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Shining a light on high volume photocurable materials

William M Palin[a*], Julian G Leprince[b,c] and Mohammed A Hadis[a].

[a] College of Medical and Dental Sciences, Institute of Clinical Studies, University of Birmingham, UK
[b] School of Dental Medicine and Stomatology, Université catholique de Louvain, Belgium
[c]Advanced Drug Delivery and Biomaterials (ADDB), Louvain Drug Research Institute (LDRI), Université catholique de Louvain, Brussels, Belgium

*corresponding author:

William M Palin, Professor of Biomedical Materials Science
University of Birmingham
College of Medical and Dental Sciences
Institute of Clinical Sciences
Birmingham Dental School & Hospital
5 Mill Pool Way
Edgbaston
Birmingham B5 7EG
+44 121 4665547
+44 121 4665542

Abstract

Spatial and temporal control is a key advantage for placement and rapid setting of light-activated resin composites. Conventionally, placement of multiple thin layers (<2 mm) reduces the effect of light attenuation through highly filled and pigmented materials to increase polymerisation at the base of the restoration. However, and although light curing greater than 2 mm thick layers is not an entirely new phenomenon, the desire amongst dental practitioners for even more rapid processing in deep cavities has led to the growing acceptance of so-called “bulk fill” (4-6 mm thick) resin composites that are irradiated for 10-20 s in daily clinical practice. The change in light transmission and attenuation during photopolymerisation are complex and related to path length, absorption properties of the photoinitiator and pigment, optical properties of the resin and filler and filler morphology. Understanding how light is transmitted through depth is therefore critical for ensuring optimal material properties at the base of thick increments. This article will briefly highlight the advent of current commercial materials that rationalise bulk filling techniques in dentistry, the relationship between light transmission and polymerisation and how optimal curing depths might be achieved.

Contents

1.0 Introduction

2.0 Curing light transport through depth

- 2.1 Light transmittance, irradiation time and degree of conversion of dental resin composites
- 2.2 Light irradiance, depth and photoinitiation related to Beer-Lamberts Law
  - 2.2.1 Molar absorptivity
  - 2.2.2 Photoinitiator concentration
  - 2.2.3 Competitive light absorbers
  - 2.2.4 Light refraction and scattering

3.0 The era of “bulk-fill”

4.0 Emerging and potential future technologies for deep photocuring

5.0 Conclusions

1.0 Introduction

For many decades the development of materials science, polymer- and photochemistry, photophysics and optical engineering has allowed a variety of industries to improve curing strategies for materials processing. Photopolymer systems have been developed for applications that require highly sensitive materials and processes, the most notable of which include photoresists for microelectronics and surface patterning, printing inks and coatings, 3D-imaging, and, of course, medical uses, including dentistry.

The ability to harness light energy for crosslinking polymer materials provides significant advantages in terms of spatial and temporal control of the setting reaction. In many applications such as photoresists and coatings, the photopolymer is crosslinked as a superficial surface layer, often in micrometer scale film thicknesses, and for most of these uses free radical photoinitiators with wavelength sensitivities in the ultraviolet (UV) spectrum are most often utilized due to the higher energy of UV photons. Indeed, the first types of light curable dental resin composites (c.1970) were formulated with UV photoinitiators such as benzoin methyl ether with an absorption maxima at ~365 nm and polymerized using a high-pressure mercury arc light source (~340-380 nm) channeled through a quartz rod. The recommended irradiation times of 60 s produced a limited curing depth of ~1.5 mm (or 750 µm by modern ISO standards), most likely due to the low curing irradiance (50x less than a typical modern LED curing unit; Figure 1), high molar absorptivity associated with Type I photoinitiators and greater scatter of shorter wavelengths. Therefore, the use of UV-curable dental resin composites was relatively short lived and soon after superceded by materials that contained dye-based, Type II photoinitiators allowing photopolymerization by lower energy photons of visible (blue) light that are also considered less damaging to biological cells and tissues. However, and of noteworthy mention, even during the early use of dental light curing, meticulous clinical techniques (rubber dam isolation, proper use of matrix bands, incremental placement, accurate light curing protocols, thorough finishing and polishing procedures) provided exceptional clinical success: 76% of

four different UV-cured posterior resin composites considered clinically acceptable after 17-year recall [1]. Also, a 30-year follow-up randomized clinical trial led to similar conclusions, with an overall success rate of 66.7% at 30 years, which corresponded to an annual failure rate of 1.1% [2]. The latter work included three materials: a one-paste light-cured hybrid resin-composite used with the usual 2 mm layering technique, and two chemical-cured two-pastes hybrid resin-composites.

