Review

State-of-the-art: Dental photocuring—A review

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ABSTRACT

Light curing in dentistry has truly revolutionized the practice of this art and science. With the exception bonding to tooth structure, there is perhaps no single advancement that has promoted the ease, efficiency, productivity, and success of performing dentistry. Like most every major advancements in this profession, the technology underlying the successful application of light curing in dentistry did not arise from within the profession, but instead was the result of innovative adaptations in applying new advances to clinical treatment. One cannot appreciate the current status of dental photocuring without first appreciating the history and innovations of the science and industry underlying the advances from which it developed. This review will place the current status of the art within the context of its historical progression, enabling a better appreciation for the benefits and remaining issues that photocuring has brought us. Lastly, the manuscript will present thoughts for future considerations in the field, offering suggestions as to how current advances in light-generating science might yet be adapted for dental use.

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Contents

1. Introduction to the review .................................................................................................... 40  
2. Literature ...................................................................................................................... 40  
2.1. Abbreviated history of light curing in dentistry ............................................................. 40  
2.2. UV-curing .................................................................................................................. 40  
2.3. Visible light curing .................................................................................................... 40  
2.4. Quartz–tungsten–halogen lights .................................................................................. 41  
2.5. Vital tooth bleaching and introduction of the “alternative initiators” 41  
2.6. The argon-ion laser .................................................................................................... 41  
2.7. Plasma arc lights ....................................................................................................... 42  
2.8. High-intensity QTH lights ........................................................................................ 42  
2.9. Issues resulting from application of high light levels .................................................... 43  
2.10. Light emitting diodes .............................................................................................. 43  
2.11. First generation leD ............................................................................................... 44  
2.12. Second generation LED ........................................................................................ 44
1. Introduction to the review

There is hardly a clinical procedure performed in contemporary dentistry where photocuring of some material is not required. The process seems so simple, perhaps even trivial: just point and shoot. What could be so complicated? However, there is much more to this story than that. In order to appreciate the convenience and benefits today’s state-of-the-art offers, it is first necessary to understand and value how this technology developed over time. This manuscript will provide a short review of the history underlying development of photocuring in dentistry, focusing on issues related to construction and operation of the light sources themselves. The perspective for appreciating current technology will be established, the issues that we currently face will be addressed, and those we have yet to overcome. A summary of recent key findings is provided, and thoughts on important factors affecting the future of development in dental photopolymerization are developed.

2. Literature

2.1. Abbreviated history of light curing in dentistry

From the late 1940, with power/liquid, self-curing direct, acrylic-based restorative materials, to the late 1960s when the dual-paste, self-cure composite systems were popular [1,2], use of direct polymer-based restorative materials has changed the way dentistry is performed and perceived. The early paste–paste systems, although rivaling their predecessors with respect to appearance and durability, still left many clinicians wanting for a faster process of fabricating tooth-mimicking restorations, where the working time was almost infinite [3], the setting time was only seconds and not minutes, and where colors were stable and long-term wear allowed successful use in the posterior segment [4,5].

2.2. UV-curing

Like many advances in dentistry, the technology for using light to polymerize resin-based materials did not originate within the profession, but instead was an existing technology that was adapted for dental use [6-9]. The first photocuring units were designed to emit ultraviolet light (about 365 nm) through a quartz rod from a high pressure mercury source and were introduced in the early 1970s [10]. This development was seen as a revolutionary step in dentistry, for it allowed a “cure on demand” feature, which was previously unattainable using the self-curing products. Typical exposure durations were 20 s, but 60 s provided enhanced results [11]. Filled, photcurable composites as well as sealant materials were available [11,12], and were used in very innovative ways to not only restore carious processes, but to also repair tooth fractures and provide easily performed esthetic results [13,14]. The photoinitiating system relied on benzoine ether-type compounds [1,15], which broke down into multiple radicals, without need of an intermediary component [16]. The spectral distribution of light sources of the time with the absorption spectrum of the initiator help to correlate these two parameters [17]. Although some of the restorations placed using this early technology have proven to be remarkably successful [18], in general the procedure was fraught with many issues. First, because of the limited ability of light to penetrate deep within the material, incremental buildups were required instead of bulk placement, and were limited in depth [19,20]. In addition, concerns were voiced about the potential for harmful effects of the short wavelength energy being exposed to human eyes (corneal burns and cataract formation) as well as possible changes in the oral microflora [1,21].

2.3. Visible light curing

Amazingly, only a few years following the introduction of UV radiation for curing dental restoratives, the ability of using visible radiation was introduced: February 24, 1976. On that day, Dr. Mohammed Bassouiny of the Turner School of Dentistry, Manchester, placed the first visible light-cured composite...
restoration on Dr. John Yarn, the then head of development of this effort by Imperial Chemical Industries (ICI) of England [22]. Again, development of the chemistry for doing so did not arise from the dental field, but instead adapted from use in the field of polymerizing thin resin films for printing, metal and plastic coatings, and paints [23,24]. The optimization of a visible light-curing photoinitation system composition using camphorquinone and a tertiary amine co-initiator was key to the success of this effort, and remains the most popular basic formulation in use today [15,25–28]. The unit consisted of a quartz–tungsten–halogen (QTH) source having heat absorbing glass and a bandpass filter allowing only light between 400 and 550 nm to pass: the wavelengths [29,30] required to activate the photoinitiator, camphorquinone. ICI went on to partner with Johnson and Johnson to introduce the first light-cured resin composite and light: the FotoFil system [29,31]. The advantages of using visible light were that 2-mm thick increments could typically be placed using from 40 s to 60 s exposure from a QTH source, and that the potential formation of cataracts and oral microflora alteration were minimized [31]. However, direct retinal burning and advancement of macular degeneration were now a potential for ocular damage, as the wavelength needed to initiate the visible-light initiated systems fell directly within the frequencies known to cause immediate, and permanent damage resulting from retinal burning [32,33]. Thus, practitioners were advised to place a filtering film between their eyes and the curing light to preclude ocular damage, while also allowing high levels of longer wavelengths to pass in order to adequately visualize the treatment field while curing: the so-called “blue-blockers” [34,35].

