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Argon, Krypton, and Copper Lasers

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Abstract

The treatment of vascular lesions was among the first uses for lasers, and the argon, krypton, and copper vapor lasers were chosen for this task due to their relative specificity for the hemoglobin absorption spectrum. Despite their relative specificity, the primary mechanism of these lasers was photothermal, which causes thermal injury and coagulation. By leveraging thermal injury to coagulate vascular lesions, the laser surgeon's skill in handling a laser and patient selection was critical for ensuring excellent therapeutic results and avoiding potentially significant side effects, such as scarring. Despite the risks of scarring, aesthetic outcomes could be quite good – especially when compared to other contemporary modalities, such as surgical excision. However, once the theory of selective photothermolysis was elucidated, use of these lasers diminished as safer methods of treating vascular lesions were developed.

One of the first uses identified for lasers was treating vascular lesions. The earliest lasers applied were the: argon (488, 514 nm), argon-pumped tunable dye (488–638 nm), copper vapor and copper bromide (511, 578 nm), KTP (potassium-titanyl-phosphate; 532 nm), and krypton (568 nm) lasers. Although designed to correspond with the hemoglobin absorption spectrum, the continuous or quasi-continuous wave beams functioned by inducing thermal damage to the surrounding structures, which often caused scarring and dyschromia. The key problem, a mismatch between laser pulse duration and target tissue thermal relaxation time, was effectively resolved once the principles of selective thermolysis were elucidated and the laser pulse duration was tuned to match the thermal relaxation time of the targeted tissue.

Laser Thermal Effects

While the argon, krypton, and copper lasers were chosen for their relative specificity in excitation of hemoglobin, thereby efficiently transferring the energy of the laser to the target chromophore [1], their primary mechanism of action was delivery of thermal energy to the lesion.

The thermal effect of lasers is mediated by the absorption of laser light energy and represents the primary method by which the clinical effects of the laser are achieved and results in conversion of the absorbed energy into heat. While the thermal mechanism of lasers has been reviewed earlier, it is important to highlight how the photothermal effects of photocoagulation occur, as this is of primary importance in vascular lesions.

The extent of thermal injury with a laser is proportional to the magnitude and duration of a temperature increase because heat increases the rate at which molecules denature. This effect is demonstrated by the Arrhenius model mathematically, with the rate of denaturation exponentially related to temperature [2]. Cells respond to heat injury by altering protein levels, upregulating expression of heat shock proteins, which confer mild resistance to thermal injury [3]. One example of the clinical response to mild heat injury is erythema ab igne, which involves exposure to heat below 45°C (113°F) and the appearance of reticulated erythema and hyperpigmentation.

When tissue is heated with lethal doses of energy to above 60°C (the melting transition point of fibrillar type I collagen), coagulation occurs with DNA and protein denaturation, resulting in loss of function via unfolding and coagulation of macromolecules. Thermal coagulation yields cell necrosis and, if widespread, a burn. Studies have shown that the accumulation of denatured material rises exponentially with temperature and proportionally with time [4].

It appears that once tissue is heated to a critical temperature, coagulation occurs. On histology, this is visible as a well-defined area of coagulation necrosis. In the dermis, type I collagen plays a key role in thermal coagulation because of its thermal instability and induction of scarring when heated above 60–70°C. Elastin, on the other hand, is quite thermally stable and does not appear to play a major role. Here, the mechanism of scarring in laser therapy becomes clear. In cases where excessive transfer of heat from the intended chromophore (e.g. hemoglobin) to the surrounding tissue and dermal collagen occur, scarring can occur. The goal then of the laser surgeon is to select a laser that delivers enough energy, converted into heat, to ablate the chromophore without transferring substantial heat to collateral structures [5].

When treating vascular lesions with argon, copper vapor, and krypton lasers, laser surgeons often used visible lesion blanching as the endpoint of acceptable thermal damage, and by extension, successful treatment. While this blanching was thought be analogous to the coagulation of protein in an egg white while cooking, it was proposed that blanching was also caused by necrosis of melanin in the epidermis, resulting in less transmission of light to the dermis [6]. However, histological studies have shown

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that with some treatment protocols there is no epidermal or perivascular dermal damage [7]. Thus, loss of epidermal viability or coagulation of collagen are not the only processes occurring in the skin when blanching is observed. Rapid sequence photography has identified a transient (2–100 ms) blanching phenomenon, representing rapid emptying of the vascular lumen following centripetal expansion of endothelial cells that block the lumen [8]. This has shifted evaluation to using this transient blanching as the endpoint.

