki used pig mandibles to determine the range of clinical incision effects in soft tissue using a 9.3-μm CO_2 laser operating in various pulsed modes. Their histologic findings indicated that with this laser "a wide range of clinical effects can be achieved consistently and predictably in soft tissue, depending on the parameter configuration selected."[53]

In 1998 Payne and associates examined the ablation processes by performing mass removal and thermal injury experiments on porcine dermis at wavelengths where tissue water is the primary absorber (10.6 μm) and where water and collagen have comparable absorption (9.5 μm). They noted that pulsed 9.5-μm laser irradiation removed tissue more efficiently and with a smaller zone of thermal injury than at 10.6 μm. They speculated that "these differences may be due to the weakening of the tissue mechanical strength by directly targeting collagen."[54]

While reporting on the incisional and collateral effects of a 9.3-μm CO_2 laser on soft tissue in 1997, Wilder-Smith *et al.* acknowledged the 1995 report by McCormack and colleagues concerning the effect of various CO_2 laser wavelengths on dental hard tissue and thereby recognized the possible multifunctionality of the 9.3-μm wavelength which "better matches the absorption characteristics of hydroxyapatite than 10.6 μm, providing the possibility for modification or efficient ablation of hard dental tissues without thermal damage to adjacent and pulpal structures. Because lasers are relatively costly devices, multiple applications are desirable if this type of equipment is to become realistically useful for clinicians."[27]

9.3-μm CO_2 LASER DEVICE CONFIGURATION

As noted above, Wilder-Smith *et al.* recognized over 15 years ago that "because lasers are relatively costly devices, multiple applications are desirable if this type of equipment is to become realistically useful for clinicians."[27] The inherent device design flexibility required to accommodate "multiple applications" is as challenging as the laser physics explained above. Melding the relevant technical discoveries and benefits found in the 27 years of research into a laser device platform creates multiple challenges. The overview listed below explains the importance of the relevant research discoveries, the related clinical benefit, and the associated device design feature to achieve the clinical benefit:

1. For the rapid ablation of enamel, dentin, and soft tissue with hemostasis,[51] the device should operate with a wavelength that has high absorption in hydroxyapatite, collagen, and water.[55] The clinical laser system should operate at a 9.3-μm wavelength to maximize the absorption in hydroxyapatite, collagen, and water and maximize the ablation of all tissue types.
2. For fine detail cutting of enamel the fluence should be about 20 J/cm^2 with a spot size of less than 0.3 mm.[37, 44] The clinical laser system should have the flexibility to operate with pulse energies from 500 μJ to above 10 mJ with focused spot sizes of roughly 0.25 mm at 100 to 10,000 Hz.

3. To ensure pulpal safety, the pulse widths of the laser must approach the relaxation time of both hard and soft tissue.[30, 56] The clinical laser system should have the flexibility to operate with pulse widths for hard tissue at 30 µs or less and pulse widths for soft tissue at 500 µs or less.
4. For interproximal cutting the device must have a long working distance while maintaining an ablation fluence level above 10 J/cm^2.[45] With a focused spot size of roughly 0.25 mm, the clinical laser system should operate with fluence above the hard tissue ablation level at 10 mm either side of the focused spot position.
5. To finely remove or "shave" enamel the device must operate with a very shallow absorption depth.[57] With very tightly focused spot sizes of 0.25 mm or less, the clinical laser system must operate with variable energy levels of 250 µJ to 20 mJ with repetition rates of 100 to 10,000 Hz.
6. For a fine surface finish on enamel the 9.3-to-9.6-µm wavelength has to be used with a very shallow absorption depth. Pulse energies of 5 mJ to 25 mJ/pulse with repetition rates over 300 Hertz are used.[51] For clean margins and flat bottom surfaces, the clinical laser system should integrate computer-controlled patterns that link the beam movement to the computer's laser pulse repetition rate generation.
7. Caries tissue is composed primarily of destructed hydroxyapatite and water. To remove carious hard tissue, the clinical laser system must be highly absorbed by both hydroxyapatite and water.[58] The clinical laser system must operate with tightly focused spots at 9.3 to 9.6 µm with long pulse widths out to 500 µs.
8. To ensure proper X and Y movement with a highly focused laser spot, computer control of the focused spot position is required.[44] The clinical laser system requires a closed-loop feedback system, like galvanometers, that control the laser beam position on the focus optic to the central laser system control box and customer graphical user interface.
9. To ensure rapid ablation without pulpal heating an integral water spray system is required.[44, 59] Because the required laser wavelength is required to be absorbed in water in order to achieve char-free cutting of soft and hard tissue, the clinical laser system should have integral fine misting that avoids the laser beam path as much as physically possible.