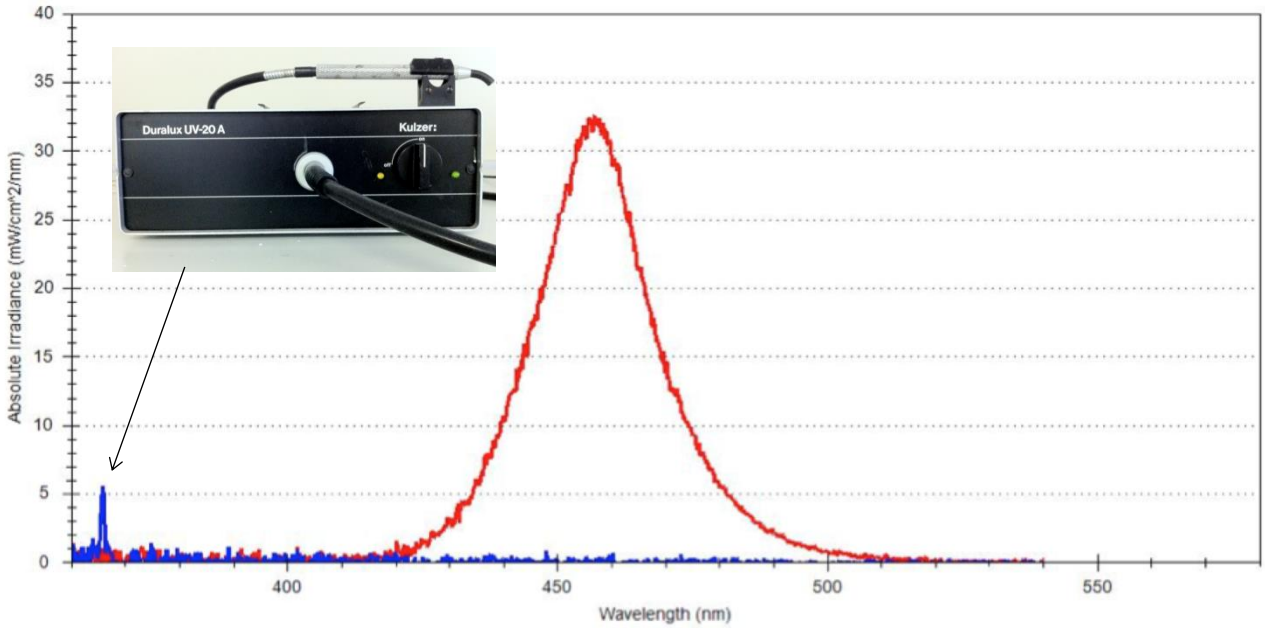


Figure 1: Spectral irradiance of a traditional mercury arc dental curing lamp (lambda max. 365 nm) compared with a modern single diode LED curing unit (lambda max. 455 nm). Absolute irradiance ~20 and 1000 mW/cm^2, respectively

Forty years on and modern clinical studies that report similar longevity of light activated resin composite materials might be considered an outstanding success. However, longitudinal studies in controlled clinical trial settings conducted by highly skilled practitioners might not reflect the situation in general dental practice, and certainly without the pressures of time and percentage profit return in private clinical businesses. The research, development, manufacture and marketing of dental resin composite technology is therefore not always based on significant steps towards material property development, but directed more towards convenience of the dentist and improved technique, reducing the tedium of multiple placement steps, shortening suggested irradiation times and ultimately attempting to save chairside time.

Examples of technological developments that attempt to improve placement procedures of dental resin composites and reduce chairside time have included self-etch adhesives that eliminate the enamel and dentine conditioning stage, higher power light curing regimes that incorrectly assume equivalent material properties using a concomitant reduction in irradiation time, and most recently, so-called, "bulk-fill" resin composites that lessen the requirement for multiple incremental placement in large cavities. These examples do not represent an exhaustive list of developments, but signify the modern appetite of dental practitioners for time-saving clinical procedures, the latter of which is fueling a shift in modern dental resin composite manufacture. It is therefore timely to review the current light curable resin composite materials that rationalize bulk filling techniques in dentistry, and this article will consider how light is transmitted through resin composite materials, the factors that affect light transport and polymerization, and how optimal curing depths might be achieved using modern and future technologies.

2.0 Curing light transport through depth

2.1 Light transmittance, irradiation time and degree of conversion of dental resin composites.

The changes in light transmission throughout polymerization of a light curable resin composite are complex and significantly affected by the constituents and chemistry of the material. The measured light irradiance, that is, the radiant flux *received by a surface* per unit area (W/m^2), which is a critical parameter that determines the extent of cure, is often inappropriately or inaccurately measured, or not recorded at all. Although methods of light irradiance and transmission measurements are not the main focus of this article (related works can be found elsewhere [3], briefly, irradiance is commonly reported at the tip of the curing unit, although proper measurements should be taken at the surface of the specimen. Even then, the average measured irradiance is affected by the distribution of light across the face of the curing tip (most curing lights do not exhibit a high "top-hat" factor, or homogeneous irradiance distribution) and the type (spectral sensitivity) and size of the sensor area (smaller diameters will overestimate irradiance of beam profiles with Gaussian distribution) [4,5]. Notwithstanding the possible effects of varied methods of light measurement, if absolute irradiance is measured at the lower

surface of a thick photocurable resin composite, the extent of light loss through a 5 mm thickness of 'bulk fill' (BF) resin composites is substantial (79.2 – 89.0%; Table 1).

			Irradiance / mW.cm^{-2}						
Mould Thickness (mm)	Measurement time (s)	CR 1	CR 2	BF 1	BF 2	BF 3	BF 4	***high power LED (without material)***
1	1	353 (24)^{b}	430 (39)^{a}	508 (6.4)^{b}	363 (24)^{b}	452 (16)^{a}	378 (42)^{a}	*2162 (7.6)^{a}*
	10	429 (32)^{ab}	475 (57)^{a}	558 (15)^{a}	432 (19)^{a}	465 (3.6)^{a}	388 (41)^{a}	
	60	496 (46)^{a}	490 (48)^{a}	578 (9.2)^{a}	465 (30)^{a}	479 (12)^{a}	405 (37)^{a}	
3	1	34 (3.0)^{c}	81 (1.9)^{c}	192 (8.1)^{b}	91 (9.1)^{b}	146 (24)^{a}	98 (14)^{a}	*1848 (2.0)^{b}*
	10	53 (5.7)^{b}	99 (1.0)^{b}	238 (9.1)^{a}	123 (6.3)^{a}	178 (22)^{a}	116 (17)^{a}	
	60	112 (2.6)^{a}	114 (2.4)^{a}	258 (12)^{a}	143 (9.4)^{a}	198 (27)^{a}	133 (16)^{a}	
4	1	12 (1.0)^{b}	29 (1.0)^{c}	107 (4.1)^{c}	52 (1.8)^{c}	83 (9.2)^{c}	68 (8.8)^{b}	*1468 (1.6)^{c}*
	10	18 (1.0)^{b}	47 (3.1)^{b}	161 (2.7)^{b}	73 (3.2)^{b}	114 (5.6)^{b}	77 (9.7)^{ab}	
	60	54 (8.4)^{a}	60 (1.8)^{a}	183 (4.8)^{a}	93 (8.4)^{a}	136 (6.7)^{a}	91 (9.2)^{a}	
Recommended thickness (mm)		2	2	4	5	4	5	