2.4. Quartz–tungsten–halogen lights

The QTH light source was not developed for this purpose, but was, instead, advanced by engineers at General Electric for use in aircraft lights, where small but very bright and durable sources were needed [36]. However, the QTH light became the mainstay of dental light curing for many years, into the late 1990s. During that time advances included a wide range of adaptations for convenience and efficacy. Bulb wattage increased from 35 W up to 100 W for hand-held units [3], and up to 340 W for table-top models [3]. Output values ranged from an average of 400 to 500 mW/cm² up to an extreme of 3000 mW/cm² from one unit: The Swiss Master Light, Electro Medical Systems, Nyon, Switzerland. The filtered spectral emission matched the absorption profile of camphorquinone very well [17]. Units were either hand-held (gun style) using a light source within the gun, from which light was emitted to the target through a bundled glass fiber light guide. The gun was connected to a base power unit using a flexible cord containing electrical wires. The adaptation of a curing gun to accept a wide variety of sizes and styles of different light guides fitting with similar focal points lead to the customization of specific types of light delivery to oral locations [37]. The other basic model was a table top unit, which contained a high-Wattage source directing filtered radiation to a flexible glass bundle cord [38], or a liquid light guide. The QTH sources in the hand-held units lasted from 30 to 50 h [30], and utilized the halogen cycle to remain clear from tungsten contamination on the inner wall of the quartz envelope [30]. Sources were readily available, easy to install, and relatively inexpensive. Typical exposure duration times to provide uniform properties in a 2-mm thick composite increment ranged from 40 s to 60 s [39], and controls allowed selected exposure intervals [40]. Units had to be fan-cooled in order to extend the working time of the source and allow the halogen cycle to operate correctly.

2.5. Vital tooth bleaching and introduction of the “alternative initiators”

Vital tooth bleaching was becoming an overnight success in the early 1990s. However, following this type treatment, manufacturers were not able to provide direct esthetic restorative materials that were of high enough value to match the newly bleached teeth. This situation arose because the photoinitiating system of that time only used camphorquinone, which is a bright canary yellow, and photobleaches only slightly upon exposures within clinically relevant times [16,41]. Thus, after curing, the restoration tended to have a yellow, residual tinge. In order to provide restorative materials of high value, manufacturers resorted to utilizing other photoinitiators that were used for the UV coating and printing industries, which had a small portion of their absorption range within the short, visible light spectrum [42]. These compounds were a class of initiator that directly broke into multiple radicals without need for any co-initiator, and, although they were pale yellow in color, photobleached to clear once utilized [43]. Examples of such compounds are bis(2,3,6-trimethylbenzoyl)-phenylphosphineoxide [15,44] (Ciba Specialty Chemicals, Inc., Basel, Switzerland), also known as Irgacure 819, and 2,4,6-trimethylbenzoyl-diphenylphosphate oxide [15,45] (BASF Corporation, Charlotte, NC), commonly referred to as Lucerin® TPO. Later, an yellowish oily initiator (1-phenylpropane-1,2-dione, FPD), was introduced to help broaden the overall absorption of initiators between 400 nm and 500 nm [15]. When combined with CQ, a synergistic effect was found, allowing the concentration of CQ to be reduced, thus decreasing the yellow residual color following photocuring [46–48], however, others found no reduction in yellow [48]. Also, to reduce the time needed to obtain maximal bleaching effect, manufacturers of bleaching agents were advocating exposure of solutions to bright light in order to heat the components and hasten their breakdown into perox radicals. The wavelengths required for this process were short, near the violet and near ultraviolet, and as well as within the blue region. Thus, light curing units now serve a dual-fold function: to photobleach and to activate resin-based restorative materials, and provide an energy source to hasten vital tooth bleaching.

2.6. The argon-ion laser

As is typically the case, the use of the argon laser to photopolymerize acryl<sup>•</sup>s did not arise within the dental community, but was instead a product of industrial development in the coating industry. The argon-ion laser was marketed first to enhance the effects of vital tooth bleaching in Europe, and is still used for that purpose today [49]. However, when first introduced to the United States [50], only the clinician could operate the unit, and not auxiliary personnel [51], so the light became more profitable to use as a dental polymerization
source. The initial delivery of power from the laser to the tooth was directly through the end of a fiber optic cable. However, due to the divergent nature of that radiation, other methods were developed in an attempt to make a fairly collimated beam of coherent energy, whose target power was not related to the tip-to-tooth distance [52], as was the case with conventional QTH light guides. The devices proved very effective in providing high physical properties in dental restorative materials [53], and because of the unit’s tremendous radiant output, some noted that shorter exposure times were needed to provide similar properties as a QTH source [54], while some did not [55]. The size and footprint of the unit were very large at first [56], but a single source could be adapted so that one laser could feed multiple operators using a network of fiber optic cabling [56]. Over time, the unit became much smaller, and could easily fit into an operator, but because of weight, needed to be on a cart, if moved from room-to-room. However, owing to the high expense (near $5000) of a typical unit, the inability to replace the source by office personnel, and the elevated room temperatures resulting from the device use, this curing system became outdated in a short time. The emission profile of argon-ion lasers marketed in the United States eliminated the strongest output at 514 nm for some reason. This wavelength was typically used to provide hemostasis [52], and was present as a selectable feature on units marketed in Europe. Excluding the 514 nm peak, the major output from dental argon lasers in the US occurred at 488 nm, just on the sharply declining, long wavelength leg of camphorquinone absorption. Other minor emission peaks at 457.9 nm, 468 nm, and 476.5 nm do occur within high CQ absorption levels, and are thus effective in curing, but others emissions are not: 496 nm [52]. High speed pulsing of the light was found to provide enhanced surface and depth conversion over continuous exposure [57], and some researchers advocated that each brand and shade of composite required an individualized energy delivery for optimal performance [58]. Designs for a portable diode laser for dental light curing have been developed, but to date, no commercial product has arisen [59].