CLINICAL DEVELOPMENTS AND USES

At this time there is only one 9.3-µm CO_2 laser system cleared by the U.S. Food and Drug Administration (FDA) for use in dentistry. The system (Solea, Convergent Dental, Natick, Mass., USA) was introduced for live patient use in August of 2013 (Figure 3). This device received FDA clearance for ablation of hard tissue for caries removal and cavity preparation, and for incision, excision, vaporization, coagulation, and hemostasis of soft tissue in the oral cavity.

Figure 3: *Solea 9.3-µm CO_2 Laser System*

This laser uses a modified CO_2 laser supplied by Coherent, Inc. of Santa Clara, California to produce the desired 9.3-µm wavelength. The laser is modified to produce the 9.3-µm wavelength by using an isotopic gas as discussed above. A second laser at 532 nm is incorporated into the system to provide an aiming beam since 9.3 µm is outside of the visible spectrum. The two beams are combined on an optical plate mounted above the lasers and delivered via an articulated arm and handpiece. The complete optical chain utilizes a total of 12 mirrors and 4 lenses.

Unlike other clinical dental lasers, this device uses extensive computer technology to manage the delivery of the beam, making it more manageable for the user and optimizing performance in the mouth. The computer is used to drive galvanometers, or galvos, small motors located in the handpiece. The last two mirrors outside the patient's mouth in the optical chain are mounted on the galvos perpendicular to each other. The galvos can oscillate up to 10,000 times in a second, manipulating the aiming and cutting beams into circular patterns of various sizes. The computer-driven galvos play a key role in providing a choice of spot sizes to suit the user and situation, and distributing the energy over a large area with more speed and uniformity than is possible with the human hand. This allows the tissue more time to cool as the beam is moved from one point to the next in the pattern, resulting in smoother margins and bottoms in hard tissue and more precise incisions in soft tissue. This laser's computer additionally controls the system's laser patterns depending on the variable-speed foot pedal's input, similar in form and function to that of the traditional dental drill. As the user increases pressure on the foot pedal, the computer increases the speed with which the selected pattern is executed. At maximum speed the laser will fire in bursts of up to 10,000 pulses per second. When the time required to jump from one point in a pattern to the next is factored in, the maximum average repetition rate is 2,200 pulses per second. By contrast, erbium lasers typically use frequencies of 50 pulses per second or less. In the mouth, dentists report utilizing the full range on the foot pedal from single digit up to the maximum repetition rate mentioned above.

The combination of easily controlled pulse durations, variable-speed foot pedal, and adjustable spot sizes are changing the way lasers are used. The dentist can easily adjust pulse durations from 1 to 500 µs and repetition rate from 1 to 10,000 Hz with the variable-speed foot pedal to effectively and quickly change settings or power (J/cm^2). This permits the operator to start procedures at lower power levels, facilitating a level of laser-induced analgesia for the patient prior to increasing the speed of ablation of enamel, dentin, and caries or excision/incision of soft tissue without having to stop and adjust settings on the laser. These are critical factors in achieving pain-free treatment without the need for local anesthesia. In children this is a major benefit because it eliminates the potential for injury associated with the child biting his or her lip or tongue before the numbness from the anesthesia dissipates. All these characteristics make using lasers more user-friendly and efficient when treating patients (Figures 4-7)

Figure 4: Case #1: Occlusal Caries Removal

Figure 4a: Preoperative view

Figure 4b: Mid-preparation view

Figure 4c: Mid-preparation view, continued, prior to caries disclosure

Figure 4d: Completed restoration

In case #1 (Figure 4), a 26-year-old female presented with occlusal caries in tooth #3, confirmed with a tactile explorer stick and a reading of 28 from a caries detection device (DIAGNOdent, KaVo Dental, Charlotte, N.C., USA). Neither injected nor topical anesthesia was administered. The 9.3-µm CO_2 laser-only (Solea, Convergent Dental, Natick, Mass., USA) preparation of the tooth proceeded with a dentin setting, 0.5-1.0 mm spot size, 5-10 mm tip-to-tissue distance, 30-100 µs pulse duration, 100% water, variable foot pedal speed of 30-90%. The tooth was then restored with a dual-curing self-etch bonding agent (Futurabond® DC, VOCO, Cuxhaven, Germany) and a bulk fill composite restorative material (SonicFill, Kerr, Orange, Calif.). The complete procedure, from seating of the patient to dismissal, took approximately 10 minutes. The patient reported no sensitivity to the extent that she was unsure whether the procedure was being performed on a maxillary or mandibular tooth.