Table 1: Light transmittance (at 448 nm) through two curing 'conventional' resin composites (CR) and four 'bulk-fill' (BF) types and without material at 1, 3 and 4 mm thicknesses. BF 1 = low viscosity, translucent; BF 2-4 = higher viscosity paste. All materials were stated as A3 shade, or equivalent, except BF 1. Light transmittance was measured through 6 mm diameter white nylon ring moulds using a fibre-coupled spectrometer with a 4 mm diameter cosine corrector sensor area. Generally, light transmittance increased at prolonged irradiation times for each thickness. Note the larger decrease in transmittance with depth of CR 1 and CR 2 (designed to be cured in 2 mm increments) compared with BF types. Longer measurement (and exposure) time of the CR materials at 4 mm thickness revealed significant increases in light transmittance. The introduction of some recent commercial BF materials are designed to exhibit reduced translucency for cosmetic purposes, highlighted here by the smaller, non-significant change in light transmittance at 1 and 3 mm throughout cure of BF 3 and BF 4; a result of finer control in optical matching of the filler and resin (Section 2.2.4). Non-significant differences identified by one-way ANOVAs and post-hoc Tukey comparisons ($P<0.05$) are represented by similar superscript letters within each material and thickness.

Incident light irradiance acting towards the upper surface of a resin composite restoration is greater compared with the bulk of the material due to attenuation and reduced irradiance throughout increasing depth. Surface reflection, light scattering and absorption by the resin components, filler particles, dyes, pigments and structures surrounding the restoration (e.g. Mylar, matrix band material, supporting structural tissue or existing restorative materials, etc) will also account for changes in irradiance and the complex initiation profile through depth. For most commercial resin composite materials light transmission increases throughout and following irradiation (Table 1; compare irradiance at 1, 10 and 60 s measurement time) as a result of dynamic, complex changes with decreasing photoinitiator concentration (Section 2.2.2) and optical pathlength due to shrinkage and possible lensing effects [6], increasing polymer density and refractive index, which approaches that of the filler phase (Section 2.2.4). However, for some modern resin composites, especially so-called "bulk-fills", manufacturers have recently designed materials to exhibit less change in translucency in surface layers (BF 3 and BF

4; Table 1) in order to better mimic the natural appearance of tooth tissue. It is therefore critical to achieve balance between opacity (for cosmetic purposes) and translucency to allow sufficient light transmittance and radical generation in deeper layers in order to achieve effective polymer conversion within an acceptable clinical timeframe (Figures 2 and 3). It is important to clarify that irradiation time is the most important parameter for effective curing of current resin composites [7,8], whether they be perceived as 'bulk fill', or not. Although the rate of polymerization is substantially reduced with increasing thickness for most materials, a similar degree of conversion can be achieved at 4 mm depth compared with 1 mm for conventional resin composites, albeit at clinically unfavorable curing times (~100s; Figure 2), and increased risk of over-heating the pulp. The expected higher temperatures associated with prolonged exposure will have also contributed to the increased conversion. The recorded inflection in degree of conversion at 360 s (light off) was the result of an experimental artifact and the abrupt change in temperature after the light was switched off, consequent change in volume and optical properties of the specimen (Figure 2). Following 10 s exposure of CR 1, the degree and rate of conversion at 4 mm was substantially reduced compared with that at 1 mm thickness, which is expected since conventional resin composite materials were not designed for curing with short exposure times at depths greater than 2 mm. However, for CR 2, the reduction in degree of conversion at 4 mm compared with 1 mm following 10 s exposure was similar to that of the bulk fill materials, BF 2-4, which is surprising given the requirements for the latter to at least double the depth of cure. Previous work has demonstrated significant 'dark' cure of bulk fills, 24 h after exposure [9, 10], although other work has identified the variation in degree of conversion and solvated hardness measurements (crosslinking) of bulk fill materials at depth [11], which may risk under-cured restorations, especially with low irradiance or short exposure times. The reduction in DC of BF 1 at 4 mm was negligible (Figure 2) since the material was more translucent and less viscous (lower filler loading).