Lasers

The argon, krypton, and copper vapor lasers were among the earliest lasers employed to treat cutaneous lesions. Composed of a gas medium that is excited by a source of pump energy to generate light, these lasers emit a continuous beam or pulses of light. As continuous lasers, they usually have limited power, unlike more contemporary lasers, whose extremely high peak power can be emitted during a short laser pulse [9]. Of note, some of these lasers emit a relatively rapid (50–200 ms) train of low-energy pulses that behaves surgically like continuous-wave lasers and were called quasi-continuous. The pulse duration of these lasers, however, were still too long to observe any additional benefit [10].

Argon

The argon laser was first used in 1965 for the treatment of diabetic retinopathy and first sold commercially in 1971 [11]. The argon laser was applied in dermatology to vascular lesions, such as port-wine stains (PWS), capillary/cavernous hemangiomas, and telangiectasias, which benefit from its coagulation effect.

Argon lasers (e.g. Cooper Lasersonics Model 770 or Coherent Radiation Dermatologic Model 100) produce a visible blue-green light with six different wavelengths ranging from 457.9 to 514.5 nm, though two peak wavelengths at 488 and 514.5 nm account for over 80% of the laser energy output. This blue-green light corresponds to a relative peak of absorption by hemoglobin and melanin. When low levels of this blue-green light interact with highly pigmented (melanin or hemoglobin-containing) tissue, the result is a sufficient level of localized heat generation which makes the argon laser a highly effective coagulator by producing thermal damage of superficial blood vessels and (via conduction) coagulative necrosis of the epidermis and subadjacent dermis through the entire exposed field. This principle of selective absorption was the theoretical basis of photocoagulation of vascular lesions, such as PWS and telangiectasias. The argon beam can pass through the overlying skin without substantial absorption and reach the chromphore within the lesion, causing protein coagulation in these layers. However, the theory behind argon laser-induced damage fell short in practice, as more indiscriminate thermal damage was observed in practice than originally hypothesized.

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Equipment

The advent of fiber optics to transmit light made it possible to use the argon laser through flexible scopes. The beam could be transmitted through fibers that were attached to a handpiece that included a lens system. When used with a microscope, the micromanipulator included an internal shutter that protected the user from laser light when the laser was in use. Without a microscope, however, special glasses were required to protect the eyes from the laser light. Together, this self-contained and adjustable unit allowed for the use of this laser in dermatology.

Technique and Settings

The spot size used with argon lasers varied from 0.2 to 5.0 mm. A smaller spot size was used when more precise laser injury was desirable for single or smaller lesions (e.g. telangiectasias), while larger spot sizes were employed to cover greater surface areas when treating PWS. When the argon beam is focused on a very small spot (or total power is increased sufficiently), the power density can be high enough to vaporize tissue. The power ranges (1.0 and 2.5 W) used were lower for lighter lesions and higher for darker lesions; however, units with a power range as high as 10 W were also available. As a general rule, the minimum power necessary for clinical blanching or vaporization was employed. This was typically insufficient to produce scars. The pulse duration most frequently used was 0.2 s, but increased to 20 s of continuous exposure time for greater concentration of laser power in hypertrophic, spongy, and thick hemangiomas. The choice of continuous versus pulsed mode often depended on the laser surgeon's preference. The intermittent or pulsed technique allowed for a time setting (e.g. 0.2 s), spot size (e.g. 2 mm), and power setting (0.6–0.8 W). In either case, the laser stylus was held approximately 2–4 cm from the skin surface to ensure perpendicular delivery of the laser light. Then, a spot test could be performed, observing for whitening of the skin surface. Graying or blackening of the skin was an indication of excessive irradiance. A lack of blanching indicated a powder density that was too low. Observing for these effects, the laser surgeon would gradually adjust upward or downward from the initial level with successive spots until the appropriate clinical endpoint was achieved. With the power level set, the interval would then be set (e.g. 0.5–1.0 s). Typically, the fluence was 30 J/cm^2.