2.7. Plasma arc lights

Because dental auxiliaries were prohibited from using the argon laser in the US, technology was imported from Europe (French patent #2,305,092, 1976), that had already met with a great amount of success: the plasma-arc curing light. Developed in the mid 1960s [60], this type source was not initially developed for dental purposes, but instead was used to provide broad-banded radiation for visualization of operating fields (endoscopy and colonoscopy, for examples) as well as for minimally invasive medical procedures. The source consists of two tungsten electrodes separated by a small distance, encased in a high pressure gas-filled chamber, having a synthetic sapphire window through which the light emission was directed from a parabolic reflective surface [50]. A high electrical potential is developed between the two electrodes, which then forms a spark, ionizing the gas, and providing a conductive path (a plasma) between the electrodes. Once the initial spark is established, electronics then adjust the operating current to maintain light generation through a variety of sophisticated feedback system [30]. The gas originally used in dental PAC lights in Europe contained argon, and was extremely high in output, allowing claims of “sub-second” exposures in advertisements to be used to replace conventional 40–60 s exposures using the QTH light. These units had to be highly filtered, as they generated tremendous amounts of infrared light (resulting in heat generation at the target) as well as ultraviolet (having dangerous ozone formation potential). In dental applications, these types of light sources were pulsed and initially developed for curing UV-polymerizable resins, although no commercial unit could be found [19]. They were later adapted for use in visible-light cured products [61]. Thus, the table top unit contained the conventional heat-absorbing glass and visible bandpass filters, but the light emitted after this entered a three-foot-long liquid light guide [62], which acted as a very long optical filter, eliminating UV and IR radiation, and passing only visible light [30]. The typical output from these types of lights was near 2000 mW/cm², and was broad banded: from 380 to 500 nm, with a naturally occurring peak near 460 nm, where CQ has its optimal absorption. Originally designed in Europe as the Argo HP, claiming the ability to cure composite using “sub-second exposures”, the unit was attempted to be marketed in the US by DMDS, Inc, of Salt Lake City, UT, as the Apollo 9500. It is conjectured that the source in this unit contained argon, which allowed it to emit higher than conventional xenon, but did not last as long. However, issues with licensing led to the US version of the light introduced in August, 1998 as the Apollo 95e [51]. This unit was quite small, and touted provision of adequate exposure times of 3 s as being equivalent to that of a 40 or 60 s exposure from a QTH light. However, the reason why this unit provided no longer continuous exposure than 3 s did not arise from its inadequacy of providing sufficient energy to optimally cure a composite within that time, but instead was attributed to the power supply driving the source not containing sophisticated enough feedback to allow continuous output for any longer time. To perform adequately clinically, multiple 3-s exposures were typically required [63,64]. The unit offered a variety of removable curing tips at the end of the light guide: curing tips: 430 nm or 460 nm for resin curing and 400–500 nm for bleaching. It was from the confusion concerning tip selection for curing specific restorative materials that the confusion started related to the issue of what light will work with which composite or bonding resin. Subsequent PAC units overcame many of the initial shortcomings of the Apollo 95e: broad banded output (not requiring a special bandpass tip end), longer, continuous exposure times, and a general tendency to provide adequate composite curing using a single 10-s exposure [30].

2.8. High-intensity QTH lights

The QTH light was then competing with PAC units for market share. In order to respond to the new, high output device, manufacturers of QTH lights used different mechanisms to increase the overall output of their units, and claimed they were “equivalent” to a PAC unit with respect to performance. First, source settings were available to “boost” light output over normal values. This mechanism really just drove the filament at a higher voltage, exceeding accepted tolerance values than the bulb manufacturer suggested. Settings in such output modes were no longer than 10 s, as longer exposures would
severely degrade the operating lifetime of the source. Another mechanism of increasing output was the development of the “turbo-tip”. This device was a rigid glass fiber bundle, but had the individual fibers drawn while hot, so that the bundle diameter was smaller at the distal, emitting end than at the proximal, receiving end of the guide. By doing so, the same amount of power was present at both ends, but was distributed over a much smaller area at the emitting end, resulting in about a 1.6 times increase in irradiance [65]. This type tip is still widely used in contemporary LED units to help increase overall output values. However, even with both of these features in operation, the output of the QTH did not match that of a typical PAC light of the day [30]. The future for the QTH light is not “bright”, as many governmental agencies are banning this tremendously inefficient incandescent light sources from general usage soon. Thus US government has stipulated deadlines by which incandescent light sources will stopped being marketed: elimination of the 100-W bulb in 2012 and ending with removal of the 40-W source by 2014 [66].

2.9 Issues resulting from application of high light levels

During this time of application of ever increasing irradiance levels to teeth and gingiva, issues arose related to the application of excessive heat to the target. Clinicians were admonished to direct light output on their thumb nail to sense the level of “heat” evolved from a light. In addition, with such high intensity levels, the PAC lights were now noted to cause a high degree of polymerization stress, as the resin was polymerized so fast that relaxation was not allowed to occur in the polymer network before it became vitrified. To overcome this problem, researchers found that if the light intensity was delivered in such a manner as to control the rate of curing, stress relief would occur by allowing the resin to flow prior to vitrification. It was hoped that in so doing the potential for debonding between the restoration and the tooth would be much lower [67]. In addition, less heat would generate in the target during the restorative process [68]. However, little difference was noted in microleakage studies [69–71]. Thus arose the concept of “soft-start” polymerization.