Figure 5: Case #2: Distal Caries Removal

Figure 5a: Preoperative view

Figure 5b: Mid-preparation view

Figure 5c: Interproximal matrix band placed prior to restoration

Figure 5d: Completed restoration

In case #2 (Figure 5), a 23-year-old male patient presented with distal caries in tooth #20. No anesthesia (injection or topical) was administered. The laser-only preparation of the tooth was performed with a dentin setting, 0.25-1.0 mm spot size, 5-10 mm tip-to-tissue distance, 30-100 µs pulse duration, 100% water, variable foot pedal speed of 30-90%. An interproximal matrix band (Palodent, DENTSPLY Caulk, Milford, Del., USA) was placed and the tooth was then restored with a dual-curing self-etch bonding agent (Futurabond® DC) and a bulk fill composite restorative material (SonicFill). The complete procedure lasted approximately 10 minutes. The patient reported initial sensitivity but then was able to tolerate a lower pulse duration (80 µs). After the procedure, the patient requested that this laser be used for future restorations.

Figure 6: Case #3: Lingual Frenectomy

Figure 6a: Preoperative view

Figure 6b: Immediate postoperative view, no bleeding evident

Figure 6c: One-week postoperative view showing uneventful healing

In case #3 (Figure 6), a 4-1/2-year-old child was referred by a physician for a lingual frenectomy to help correct speech difficulties. No local anesthesia was administered; instead, topical EMLA (eutectic mixture of local anesthetics) cream (lidocaine 2.5% and prilocaine 2.5%) was applied, followed by 1-minute exposure to the laser's low-level 532-nm aiming beam to enhance the analgesic effect of the cream. The laser-only surgical procedure was then performed with a soft tissue setting, 0.25 mm spot size, 25-40 mm tip-to-tissue distance, 77 µs pulse duration, no water, variable foot pedal speed of 1-40%. The duration of laser exposure for the surgical procedure was approximately 25 seconds. The patient reported no discomfort either during or after the procedure. Afterward, the patient demonstrated greater tongue mobility and improved articulation, and could immediately proceed with speech therapy as needed. The parents were instructed to elevate the lip twice daily, apply vitamin E to the surgical site to promote healing, and to avoid spicy foods for several days. The one-week postoperative examination revealed uneventful healing, and the parents and child were pleased with the results. No discomfort was reported during the healing interval.

Figure 7: Case #4: Maxillary and Lingual Frenectomy

Figure 7a: Preoperative view of lip frenum

Figure 7b: Preoperative view of lingual frenum

Figure 7c: Immediate postoperative maxillary view, no bleeding

Figure 7d: Immediate postoperative lingual view, no bleeding

In case #4 (Figure 7), a 1-month-old infant was referred because of breastfeeding difficulties. Clinical examination indicated that both the lingual and maxillary lip attachments were interfering with the ability of the baby to achieve a good latch. Both mother and infant were experiencing difficulties with breastfeeding. No anesthesia (injection or topical) was administered. The laser-only surgical procedure was performed with a soft tissue setting, 0.25 mm spot size, 25-40 mm tip-to-tissue distance, 77 µs pulse duration, no water, variable foot pedal speed of 1-40%. The duration of laser exposure for the surgical procedure was approximately 25 seconds for each site. No bleeding was evident upon completion. The parent was instructed to elevate the tongue and lip twice daily. At one-week follow-up, uneventful healing was underway, and the mother reported that the infant was able to breastfeed immediately after surgery without difficulties, and all presurgery problems were resolved.