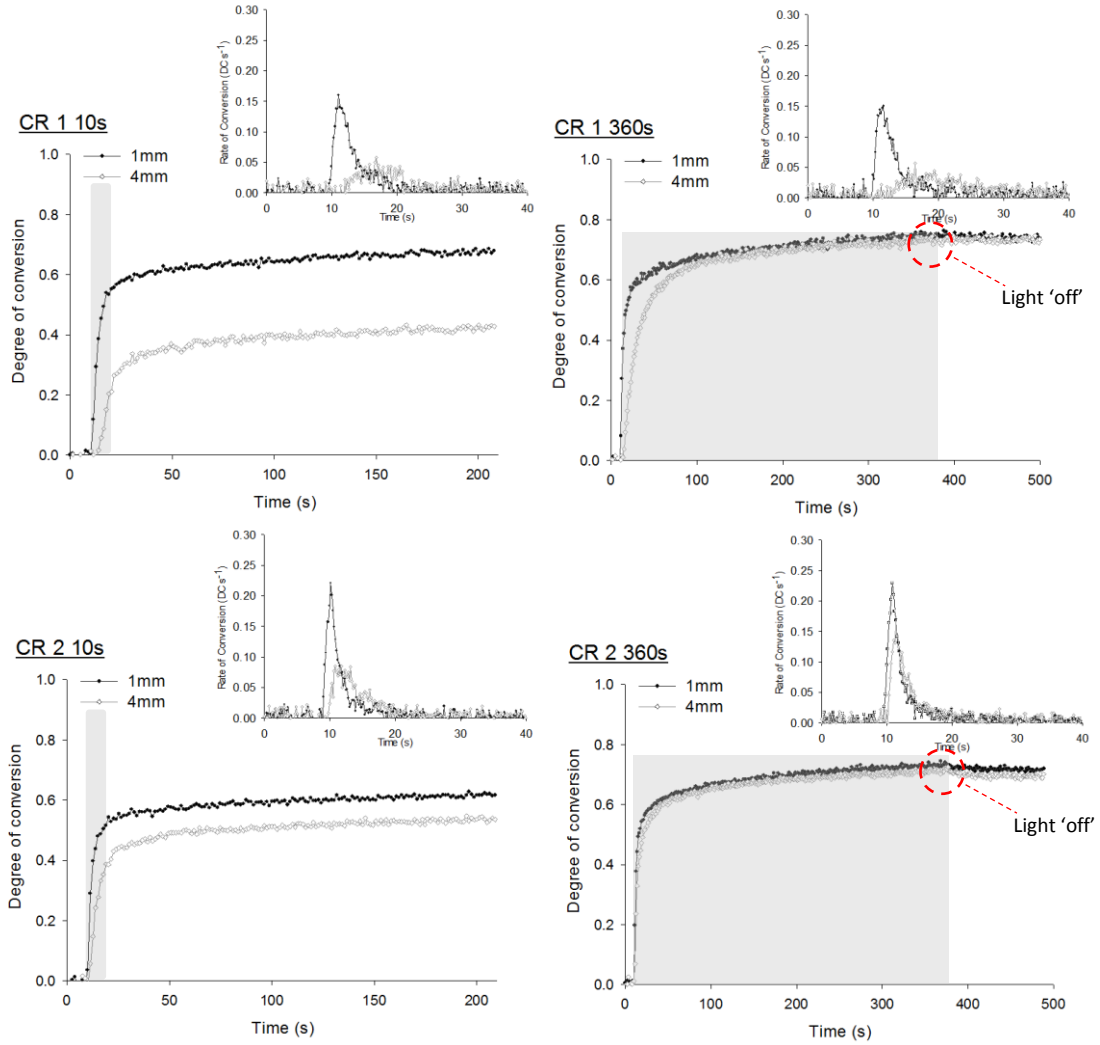


Figure 2: Degree of conversion and the rate of conversion (inset) measured on the lower surface of 1 mm thick and 4 mm thick (6 mm diameter white nylon ring moulds) conventional resin composites (CR) using attenuated total reflectance (ATR) Fourier Transform infrared spectroscopy (FTIRS) and cure times of either 10 s or 360 s (indicated by the shaded grey areas). Degree of conversion was assessed by measuring aliphatic carbon double bond absorption (1638cm^{-1}) using an internal aromatic isosbestic point (1598 cm^{-1} to 1608 cm^{-1}). Generally, degree of conversion and rate of conversion decreased with increasing thickness, but provided the exposure time is long enough, comparable degree of conversion can be achieved in increments of up to 4 mm.