When first introduced, this exposure method was added as a selectable feature on QTH lights, and resulted in an initial short (10 s) low-level output near 100 mW/cm², which was followed by an immediate jump to full output power for the remainder of the exposure: step curing [72]. Later changes resulted in a time-based increase of the light initially applied, with the remainder of exposure providing full output: ramp curing [73]. Still another modification was provided as the “pulse-delay” technique, where, for the last composite increment to be cured, a very low level, short duration exposure was given (3 s at 200 mW/cm²). The clinician was advised to treat another patient (for 5–10 min) while the composite was allowed to flow and relieve stress, after which a bolus of higher output power was provided to complete the cure (30 s at 500 mW/cm²) (BISCO VIP). However, the results did not produce any remarkable differences to what was being used conventionally [74].

An issue with use of this type exposure delivery is the adequacy of energy delivery at the bottom of an increment. Thus, the possible correlation between enhanced marginal adaptation, reduced shrinkage, and lowered curing stress using this curing protocol might have been related to an overall lower extent of cure at the bottom surface [75,76]. As long as an adequate exposure is provided during the high-output phase of this technique, concerns about under polymerization are not an issue [77]. Variations of a soft-start feature are still offered on many contemporary light curing units, even LEDs. Comparison of material conversion between conventional 40 s exposure and that when using a ramped soft-start procedure did not indicate any significant different difference [73], however, contraction strain and the polymerization exotherm were decreased [73,78]. To remain competitive, manufacturers of PAC lights also attempted to utilize a soft-start mode to reduce the tremendously high stresses and temperature rises developed from their use. However, the operational characteristics of the PAC light do not provide for low radiant output, as the spark arising from even the minimal voltage needed to maintain its presence produces more irradiance than does a conventional QTH light using regular output. Thus, benefits of soft-start curing with these types of units could not be realized [79]. However, one manufacturer incorporated a modulated level of output during exposure in order to maintain an overall high irradiant output, but also to decrease target temperature [80].

2.10 Light emitting diodes

The 50th anniversary of the invention of the light emitting diode (LED) is fast approaching [81]. Use of this technology has truly changed our lives, and will continue to do so, as it proves to be an efficient, cost-effective lighting source. These solid state devices rely on the forward-biased energy difference (band gap) between two dissimilar semiconductor substrates (n-type conduction band, and p-type valence band), to determine the wavelength of emitted light [30]. They are much more efficient than previous types of dental light sources, are light weight, and can be easily battery powered for portability. During the middle-to-late 1990s, tremendous strides were made in the video display, optical communications and solid-state lighting industries with the introduction of light emitting diodes (LEDs). At that time, the red laser was used for providing the source of reading optically encoded video discs. Because the wavelength of red is longer than that of blue, the physical size of the hole through which the light had to pass in order to create a digital signal was also large. With development of shorter wavelength, blue LEDs, smaller sized holes were required than with red, and literally many more openings could be placed into the same physical area occupied by fewer and larger holes that transmitted red light. Thus, more digital information could be stored on smaller sized media, and development in LEDs within this wavelength range became a focus of interest [82].

Blue emissions were developed using indium–gallium–nitride (InGaN) substrates in the early 1990s [81]. This was also a color needed to activate phosphors to emit yellow, enabling the first white-appearing LED in history [81,83]. It was thus coincidental that the output range of the newly developed blue LEDs fell within the maximal absorption of
CQ, enabling the possibility of using blue LEDs as a dental light-curing source. As has been mentioned previously, it did not take long before the blue LED made its way into a dental curing light [30,84].

2.11. First generation LED

These lights were first introduced to the commercial market at the end of 2000 [85], with the introduction of the LUX®MAX light (Akeda Dental A/S, Lystrup, Denmark), however the concept was developed much earlier: 1995 [86,87]. A typical design of first generation LED curing lights used an assemblage of multiple, individual LED, single-emitting “5 mm lamp” LED elements (each chip providing 30–60 mW) into a focused, axial [88,89] or planar array [87], arranged such that the combined output of the luminaire was sufficient enough to provide sufficient energy to activate CQ. Arrays of from seven (Freelight, 3M/ESPE, St. Paul, MN) to 64 (GC-e-light, GC America, Alsip, IL) were available, but even with those numbers, the extent of radiation was not sufficient to warrant competition with respect to providing decreased exposure time over that of the standard QTH light. In addition, battery technology relied upon use of NiCad cells, which were plagued with poor performance and memory effects. However, because some photo-curable restorative products used only the short wavelength, alternative initiators, the new blue LED lights would not polymerize them, nor would the argon-ion laser. This situation started the great confusion over what LED lights would not polymerize them, nor would the argon-ion laser. This is related to the development of the technology: the “second generation” LED curing lights, developing what is referred to as the “second generation,” LED lights. Chip manufacturers were now fabricating chips specifically within the wavelength requirements for dentistry, and labeling them as “dental LED blue”. Luxeon is Philips Lumileds Lighting Company’s trade name for their high-power LEDs which dissipate 1 W or more. At first, there were two types of Luxeon chips used in second generation lights, according to the development of the technology: the 1 W chip (Luxeon LXHL-BRD1 or –MRD1 generating 140 mW output), and the 5 W chip (Luxeon LXHL-PRDS or –MRDS, generating 600 mW output). Characteristic of these units is the great increase in output power over that of the first generation [81], with one 5-W chip providing similar luminance as 10–20 of the individual 5-mm types of the first generation. However, a similar wavelength range output persisted as with the first generation, resulting in a continued inability to photocure restorative materials using only short wavelength initiators. Battery technology also improved, and NiMH energy packs became the typical power source. However, because of the generation of tremendously high power in such a small area, the temperature within the chip was now a concern, as too high a temperature would result in permanent chip damage, and this thermally cut off prior to that point [84]. Thus, units had heavy metal heat sinks or large metal surfaces to dissipate chip heat [81,92]. In addition, the return of fans [92] to curing lights was seen – something everyone thought was gone forever. Recently, a 10 W blue LED has become available especially for dental applications (LZ4-00DB10, LedEngin, Inc., Santa Clara, CA, as well as a 15 W unit (LZ4-00CB15, LedEngin, Inc.), and can provide up to 4.2 W and 5.6 W of radiant flux, respectively. With the dramatically increased chip output, second generation LED lights were now reliably able to outperform their commercial competitive curing light types using shorter exposure times [93,94].