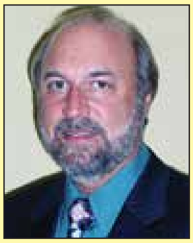
CONCLUSION

Practitioner and patient experience with the 9.3-µm CO_2 laser confirms the ability of this unique wavelength to safely, effectively, quickly, and comfortably perform caries removal, cavity preparation, and intraoral soft tissue surgery with reduced need for conventional drills to complete the procedures. Compared to other CO_2 laser wavelengths, the isotopic 9.3-µm wavelength combines high absorption and low reflectance coefficients and shallow absorption depth in hydroxyapatite to make it the ideal wavelength for efficient ablation of hard dental tissue. Effective soft tissue incisions and excellent hemostasis are enabled by this wavelength's high absorption into water and collagen.

The current implementation of this new laser is based upon the findings of nearly three decades of scientific research. The pulse durations of this 9.3-µm dental laser are specifically designed to correlate with the thermal relaxation time of hard and soft tissue to help ensure pulpal safety while the simultaneous water misting cools the targeted hard tissue. The pulse energies and repetition rates along with variable spot sizes enable fine detail cutting of enamel. The computer-controlled, galvanometer-guided laser beam and variable-speed foot pedal provide the clinician with unprecedented levels of precision and control.

AUTHOR BIOGRAPHIES

Dr. David Fantarella has been a practicing general dentist for the past 15 years in Hamden and North Haven, Connecticut. He specializes in general, implant, cosmetic, and laser dentistry. He received his Bachelor of Science degree from Dartmouth College while majoring in physics. He was awarded the Barry Goldwater Award for Excellence in Science. Dr. Fantarella completed his graduate studies at the University of Connecticut (UConn) School of Dental Medicine and was invited to do his residency at the UConn Medical Center. Upon receiving his Doctor of Dental Medicine degree from UConn, he was awarded the American Association award for excellence in Oral and Maxillofacial Surgery, the National Society of Dental Practitioners award in Law and Ethics in Dentistry, and the American Academic Periodontology award for outstanding achievement in periodontics. Dr. Fantarella may be contacted by e-mail at dfantarella@me.com.



Dr. Larry Kotlow is a Board-certified Pediatric Dentist practicing in Albany, New York. He is a graduate of the State University of New York (SUNY) Buffalo Dental School and received his pediatric training at the Cincinnati, Ohio Children's Hospital. He has advanced proficiency certification in the use of Er:YAG lasers and standard proficiency certification in the use of the Nd:YAG and diode family of lasers from the Academy of Laser Dentistry. Dr. Kotlow has achieved Mastership status in the Academy of Laser Dentistry. He has written more than 30 papers on using lasers and pediatric dentistry and has contributed chapters in three textbooks on using lasers in pediatric dentistry. He has been using lasers since 2000. He has lectured worldwide in Australia, Israel, Taiwan, France, Italy, and Canada as well as in meetings of the Academy of Laser Dentistry (ALD) and the American Academy of Pediatric Dentistry. He is the co-recipient of the ALD's Leon Goldman Award for Clinical Excellence in Laser Dentistry. Dr. Kotlow may be contacted by e-mail at kiddsteeth@aol.com.

Disclosures: Dr. Fantarella is a consultant to Convergent Dental on the development of the Solea CO_2 laser. He is an investor in Convergent Dental and serves on their Clinical Advisory Board. He also demonstrates technology for Henry Schein, and is a former trainer for Biolase.

Dr. Kotlow is a consultant to Convergent Dental on the development of the Solea CO_2 laser. He is an investor in Convergent Dental and serves on their Clinical Advisory Board. He has provided Technology 4 Medicine with consultation services on the XLASE™ and the LightWalker® lasers.