In the continuous setting, the same power settings were applied with the practitioner looking for the same whitening skin changes. Once this was attained, the setting was switched from intermittent to continuous and the laser source was waved over the skin (more rapidly with continuous mode, slower for pulsed mode). The speed of moving the laser over the skin was determined by simultaneously observed blanching with PWS and hemangiomas or vaporization with tattoos, moving faster if the spot appeared to get larger and slower if it appeared to get smaller. Total exposure typically averaged 100–125 J/cm^2 for the treatment area.

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Large Lesions

When treating larger lesions, it helped to divide the lesion into treatment segments of 2–3 square inches [2], typically corresponding to the natural anatomic areas of the face. While potentially easier for the laser surgeon to perform the task of treating the vascular lesion, treatment in segmental or intermittent patterns did not prove to reduce posttreatment complications substantially. In studies of large series of patients, the rate of complications (scars, hypopigmentation, and texture changes) for those treated in alternating patterns of rows and stripes (the stripe treatment method) was very similar to the rate for patients treated in solid block areas [1]. However, the possibility of a residual pattern of stripes and the unusual or bizarre appearance of the patient while undergoing treatment were disadvantages of the stripe treatment method.

Histology

The histology of argon laser treatment indicated significant epidermal damage with diffuse dermal collagen coagulative necrosis [12, 13]. The depth in the dermis depended on the incident energy density. Destruction of hair follicles, sweat glands, and sebaceous glands was also seen. Because of this damage, hypertrophic scars could occasionally occur.

Side Effects

Considering the hazards and complications of argon therapy, scarring was the most disconcerting occurrence (6% hypertrophic scarring under the best conditions) [14]. Hyperpigmentation and hypopigmentation usually resolved in 8–12 months, though in Caucasians with darker skin this could be a noticeable problem. Darker skin types (types 4–6) tolerated the hypopigmentation poorly. Pain was minimal compared to thermal burns of a similar extent. Time lost from work might be 2–5 weeks and infection was sometimes a problem, given the laser-denuded skin surface.

Scarring was a dreaded complication. Despite selective absorption by hemoglobin in vessels, it was evident that sufficient absorption of the epidermis and upper layers of the dermis did occur to produce thermal injury extending into the papillary dermis [15, 16]. In these unfortunate cases, the patient healed with scarring. Ultimately, favorable or unfavorable outcomes depended on a range of factors including the patient's healing tendency, presence of infection, occurrence of trauma to the healing wound (e.g. picking of crusts), presence of motion or traction on the healing wound, and presence of foreign bodies (e.g. dust, cosmetics). On a purely histologic basis, all patients treated with argon lasers had scarring. The practical implication of this fact was whether this scarring was objectionable or more prominent than the previous vascular lesion. Studies of patient perception indicated that 86% believed that they had scarring, defined as any area of the lesion that was less satisfactory in appearance than the patient had expected, and as an area that was elevated or indented, darker or lighter in color, rougher or smoother, or in any way noticeable [17]. In only 6% of these patients was the recognizable area ultimately considered to be true scarring and objectionable upon a physician's evaluation.

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Proposed methods of limiting the argon laser's thermal effects on tissue included reducing the tissue temperature, decreasing the laser power output, shortening the pulse duration, and limiting the number of impulses or exposure time to a given target. Gilchrest et al. [18] studied 30 consecutive patients with PWS to determine whether chilling lesional skin at the time of treatment could improve the outcome of argon laser therapy. On the basis of an approximate thermal relaxation ('cooling') time of 1 ms for the average ectatic PWS vessel, 1 ms was estimated as the optimal exposure for inducing specific thermal damage. During the 1970s and 1980s, pulse durations this short were not attainable on lasers available for clinical use given very high power outputs required of the laser machines in order to deliver sufficient energy to coagulate vascular structures. Arndt [19] showed that even shortening the pulse duration to 50 ms (still well above the thermal relaxation time) was not enough to achieve recognizable improvement in patient outcome. With this older technology, the continuous beam delivered with a 200-ms pulse duration was deemed equally good or superior to a 50-ms pulse duration because it was less tiring for the laser surgeon, allowed larger areas to be treated within a given amount of time, and often produced less mottling and more uniform results.