2.13. Third generation LEDS

In order to enable polymerization of restorative materials utilizing more than only CQ as initiator, curing light manufacturers resorted to providing LED chip sets that emitted more than one wavelength [95]. The first of these units utilized a separate 5 W blue LED central chip surrounded by four low power violet (around 400 nm) LEDs: the Ultralume 5, Ultradent Products, South Jordan, UT. Thus, by combining the output from these two wavelengths, light was provided at wavelengths that were effective for not only CQ, but for the alternative set of initiators as well, creating the equivalent of a “broad banded LED” curing light. Other manufacturers incorporated the violet chips located next to other blue chips within a single LED element: LZ4-00D110, High Efficiency Dental Blue + UV LED Emitter, LedEngin, Inc. with 1 UV die emitting 0.76 W and 3 blue dies each emitting 3 W. This ability to generate multiple wavelengths from a single LED light led a “third generation” of LED dental curing lights. Typically, these units use either NiMH or Li-ion battery technology. One manufacturer currently incorporates three different wavelength chips: VALO, Ultradent Products. Units of this generation are quite effective in providing sufficient irradiance at appropriate wavelengths to be able to polymerize any type of dental restorative material.

2.14. LED construction styles

Contemporary LED lights are either those styled after the old QTH guns, with a turbo-tip fiber optic bundle, or are a pencil-type unit, having the emitting chip set at the distal end of the unit, or a similar fiber optic guide. The lights are either totally driven by mains connection, are battery operated, or both. Those with the chips at the distal end of the unit either have no coverage over the elements, a sealed or removable clear plate covering it, or utilize a fixed lens. Advantages of the pencil style body include the ability to gain access to many more locations in the mouth than possible with the hard, long
glass bundled guide, as well as less possibility of breakage, should the unit drop.

2.15. Summary of history

Even with all the advantages that solid state light curing has brought to dentistry, many of the issues correlated with other types of light sources (polymerization shrinkage and stress, intrapulpal temperature rise, and confusion over method of light intensity application) persist at present. Misunderstanding and frustration still exist over compatibilities between lights and restorative resins, as well as the advantages and disadvantages of using curing units with such tremendously high output levels [96–98]. Even with all the improvements made in photocuring over the last 40 years, after 10 years of service, only 43% of almost 100,000 resin restorations were still in service, with performance described as “comparable only to the worst performing amalgam restoration [99]”. Only with this background, can the most contemporary findings and issues can be better placed in context of the entire field.

3. Contemporary issues

3.1. Adequacy of exposure durations

There is still much confusion over the ability of a given light or a specific composite combination to be used together to provide optimal polymerization within a short clinical timeframe. Clinicians desire to perform quality dentistry using minimal chair time. Curing light manufacturers advertise use of their lights for a single duration of output, often regardless of composite type, shade, or clinical distance the tip is held from the restoration. Likewise, manufacturers of composite recommend specific exposure durations for their products that often do not match those of lights made by competitive companies [93,100–104]. Thus, the clinician is left wondering which suggested time is “correct”, and as a result, tend to over-expose restorations to be on the “safe side”. However, in so doing, the longer exposure results in generation of more heat within the tooth and surrounding, exposed tissues, leading to possible post-operative, iatrogenic complications [100,105,106]. Clinicians will often utilize a hand-held dental curing radiometer as an indicator of the potential a curing unit has for photocuring resin-based materials. However, these instruments have been shown to be inaccurate [107] and not good predictors of clinical performance [108], and are really only meant to be used as a relative indicator of light output value over time [108].

3.2. Custom, in-office exposure guide

Recently, a simple in-office test has been introduced whereby a clinician can develop a custom curing exposure guide, using their own light unit, any type of composite, and at any tip-to-composite distance using simple items readily available in a common office [109]. This method relies upon the clinician establishing the relationship between exposure duration, and tip distance on the thickness of cylinder of remaining, unscrapable composite made from exposing composite-filled, single-dose compules. The test also works for syringe material. The results from this procedure have been correlated with those of a more sophisticated, laboratory test when determining optimal exposure using biaxial flexural strength [109]. This method allows for the determination of optimal exposure time for any given combination of curing light and composite, thus avoiding the confusion of relying on manufacturer-recommendations for each. However, the test requires time as well as use of an existing stock of composite material. If the test is performed on a periodic basis, the clinician can also know exactly how to adjust exposure duration of a light, found to be decreasing in output, to provide optimally effective results once again. In addition, the relative performance of a given composite when using two or more different light units can be determined, as well as relating the performance of different composites when using a given curing light.

3.3. Photocuring training device and precision curing unit evaluation

A newer device, combining precise, laboratory spectral technology with clinically relevant measuring conditions within prepared dentiform teeth in a manikin head has been developed (MARC (Managing Accurate Resin Curing), BlueLight analytics inc., Halifax, Nova Scotia, Canada). This device uses cosine correctors to capture emitted light from curing units, and directs the output to a calibrated spectral radiometer embedded within the manikin head. The sensors can be placed anywhere in the dentiform, replicating a variety of different classifications of tooth preparations, locations, and preparation depths. Output from the radiometer is fed into a laptop computer, where custom software is used to provide a myriad of real-time and accumulated data: spectral irradiance, total energy delivered over a given exposure duration, and the estimated exposure duration needed to deliver a specified energy dosage. Training office personnel to properly perform light curing is very effective using a device such as this. The effects of minor alteration in tip position, movement, or location are instantly displayed in real-time, and the ultimate consequence in terms of altered energy delivered is determined. The unit can also be used in a “scientific mode”, whereby a calibrated, high resolution spectral irradiance plot is available for display or digital output to a spreadsheet device. The device has been successfully used to test literally hundreds of clinicians in a “before training” and “after training” mode, and has shown that, with effective education of the operator with respect to hand stabilization, tip angulation and position, and direct visualization the operating field through blue blocker glasses, use of the instrument provides the optimal result in energy delivery [110]. The device can also discriminate the ability of different lights to deliver adequate energy levels between various tooth locations using this simulated clinical model [111]. Accurate, recordable curing light performance over time is also capable of being determined using the same device, for without such determination, clinicians are totally with the knowledge of the condition of their photocuring unit [112].