REFERENCES

1. Maiman TH. Stimulated optical radiation in ruby. *Nature* 1960;187(4736):493-494.
2. Patel CKN. Continuous-wave action on vibrational-rotational transitions of CO_2. *Phys Rev A* 1964;136(5):1187-1193.
3. C. Kumar N. Patel. The carbon dioxide laser. In: Hecht J. *Laser pioneers,* revised edition. Boston: Academic Press, Inc. 1992:191-204.
4. Glotz M. CO_2 lasers. Chapter 1-2.4.4 in: Berlien H-P, Müller GJ, editors. *Applied laser medicine.* Berlin: Springer-Verlag, 2003:33-36.
5. Migliore LR. Industrial lasers. In: Driggers RG, editor. *Encyclopedia of optical engineering,* Vol. 1. New York: Marcel Dekker, Inc., 2003:880-888.
6. Atkinson TJ. Fundamentals of the carbon dioxide laser. Chapter 5 in: Catone GA, Alling CC III, [editors]. *Laser applications in oral and maxillofacial surgery.* Philadelphia: W.B. Saunders Company, 1997:73-96.
7. Joffe SN. Lasers in medicine. In: Driggers RG, editor. *Encyclopedia of optical engineering,* Vol. 2. New York: Marcel Dekker, Inc., 2003:1045-1056.
8. Polanyi TG. Foreword. In: Shapshay SM, editor. *Endoscopic laser surgery handbook*. New York: Marcel Dekker, Inc., 1987:iii-ix.
9. Yahr WZ, Strully KT: Blood vessel anastomosis by laser and other biomedical applications. *J Assoc Adv Med Instrum* 1966;1:28-31.
10. Polanyi TG, Bredemeier HC, Davis TW Jr. A CO_2 laser for surgical research. *Med Biol Eng* 1970;8(6):541-548.
11. Strong MS, Jako GJ. Laser surgery in the larynx: Early clinical experience with continuous CO_2 laser. *Ann Otol Rhinol Laryngol* 1972;81(6):791-798.
12. Shafir R, Slutzki S, Bornstein LA. Excision of buccal hemangioma by carbon dioxide laser beam. *Oral Surg Oral Med Oral Pathol* 1977;4(3):347-350.
13. Adrian JC. Effects of carbon dioxide laser radiation on oral soft tissues: An initial report. *Mil Med* 1979;144(2):83-89.
14. Strong MS, Vaughan CW, Jako GJ, Polanyi T. Transoral resection of cancer of the oral cavity: The role of the CO_2 laser. *Otolaryngol Clin North Am* 1979;12(1):207-218.
15. Tuffin JR, Carruth JAS. The carbon dioxide surgical laser. *Br Dent J* 1980;149(9):255-258.
16. Horch HH, Gerlach KL. CO_2 laser treatment of oral dysplastic precancerous lesions: A preliminary report. *Lasers Surg Med* 1982;2(2):179-185.
17. Pecaro BC, Garehime WJ. The CO_2 laser in oral and maxillofacial surgery. *J Oral Maxillofac Surg* 1983;41(11):725-728.
18. Frame JW, Das Gupta AR, Dalton GA, Rhys Evans PH. Use of the carbon dioxide laser in the management of premalignant lesions of the oral mucosa. *J Laryngol Otol* 1984;98(12):1251-1260.
19. Frame JW. Carbon dioxide laser surgery for benign oral lesions. *Br Dent J* 1985;158(4):125-128.
20. Frame JW. Removal of oral soft tissue pathology with the CO_2 laser. *J Oral Maxillofac Surg* 1985;43(11):850-855.
21. Horch HH, Gerlach KL, Schaefer HE. CO_2 laser surgery of oral premalignant lesions. *Int J Oral Maxillofac Surg* 1986;15(1):19-24.
22. Pick RM, Pecaro BC. Use of the CO_2 laser in soft tissue dental surgery. *Lasers Surg Med* 1987;7(2):207-213.
23. Pogrel MA. The carbon dioxide laser in soft tissue preprosthetic surgery. *J Prosthet Dent* 1989;61(2):203-208.
24. Kamami YV. Outpatient treatment of sleep apnea syndrome with CO_2 laser, LAUP: Laser-assisted UPPP results on 46 patients. *J Clin Laser Med Surg* 1994;12(4):215-219.
25. Kverno D, Nolen J. The CO_2 laser. Molecular structure: Vibration and rotation. Davidson College Web site. http://www.phy.davidson.edu/stuhome/jimn/co2/pages/co2theory.htm. Accessed January 13, 2014.
26. Isotopes of oxygen. Wikipedia Web site. http://en.wikipedia.org/wiki/Isotopes_of_oxygen. Updated January 3, 2014. Accessed January 13, 2014.
27. Wilder-Smith P, Dang J, Kuroski T, Neev J. The influence of laser parameter configurations at 9.3 μm on incisional and collateral effects in soft tissue. *Oral Surg Oral Med Oral Pathol Oral Radiol Endod* 1997;84(1):22-27.
28. Nelson DG, Shariati M, Glena R, Shields CP, Featherstone JD. Effect of pulsed low energy infrared laser irradiation on artificial caries-like lesion formation. *Caries Res* 1986;20(4):289-299.
29. Featherstone JDB, Nelson DGA. Laser effects on dental hard tissues. *Adv Dent Res* 1987;1(1):21-26.