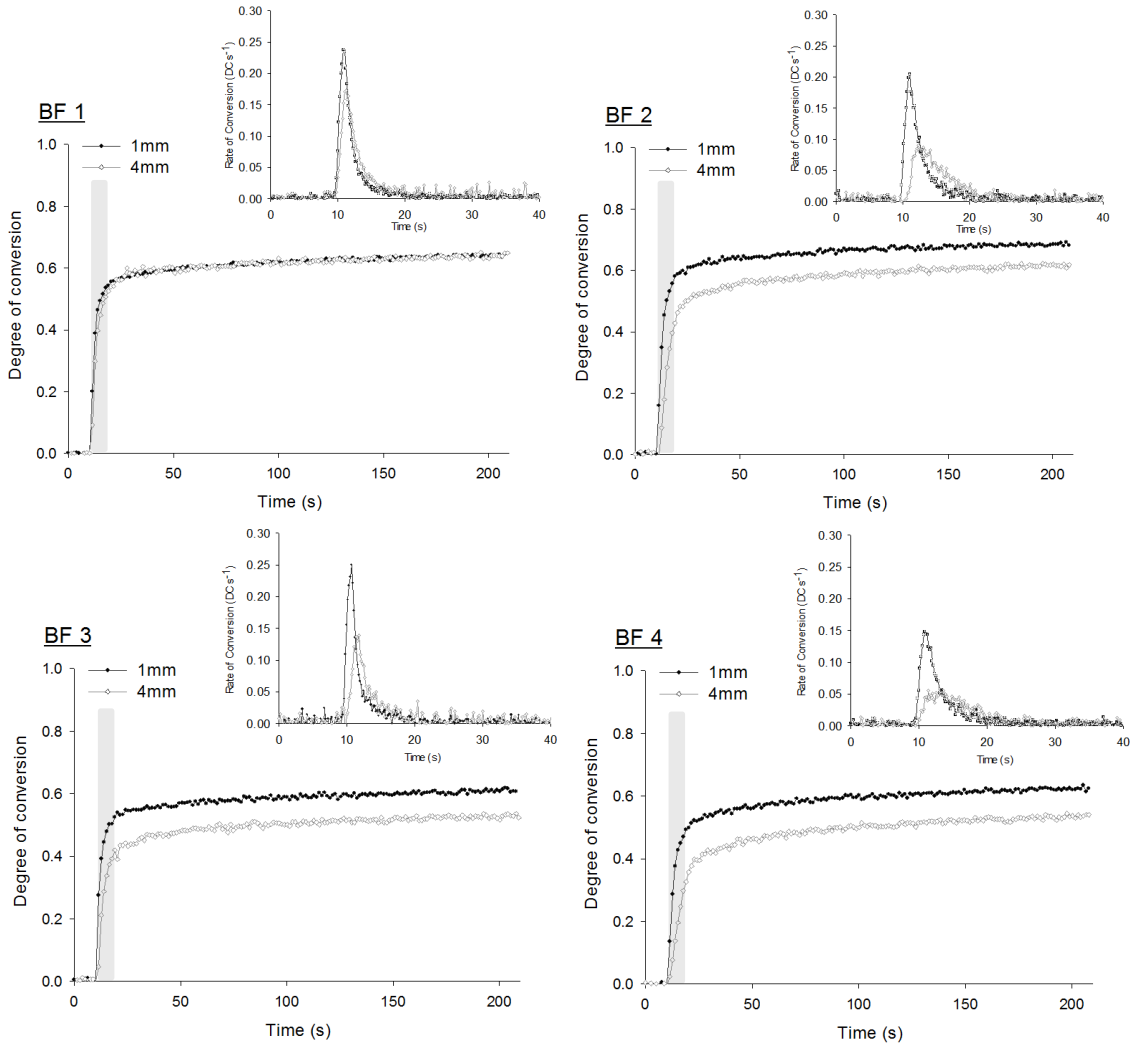


Figure 3: Degree of conversion and the rate of conversion (inset) measured on the lower surface of 1mm thick and 4mm thick (6mm diameter white nylon ring moulds) 'bulk-fill' (BF) resin composites using attenuated total reflectance (ATR)-FTIRS and cure times of 10 s (indicated by the shaded areas). Degree of conversion was assessed by measuring aliphatic carbon double bond absorption (1638cm^{-1}) using an internal aromatic isosbestic point (1598 cm^{-1} to 1608 cm^{-1}).

2.2 Light irradiance, depth and photoinitiation related to Beer-Lamberts Law.

The penetration of light towards the lower surface of a thick sample decreases exponentially due to Beer-Lambert's Law and the relationship between the light irradiance incident upon the material surface and within a depth of the material can be given by [12-14];

$$I_Z = I_0(10^{-\varepsilon C_z}) \qquad (1)$$

where I_z is the irradiance at depth z, I_0 is the incident irradiance at the surface (z = 0), ε is the molar absorptivity and C_z is the molar concentration of the photoinitiator at depth z. Upon light activation, the initial (uniform) concentration of the photoinitiator is consumed at a rate proportional to the local light irradiance, which therefore results in a complex description of an increasing initiator

concentration gradient from the illuminated surface combined with a concomitant decrease in irradiance with depth. Using a differential equation, the general assumption has been further developed for consideration of competitive light absorbers (from photolysis products), monomer resin absorption and photoinitiator diffusion, which vary with time as well as depth, such that [15];

$$\frac{\partial I(z,t)}{\partial z} = -\left[\varepsilon_i C_i(z,t) + A_m + \varepsilon_p C_p(z,t)\right] I(z,t) \qquad (2)$$

where I_z is the irradiance at depth z and time t, ε_i is the molar absorptivity of the photoinitiator, $C_i(z,t)$ is the molar concentration of the photoinitiator at depth z and time t, A_m is the absorption coefficient of the monomer and polymer (its variation is ignored), ε_p is the molar absorptivity of the photolysis product and $C_p(z,t)$ is the molar concentration of the photolysis product at depth z and time t. The following Sections 2.2.1-2.2.4 aim to address and review elements of Equation 2 applied to the use of light curable dental resin chemistry.

2.2.1 Molar absorptivity

How strongly a chemical species attenuates light at given wavelength, or the molar absorptivity (extinction coefficient) of the photoinitiator is an important consideration for photocuring thick resin materials. However, intrinsic competitive effects exist where higher ε provides an increased rate of initiation and faster rates of polymerization, yet low ε may result in a comparatively increased light transmission through the material since photon absorption is decreased [15]. The diketone Type II photoinitiator, camphoroquinone (CQ) strikes a good balance of molar absorptivity level for deep curing, and one reason why it remains state-of-the-art for commercial dental resin composites. Nevertheless, some manufacturers use multiple component photoinitiator systems that may contain additional Type I photoinitiators which absorb at shorter wavelengths, usually with much greater absorptivity, such as acylphosphine oxides. The possible reasoning here; to reduce the yellowing effect of CQ for cosmetic purposes and to improve curing characteristics through depth. The premise of the latter to provide both advantages of a higher and lower molar absorptivity (irradiated using different