The relevance of this type measurement relies on the fact that there is a critical energy level needed to optimally polymerize dental composites [113]. In addition, the estimation of
adequate energy delivery currently relies on the reciprocity between exposure duration and power density. Some literature indicated that such a relationship exists [114], while others show evidence that it does not [115,116], and that the polymer network developed using fast or slow delivery of energies is different [117]. In addition, the spectral delivery of energy within the absorptive needs of the photoinitiator system present in a given composite is also not accounted for [42,48,90,118,119]. However, the device is capable of such discriminative capacity with software improvement, and minor hardware changes should allow in situ testing of real-time energy delivery to actual resin composites while curing.

4. Current methods and findings in curing unit characterization

4.1. Hand held curing radiometers

With wide ranging claims of curing light performance touted by manufacturers, it becomes of great importance to be able to distinguish the true operating characteristics of curing lights so that valid evaluation of those claims can be made [120]. Unfortunately, many published articles rely on the validity devices called “hand-held curing radiometers” to provide accurate data from which performance among lights is correlated to some parameter measuring polymerization extent: depth of cure, flexural strength, hardness, etc. Great discrepancies among measurement of light unit output have been found using such hand-held dental curing radiometers, validating that they are not considered reliable indicators in ranking the potential for depths of cure among lights [108]. The hand held radiometers are really just a photometers calibrated in radiometric units [30,107]. A typical device consists of a detector port on which light from the emitting end of the unit is directly placed. A drawback of this type unit is that the detector must be smaller than the diameter of the tip in order for the meter to work correctly [120]. Thus, the unit is not exposed to the total emitted power, and is assuming that emissions are homogeneously distributed over the emitting face end. Secondly, there is no way to discriminate among the spectral emission of lights, as the unit only responds to all radiation passed to the photodiode through whatever restrictive bandpass filter it uses. This limitation means that not only is the distribution of power with respect to frequency is not known, but the unit may also not be responding to wavelengths outside of the bandpass which might be present. In addition, these type devices only provide an indication of exitance irradiance (the emitting tip-end value), which is not an indicator of light performance when held at a distance from the tooth [121,122]. Thus, these devices are really not meant to report accurate data to characterize light unit emission, but are instead designed to operate as a method to evaluate the periodic performance of a curing light over time for purposes of detecting output changes [107,123].

4.2. Conventional irradiance measurement

True characterization of the emissions from dental curing lights has been provided in the literature since the introduction of the devices [17]. With more current technology, these processes have become less expensive and provide higher resolution, and respond much faster than the older types of instruments [124]. A well equipped light measurement facility would first determine the total power output of the unit measured using a well calibrated thermopile. These instruments are black body absorbing plates with layers of thermocouples that respond to absorption of radiant energy in a linear manner over a very wide range of frequencies. For accurate measurement, all radiation from the exiting port must fall on an area larger than the beam. Because dental curing lights have a very wide difference with respect to beam diversion and increasing tip distance, this means that the tip end needs to be held as close to the detector plane as physically possible, without touching it [122]. Unfortunately, when following ISO 106500-1 testing standard for determining exitance radiation of curing lights, the tip ends are held at a distance from the detector plate [125]. Depending on the distance between the top plane of the thermopile to the depth where the black plate is located, the tip will be held at varying distances, and will thus provide inaccurate results for light power emission [126]. Therefore a wide diameter, shallow-welled thermopile is best for evaluation of dental curing lights. In addition, it is optimal to have at least two independent, calibrated thermopile systems so that the tolerance of calibration measurements can be checked. However, a thermopile will provide accurate indication of the emitted power, but without respect to its spectral distribution: measuring the radiant flux (mW). Knowing this value, the area of the emitting end over which light is being projected is measured accurately using a microscope and is divided into the total power to derive the irradiance or exitance irradiance (more casually referred to as “power density”): mW/cm².

4.3. Spectral irradiance

As mentioned previously, the thermopile does not discriminate with respect to frequency over which it measures power: it only measures total power. Spectroradiometers are used to evaluate dental curing radiometers and measure power generated over the visible spectrum. The spectral distribution of dental curing lights has been determined since UV units were introduced [17,21]. Early technology required large and very expensive equipment. However, advancements in the science have allowed very precise measurements to be accurately made using only a modest investment. These instruments are commonly referred to as “hand-held spectroradiometers” (model USB2000 + RAD, Ocean Optics, Dunedin, FL).

To analyze light emission, a device needs to be utilized that captures all radiant output from the emitting end. One type device is an integrating sphere, which consists of an entrance port into which the emitting end of a curing light is placed, and within which light reflects multiple times off of a highly reflective, diffusing film. A fiber optic cable is fitted to an exit port of the sphere, and connects to a small spectroradiometer. This device then separates the input light into its spectral components using a grating, and projects that beam onto a fixed, multiple element diode array. The more diodes in the array, the smaller the entrance slit size, the more finely the spectrum can be resolved. The array output is analyzed using software,
where the contents of each pixel captured over a specific time frame are determined and displayed on screen as a plot of emitted power (photon count) with respect to wavelength. These systems must be calibrated against a highly accurate light source that typically resides inside of the sphere, so that the sphere, connecting cables, and spectral radiometer are all calibrated as a single instrument. Any change in tightening or cabling will negate the calibration. This type system is bulky, as the sphere is typically 6 in in diameter, and requires a dedicated, computer-controlled power supply for the calibration lamp. A limitation of the device is that it cannot determine variation in power or wavelength values across the emitting tip, as the internal reflections average out all such differences, providing only a single, value. The output measure from using such equipment is the spectral radiance: mW/nm.