30. Featherstone JDB, Fried D. Fundamental interactions of lasers with dental hard tissues. *Med Laser Appl* 2001;16(3):181-194.

31. Vogel A, Venugopalan V. Mechanisms of pulse laser ablation of biological tissues. *Chem Rev* 2003;103(2):577-644.

32. Forrer M, Frenz M, Romano V, Altermatt HJ, Weber HP, Silenok A, Istomyn M, Konov VI. Bone-ablation mechanism using CO_2 lasers of different pulse duration and wavelength. *Appl Phys B* 1993;56(2):104-112.

33. Fried D, Borzillary SF, McCormack SM, Glena RE, Featherstone JDB, Seka W. The thermal effects on CO_2 laser irradiated dental enamel at 9.3, 9.6, 10.3, and 10.6 µm. In: Anderson RR, editor. *Laser surgery: Advanced characterization, therapeutics, and systems IV,* January 22-25, 1994, Los Angeles, Calif. Proc. SPIE 2128. Bellingham, Wash.: SPIE – The International Society for Optical Engineering, 1994:319-328.

34. Wigdor HA, Walsh JT Jr, Featherstone JDB, Visuri SR, Fried D, Waldvogel JL. Lasers in dentistry. *Lasers Surg Med* 1995;16(2):103-133.

35. McCormack SM, Fried D, Featherstone JDB, Glena RE, Seka W. Scanning electron microscope observations of CO_2 laser effects on dental enamel. *J Dent Res* 1995;74(10):1702-1708.

36. Kimura Y, Takahashi-Sakai K, Wilder-Smith P, Krasieva TB, Liaw L-HL, Matsumoto K. Morphological study of the effects of CO_2 laser emitted at 9.3 µm on human dentin. *J Clin Laser Med Surg* 2000;18(4):197-202.

37. Fan K, Bell P, Fried D. Rapid and conservative ablation and modification of enamel, dentin, and alveolar bone using a high repetition rate transverse excited atmospheric pressure CO_2 laser operating at $\lambda = 9.3$ µm. *J Biomed Opt* 2006;11(6):064008-1 – 064008-11.

38. Darling CL, Fried D. Real-time near IR (1310 nm) imaging of CO_2 laser ablation of enamel. *Opt Express* 2008;16(4):2685-2693.

39. Maung LH, Lee C, Fried D. Near-IR imaging of thermal changes in enamel during laser ablation. In: Rechmann P, Fried D, editors. *Lasers in dentistry XVI,* January 24-25, 2010, San Francisco, Calif. Proc. SPIE 7549. Bellingham, Wash.: SPIE – The International Society for Optical Engineering, 2010:754902-1 – 754902-6.

40. Takahashi K, Kimura Y, Matsumoto K. Morphological and atomic analytical changes after CO_2 laser irradiation emitted at 9.3 µm on human dental hard tissues. *J Clin Laser Med Surg* 1998;16(3):167-173.

41. Konishi N, Fried D, Staninec M, Featherstone JDB. Artificial caries removal and inhibition of artificial secondary caries by pulsed CO_2 laser irradiation. *Am J Dent* 1999;12(5):213-216.

42. Tao Y-C, Fried D. Near-infrared image-guided laser ablation of dental decay. *J Biomed Opt* 2009;14(5):054045-1 – 054045-6.

43. Hedayatollahnajafi S, Staninec M, Watanabe L, Lee C, Fried D. Dentin bond strength after ablation using a CO_2 laser operating at high pulse repetition rates. In: Rechmann P, Fried D, editors. *Lasers in dentistry XV,* January 24, 2009, San Jose, Calif. Proc. SPIE 7162. Bellingham, Wash.: SPIE – The International Society for Optical Engineering, 2009:71620F-1 – 71620F-9.

44. Nguyen D, Chang K, Hedayatollahnajafi S, Staninec M, Chan K, Lee R, Fried D. High-speed scanning ablation of dental hard tissues with a $\lambda = 9.3$ µm CO_2 laser: Adhesion, mechanical strength, heat accumulation, and peripheral thermal damage. *J Biomed Opt* 2011;16(7):071410-1 – 071410-9.