initiation wavelengths appropriate for the absorption properties of both initiators, hence the use of multiple-diode LED curing lights) in order to achieve faster polymerization rates without severely reducing irradiance and to provide appreciable initiation rates at depth [15,16]. To demonstrate these effects, we have previously measured the absorption characteristics of Type I and Type II photoinitiators at various thicknesses (Figure 4), either using CQ (with a tertiary amine co-initiator, 2-(dimethylamino)ethyl methacrylate, four times the mass% of CQ), monoacylphosphine oxide (MAPO) or a combination of both to produce a ternary photoinitiator system, hence half that of each photoinitiator alone. Light transmittance was monitored at specific wavelengths that correspond with the irradiation light for each photoinitiator (405 nm and 468 nm for MAPO and CQ, respectively; Figure 5). The significant reduction in relative light transmittance of MAPO compared with CQ at early stages of photoinitiator consumption (<10 s) and thicknesses greater than 2 mm (Figures 5a and 5b, respectively) was attributed to the substantially higher molar absorptivity of MAPO (Figure 3). However, by using a combination of photoinitiators (whilst maintaining similar total photoinitiator concentration), specifically, the first that absorbs light strongly, and the second with weaker absorbing power, light transmittance at both wavelengths decreased to lesser extent (Figure 5c and 5d) compared with either photoinitiator used alone. In the mixed CQ/TPO photoinitiator system, an equivalent transmittance (at 468 nm and 405 nm, respectively) compared with the corresponding single initiator system would be expected if the initiator concentration were similar in each case. Here, total initiator concentration was kept constant, i.e. for the mixed initiator system, CQ and TPO concentration was half that of each initiator used alone, resulting in higher transmittance, which increased with thickness (Figure 5c and 5d). In terms of photopolymerization efficiency, it could be expected that the high molar absorptivity of MAPO would provide sufficiently high photoinitiation rates in surface layers despite the reduced concentration, whilst the weaker absorption band of CQ would ensure radical generation at greater depths. Using similar photoinitiator systems and photocuring conditions as described above, but in a typical dimethacrylate resin-composite, it could be observed that combined photoinitiator systems at half the concentration of individual photoinitiators resulted in similar DC at the materials surface, but with a significant improvement of cure depth, i.e. higher DC at increased

depths (>4 mm), compared with the use of CQ or TPO alone at full concentration (Figure 6). Three different curing parameters were used, but the same trend was observed in each case. The use of a mixed photoinitiator system with very different absorptivity and quantum yield, and irradiated with two different wavelength bands will significantly affect the rate of radical formation compared with a single photoinitiator and one corresponding wavelength band absorption. The total number of radicals generated from photoinitiator molecules is likely to be greater compared with a single photoinitiator of lower molar absorptivity (in this case CQ/DMAEMA). Each converted CQ molecule generates one dormant radical and one active radical that can initiate polymerization, whereas TPO is able to generate two active radicals [17]. While the combination of photoinitiators was clearly beneficial for conversion within deeper layers of the material (>3 mm, Figure 6), lower depths of cure have been reported using single photoinitiators with high molar absorptivity (e.g. TPO alone) [18]. Indeed, the quantum yield and resulting degree of conversion are reportedly lower in thinner layers (~2 mm) for mixed initiators than for TPO alone due to the generation of fewer radicals in the former system and a possible energy transfer from the most efficient to the less efficient photoinitiator [19, 20]. The same trend is observed in the data presented in Figure 6, which supports the need to optimise the co-initiator ratio and irradiation conditions to reach the best possible compromise between depth of cure and absolute degree of conversion at the material surface. Even though the initiator concentration is halved in the mixed system, under the current irradiation parameters, sufficient radical concentration is achieved in the uppermost layers due the significantly higher quantum yield of TPO irradiated at 405 nm. The rate of photoinitiation of TPO is limited by depth due to high molar absorptivity, however, the combination with an initiator of lower absorptivity provides decreased light absorbance in deeper layers thereby increasing radical production and degree of conversion with depth compared with a single photoinitiator alone.

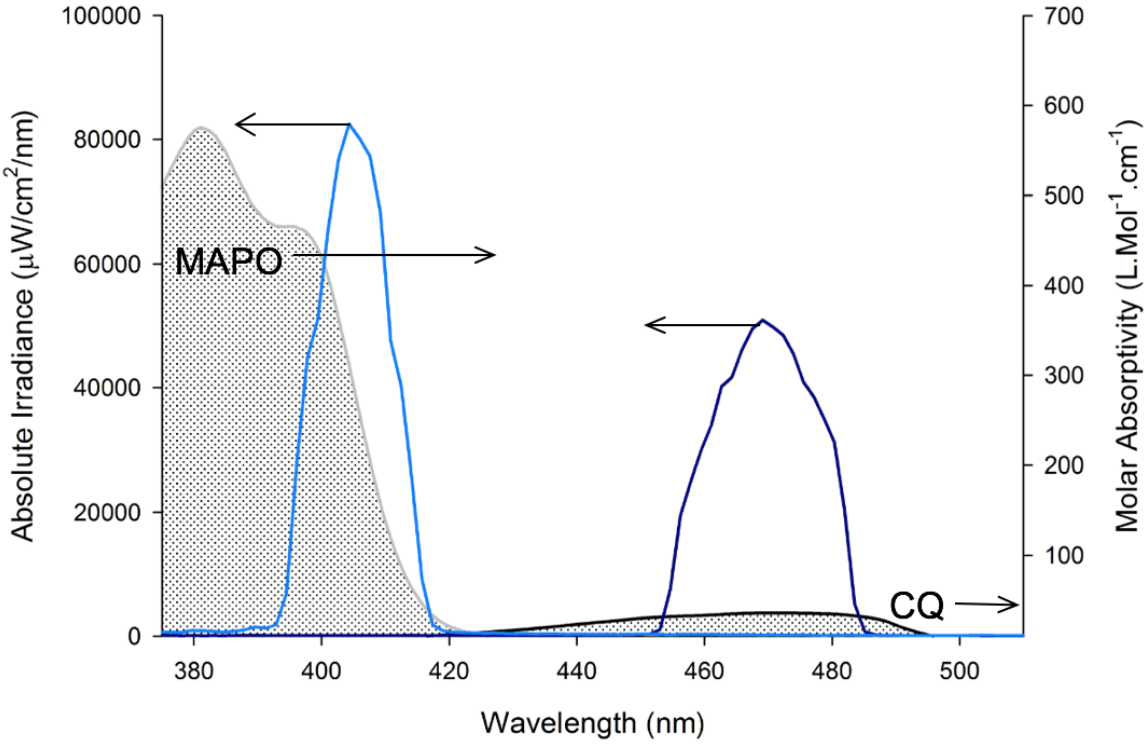


Figure 4: The spectral irradiance of the power-controlled light source (Aura, Lumencor, USA) with peak emission at 405 nm and 468 nm and corresponding absorption bands of the photoinitiators, monoacylphosphine oxide (MAPO) and camphorquinone (CQ), respectively. MAPO exhibits substantially higher molar absorptivity compared with the weaker absorption band of CQ. The initiating wavelengths approximate those used in commercial, so-called "polywave" (multiple diode) light curing units, although the irradiance of 405 nm used here is significantly higher compared with commercial devices.