A simpler experimental system involves use of a cosine corrector (CC-3, Ocean Optics) connected to a fiber optic cable, which again is attached to the spectroradiometer. The cosine corrector is a lambertian diffuser (typically opaline glass, Spectralon®, or Teflon) that captures light from all incident angles within 180 degrees from the plane on its exposed side. Typically, the optical input diameter of the corrector is approximately 4 mm, so it does not capture all emitted power from a tip larger in size, and assumes a uniform distribution of both irradiance and wavelength across the curing unit tip face. However, by providing a known capture area, the resulting spectral distribution can be reported using irradiance units, not just power. The entire system (cosine corrector, and spectrometer, must also be calibrated using a traceable source that is capable of providing uniform light of known quantity across the cosine corrector face. However, once calibrated, the instrument is portable, as it consists only of a small cosine corrector, a fiber optic cable, and palm-sized spectral radiometer, and easily operates from a laptop computer. Data generated using this type setup provides the spectral irradiance of a light curing unit: mW/nm/cm².

### 4.4 Determining beam irradiance inhomogeneity

Beam irradiance mapping (intensity contour generation) is not new to the dental field, as large discrepancies were detected in the UV curing lights [21] and as early as 1985 for visible light models [127]. Recent work [124,128,129] has developed a method for observing the uniformity of light across the emitting end of contemporary light curing units. The instrumentation needed for such work is adapted from analysis of laser beams, where knowledge of how the laser interfaces with a surface is essential to its performance: if the beam power is higher in the core or higher in the periphery. The basic setup of such an instrument consists of a CCD camera upon which the beam is targeted. The system is configured differently for analyzing dental curing light beam uniformity, as there must be a translucent target present and a lensing/iris system to adjust for focusing and saturation of the detector. The distance between the camera and target is fixed so that accurate dimensional measurements can be made of the beam image at that plane. The target material is critical to obtaining accurate results, in that the material must be opaque enough to block imaging the tip through the target, but translucent enough to provide sufficient light to pass and make an image on the camera side. A target material that seems to perform quite well for tip-end imaging is 1500 grit ground glass (DG100X100-1500, Thorlabs, Newton, NJ). First, the total power emitted from the tip end is measured using a calibrated system: either a thermopile or integrating sphere. Next, the emitting end of the curing unit is placed in direct contact with the target glass. It is essential to obtain accurate distinction from a signal and background noise, so a system must be used that provides uniform baseline values for all pixels when no signal is present (UltraCal™, Ophir-Spiricon, Logan, UT). The camera-side image is then analyzed using software and displayed on the computer screen. The lens aperture must be adjusted so that the maximal power observed by the camera is set just under the saturation limit of the detector, providing full dynamic range to scale the result. The total measured power obtained using the thermopile is entered into software, and the computer distributes that value across the pixels within the beam image, and color renders different value ranges. By providing accurate dimensional information at the target plane, the result is a calibrated, color-coded image of the beam irradiance dispersion across the emitting face.

### 4.5 Importance of beam imaging to curing light characterization

Beam imaging is a necessary component for characterizing output from a curing light. Often, it is very difficult to determine the exact area over which light is being emitted from the unit end. This situation is especially true for LED curing lights that have chips located at the distal end of the handpiece, and utilize no lensing system, but instead, have the emitters only behind a clear plastic plate, or behind nothing at all. Unless the precise emission area is known, determination of valid irradiance values are not possible to calculate. When determining the irradiance of curing lights utilizing glass bundled fiber optic guides, many researchers measure the diameter of the bundle and assume that all exiting light is actively dispersed across the entire bundle end. Using the beam profiler, recent research has identified specific incidences where the active, lighted area of a fiber optic bundle is less than that of the actual bundle area itself [124]. Without such knowledge, large errors in irradiance values assigned to such a light would occur.

A measure of the uniformity of power distribution within the beam is called the “top hat factor”, or THF. In this calculation, software determines the weighted average of each pixel making up the beam image, and relates it to the maximum response seen. Thus, a number from unity (representing perfectly uniform distribution) to zero is presented, with lower values indicating less uniformity [124].

The extremely important aspect of such type analysis is that a histogram of the irradiance distribution as a function of pixel values occupying the area on the beam face can be calculated [124]. When such procedures are performed, it is seen that many lights display very wide ranges of irradiance values, while others seem to be more narrow-ranged [124]. The critical point being made is that conventional methods for irradiance determination assign a single, average value (using an integrating sphere or cosine corrector) to a curing light. However, what should be reported is the histogram of irradiance.
distribution, as it is not normally distributed and truly indicates the range of output values the unit presents. Providing such information gives a much better comprehension of the quality of a unit than does presentation of a single, averaged value. In addition, the effect of irradiance distribution when different types of light guides are used on the same unit body can be identified and quantified [124]. The clinical relevance of such differences in irradiance is pointed out by the fact that localized areas of irradiance value deviation at the tip end have been precisely matched to the mapping of hardness values on the surface photocured by that unit, as well as at a depth of 2 mm [129]. These differences could affect the rate of localized curing, and of the resulting polymerization stress development, which could also affect the integrity of the resin-tooth interfacial bond. In addition, the differences could affect localized conversion values and wear resistance, leading to a differential in restoration surface wear over time [130].