In addition to altering technique, prevention and patient selection were the important means by which physicians ensured excellent outcomes, avoiding those who would be predisposed to keloids (using test patches or clinical exam for this purpose) and using the selection criteria mentioned earlier, which were developed after years of experience. The areas of highest risk for scarring included, in order of incidence, the upper lip, melolabial fold, medial canthus of the eye, and mandibular and preauricular areas. Scarring could be minimized by decreasing fluence, using only the minimal amount to produce blanching or using the striping method described above.

Scarring could be seen as early as 3 weeks and as late as 3 months, and the most common complaint among patients was tightness of the skin that later developed into an enlarging nodule. If scarring developed, traditional methods of inhibiting scar formation could be employed, such as intralesional triamcinolone injections or a steroid-impregnated tape cut in the pattern of the scar and changed every 12 h. For smaller, less prominent scars, steroid creams could be given to massage into the lesion. Pressure supports worked well on extremities (e.g. wrist, hand, ankle, etc.).

Despite the best efforts of physicists and engineers to tune the argon laser to its medicosurgical purpose, the ultimate benefit in outcome relied on the operator to employ craftsman-like hand-eye coordination and small hand maneuvers to expertly treat a lesion. For example, by overlapping spots to intensify effects, leveraging the diffusion of heat from the spot edge to more gently treat a sensitive area, and matching striping and continuous techniques for delicate areas on the face, operators applied an understanding of the laser-skin interaction to achieve aesthetically superior results.

This reliance on the 'art' of the operator reflected the application of an imperfect tool to the problem of vascular lesions. Ultimately, the argon laser proved to be an intermediate step in the progress of vascular laser development. Although tuned to the absorption spectra of hemoglobin and vessels, its use at pulse durations far exceed-

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ing the relaxation time of the targeted structures resulted in development of second-degree burns that are now considered excessive and unnecessary since the elucidation of the theory of selective thermolysis.

Copper Vapor

Although the argon laser was the standard of care for treating vascular lesions and PWS for over 20 years, the incidence of side-effects and greater understanding of the theory of selective thermolysis encouraged the research and development of new lasers capable of treating these lesions.

Among the newer laser systems introduced for treatment of cutaneous vascular lesions was the copper vapor laser, a metal vapor laser whose laser media was a combination of vaporized copper and neon gas [20]. Copper, in the form of metal pieces, was placed into the laser discharge tube, where it was heated to 1,500°C. Thirty to 45 min was needed to produce a vapor and represented the warm-up time of the laser apparatus [21]. Neon gas was added to improve the discharge quality of the laser. Two wavelengths were emitted by this system, and one could change from yellow (578 nm) to green (511 nm) light by switching the filters. The yellow light, in particular, offered several potential advantages over the argon laser. The 578-nm wavelength approximated the peak absorption of oxyhemoglobin at 577 nm and, more importantly, corresponded to a greater relative absorption by hemoglobin compared with melanin. Furthermore, the greater wavelength allowed for deeper tissue penetration than would be expected at argon wavelengths [22]. Initial published data also suggested that the copper vapor laser compared favorably with the argon laser in the treatment of vascular lesions.

The copper vapor laser also represented a quasi-continuous laser, producing sequential 20- to 50-ns pulses of light at a rate of approximately 16 kHz. However, the peak pulse power was higher than with argon lasers. In this way, the copper vapor laser differed from the dye lasers, which were either continuous wave or single pulsed. It was common for many laser surgeons to filter out any green light for treatment of vascular lesions and use only the yellow. The spot size typically ranged from 100 to 1,000 μm [21].

Equipment

The light from a copper vapor laser was typically delivered via a 1 mm in diameter quartz optical fiber or a robotic scanner. In the former, the end of this fiber protruded from a pen-shaped handpiece which had been ground flat and roughly polished. The latter came as both a large built-in computerized scanner or a smaller automated handpiece [21]. Light delivery was controlled by a pneumatic foot switch.

Technique and Settings

Prior to starting treatment, the periphery of the area to be treated was marked with a green pen which contrasted with the color of the lesion. The green outline provided a finishing point for the scan because the true edge of a vascular lesion was not always easily discernible when rapidly scanning the tissue. Unlike the argon laser, the handpiece

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