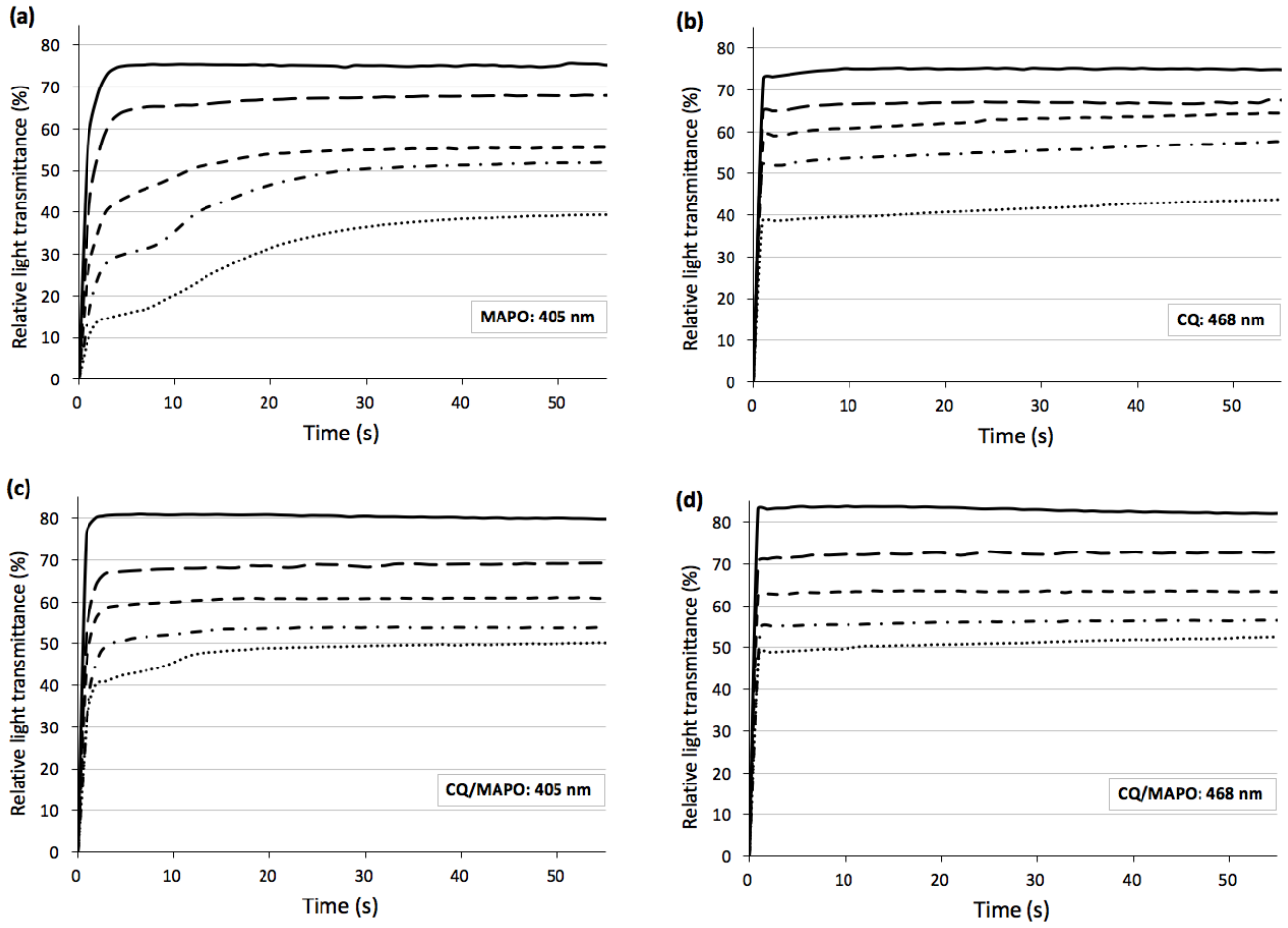


Figure 5: The percentage change in light transmittance (LT) of a) monoacylphosphine oxide (MAPO) measured at 405 nm, b) camphorquinone + 2-(dimethylamino)ethyl dimethacrylate (CQ) measured at 468 nm, c) CQ/MAPO measured at 405 nm, and d) CQ/MAPO measured at 468 nm. LT of each sample group was measured using a constant photoinitiator concentration of 0.0134 mol.dm^{-3} (equimolar for c and d) in methanol and at 5 thicknesses (1, 2, 3, 4 and 6 mm; highest to lowest LT) using an established UV-Vis spectrometry technique [17]. Each light source with a peak wavelength emission at either 405 or 468 nm (Figure 4) was used for LT measurements at a constant irradiance output of ~1500 mW.cm^{-2} for 60 s (LT change was calculated relative to irradiance measured at each thickness with an equivalent specimen without photoinitiator). Monitoring the consumption of photoinitiators by light transmittance measurements in a solvent circumvented the confounding factors associated with resin shrinkage (affecting pathlength) and interfacial scattering effects of the resin and filler in resin composite materials [6].