4.6. Nonuniformity of wavelength distribution within the beam

The beam analysis system can also be used to measure the homogeneity of wavelength dispersion across the beam tip [131]. For units having broad banded sources (QTH and PAC lights), there is no difference, as all portions of light in the beam contain all output wavelengths. However, in the third generation LEDs, the off axis arrangement of the different color chips is reproduced in the forward beam image, resulting in localized areas of radiation occupied by only one chip’s wavelength: very little frequency mixing within the beam [130]. Such a condition may result in disproportionate utilization of resin systems containing multiple photoinitiators, each requiring different wavelengths for activation, not only at the top, irradiated surface, but also within the depths of the composite. Again, conventional methods of curing light analysis would have not detected such differences, however with these new analytical techniques, clinically relevant characteristics are detected and quantified, and valid distinctions among light units can be made.

4.7. Curing tip movement

Knowing that discrepancies in both irradiance and wavelength may exist in a light beam obviates the reaction to move the beam during exposure in order to better distribute values. Large discrepancies were noted in the beam profiles for the early UV light [21], and as a result, the clinician was advised to move the tip during exposure to purposefully average out localized differences and hopefully create a more uniformly polymerized restoration [1]. Later work was done in this area using large diameter and smaller diameter QTH light guides [132]. It was found that circular movements, as opposed to fixed, or overlapping exposures resulted in a much poorer distribution of composite hardness, and that large and small diameter light guides provided the best results when used in a rigidly fixed position. Similar, but contemporary work performed using LED lights also confirmed that tip movement (either fast or slow circular or rastering motion) resulted in lower overall hardness at both the top, irradiated surface, as well as at a 2 mm depth [133]. However, in that study, all exposures lasted the same amount of time: 20 s. It was contemplated that, if longer exposure duration had been applied when moving the tip, hardness values would have been greater better due to the increased energy applied to each location.

5. Future developments

With respect to curing light characterization, the methods outlined above should be considered for incorporation into national as well as international standards, as existing standards and evaluation protocols are obviously deficient. The physical properties of resins cured with various lights have been found to be inferior, although all units passed the ISO standard for measuring depth of cure, conforming that the standard method is not truly discriminating in characterizing curing light performance [134]. In the near future, curing light manufacturers, and the clinical offices using light emitting devices will more than likely be held responsible for monitoring the radiant output from such units as a result of regulatory agency requirements. Such requirements are already being adopted in Europe for nearly all classes of products that emit any sort of electromagnetic radiation [135,136]. However, because of the well documented retinal damage of light within the spectral range currently used for photoinitiation causes, curing lights may be helped to an even higher level of scrutiny. Therefore, clinical offices may have to either pay for professional, certified services to periodically provide characterization of their lights (as they currently are required to do for their X-ray machines), or purchase a qualified device that has been certified and worthy of such measurements. With the alarmingly poor performance and maintenance noted from multiple studies assessing curing light status in private practice [137,138], accountability of the clinician to provide optimal polymerization performance should, and hopefully will, be required.

Future studies investigating material properties of resins resulting from photopolymerization by dental curing lights should provide better documentation and description related to the true output characteristics of the lights used in the study so that better correlation can be made between resulting properties and exposure to radiation. The research community must revisit current standardized methods of determining performance of restorative resins (such as determining depths of cure), and devise more appropriate testing that accounts for discrepancies in localized irradiance and wavelength discrepancy. Currently, only a 4-mm diameter composite specimen is used for this purpose. Clearly, any great off-axis distribution of either irradiance or applied wavelength may significantly alter the performance of a light, relative to one where the distributions have resulted in a high concentration of light in the center of the tip. To this end, it would be the burden of manufacturers to develop lights having more uniform output characteristics. Internationally accepted standards for allowing specified amounts of deviance in beam characteristics could be developed, permitting more valid comparison of resin specimens photocured with such units.

With world energy sources dwindling and the need for energy efficient use of power, the application of light emit-
ting diodes in medicine and dentistry will continue to grow. Already, use of light for diagnostic and therapeutic purposes has become wide spread. However, again, as the LED industry makes advances in light-generating capabilities, the dental field will typically be the first to incorporate any new technology into clinical practice. Of specific interest is the use of organic light emitting diodes (OLEDs). These products allow extremely thin and flexible video displays to be made, but the current technology is such that output levels remain much lower than those of conventional LED chips. However, it is not unreasonable to envision impression trays with walls and floors lined with these emitting films, designed to evenly irradiate all surfaces of a photo-curable impression material. In addition, use of quadrant type trays for the simultaneous and wide area illumination of teeth for photocuring of sealants or light-enhanced vital bleaching or veneer bonding would be a great advantage, as would be the application of thin, adaptable films for photocuring resins retaining orthodontic brackets.

Another area where tremendous advances have been made with incorporation of state-of-the-art lighting technology in medical therapy has been the field of quantum dots. These substances are semiconductors with conducting characteristics closely related to the size and shape of their individual crystals. Smaller crystals display a larger band gap. Thus, as the difference in energy between the highest valence band and lowest conducting band increases, more energy is needed to excite the dot, but also, more energy is released when the crystal is back in its resting state. Such technology allows fluorescence to occur at shorter wavelengths than those of excitation. This condition would allow red light exposure to result in emission of blue light, which might be used for photoinitiation. If such dots were incorporated into a photopolymerizable resin composite, it might be possible to enable the entire mass to release light within itself, resulting in a “curing from within”, which is an aspect always dreamed of, but never realized for dental applications.

6. In summary

It is truly a remarkable time in dentistry. So many recent technical advancements have become clinical realities and are now necessary tools for the contemporary practice of dentistry, it is difficult to imagine how techniques could get much more automated. However, the beauty of our profession is that, despite the complexity of instruments and tools used, the dentist is the still the health care provider who spends the most time in direct contact with a non-sedated patient. Thus, dentists tend to build loyal patient bases, because the practice of the profession remains a one-on-one, individualized treatment, and not a remote, robotic, impersonal procedure. This is truly a great time for dentistry.

REFERENCES