45. Fried D, Seka W, Glena RE, Featherstone JDB. Thermal response of hard dental tissues to 9- through 11-µm CO_2-laser irradiation. *Opt Eng* 1996;35(7):1976-1984.

46. Wigdor H, Walsh JT Jr, Mostofi R. The effect of the CO_2 laser (9.6 µm) on the dental pulp in humans. In: Featherstone JDB, Rechmann P, Fried D, editors. *Lasers in dentistry VI,* January 23-24, 2000, San Jose, Calif. Proc. SPIE 3910. Bellingham, Wash.: SPIE – The International Society for Optical Engineering, 2000:158-163.

47. Wigdor HA, Walsh JT Jr. Histologic analysis of the effect on dental pulp of a 9.6-µm CO_2 laser. *Lasers Surg Med* 2002;30(4):261-266.

48. Nair PNR, Baltensperger M, Luder HU, Eyrich GKH. Observations on pulpal response to carbon dioxide laser drilling of dentine in healthy human third molars. *Lasers Med Sci* 2005;19(4):240-247.

49. Bell PW, Fan K, Jones RS, Fried D. Analysis of peripheral thermal damage during the rapid ablation of dentin and bone using a λ= 9.3-µm TEA CO_2 laser. In: Rechmann P, Fried D, editors. *Lasers in dentistry XII,* January 22, 2006, San Jose, Calif. Proc. SPIE 6137. Bellingham, Wash.: SPIE – The International Society for Optical Engineering, 2006:61370G-1 – 61370G8.

50. Assa S, Meyer S, Fried D. Ablation of dental hard tissues with a microsecond pulsed carbon dioxide laser operating at 9.3-µm with an integrated scanner. In: Rechmann P, Fried D, editors. *Lasers in dentistry XIV,* January 20, 2008, San Jose, Calif. Proc. SPIE 6843. Bellingham, Wash.: SPIE – The International Society for Optical Engineering, 2008:684308-1 – 684308-7.

51. Staninec M, Darling CL, Goodis HE, Pierre D, Cox DP, Fan K, Larson M, Parisi R, Hsu D, Manesh SK, Ho C, Hosseini M, Fried D. Pulpal effects of enamel ablation with a microsecond pulsed λ = 9.3-µm CO_2 laser. *Lasers Surg Med* 2009;41(4):256-263.

52. Wilder-Smith P, Arrastia A-MA, Liaw L-H, Berns M. Incision properties and thermal effects of three CO_2 lasers in soft tissue. *Oral Surg Oral Med Oral Pathol Oral Radiol Endod* 1995;79(6):685-691.

53. Wilder-Smith P, Dang J, Kurosaki T. Investigating the range of surgical effects on soft tissue produced by a carbon dioxide laser. *J Am Dent Assoc* 1997;128(5):583-588.

54. Payne BP, Nishioka NS, Mikic BB, Venugopalan V. Comparison of pulsed CO_2 laser ablation at 10.6 µm and 9.5 µm. *Lasers Surg Med* 1998;23(1):1-6.

55. Fried D. IR laser ablation of dental enamel. In: Featherstone JDB, Rechmann P, Fried , editors. *Lasers in dentistry VI,* January 23-24, 2000, San Jose, Calif. Proc. SPIE 3910. Bellingham, Wash.: SPIE – The International Society for Optical Engineering, 2000:136-148.

56. Fried D, Zuerlein MJ, Le CQ, Featherstone JDB. Thermal and chemical modification of dentin by 9-11-µm CO_2 laser pulses of 5-100-µs duration. *Lasers Surg Med* 2002;31(4):275-282.

57. Zuerlein MJ, Fried D, Featherstone JDB. Modeling the modification depth of carbon dioxide laser-treated dental enamel. *Lasers Surg Med* 1999;25(4):335-347.

58. Fried D. Laser processing of dental hard tissues. In: Fieret J, Herman PR, Okada T, Arnold CB, Bachmann FG, Hoving W, Washio K, Lu Y, Geohegan DB, Träger, Dubowski JJ, editors. *Photon processing of microelectronics and photonics IV,* January 24, 2005, San Jose, Calif. Proc. SPIE 5713. Bellingham, Wash.: SPIE – The International Society for Optical Engineering, 2005:259-269.

59. Fried D, Ashouri N, Breunig T, Shori R. Mechanism of water augmentation during IR laser ablation of dental enamel. *Lasers Surg Med* 2002;31(3):186-193.