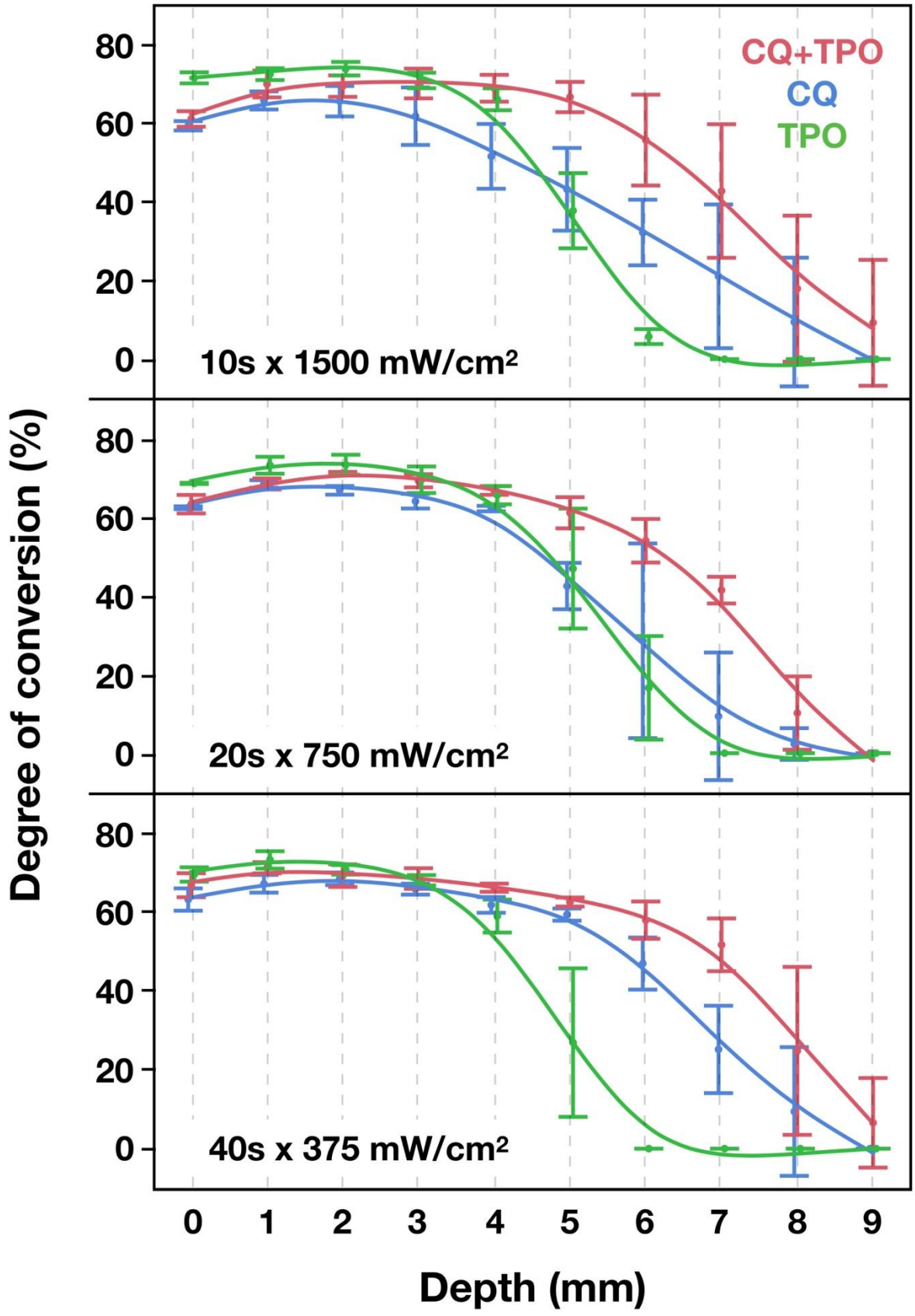


Figure 6: The degree of double bond conversion (n=3) of three model resin composite materials using similar photoinitiator systems (either CQ/DMAEMA, TPO, or a combination of both) and wavelength bands (either 468 nm, 405 nm or mixed, respectively) as in Figure 5. Resin composites were formulated with a 60/40 w/w of bisGMA/tegDMA and filled to 60 vol% using 1 µm diameter silicate (55 vol%) and 0.05 µm diameter fumed silica (5 vol%) filler particles. Degree of conversion was measured using Raman spectroscopy (780 nm laser, 5 scans, 400 lines/mm grating) by calculating the reduction of aliphatic C=C at 1640 cm^{-1} wavenumber using the isosbestic point of the aromatic C=C at 1608 cm^{-1} as an internal calibration standard. Three curing protocols were used, with similar radiant energy provided (15 J/cm^2).

Previous work has also identified the curing depth limitations of using high molar absorptivity photoinitiators alone [18, 19, 21]. Others have reported improved depths of cure when high and lower absorptivity properties are combined with dual initiation, although not usually exceeding those materials containing only CQ [22,23]. The present data, which have demonstrated significantly increased degree of conversion at depths greater than 4 mm for CQ/TPO compared with CQ alone (Figure 5), were acquired using higher irradiance at 405 nm compared with that generally used in commercial, so-called "polywave" (multiple diode), dental light curing units. Also, a reduction of the concentration of each PI may be beneficial (Section 2.2), and explain the improved depth of cure observed in the present proof of concept (Figure 6). Notwithstanding the potential limitations of decreased depth of cure, high irradiance curing (at 405 nm) with appropriate shorter wavelength band photoinitiators resulted in significantly increased polymer conversion and minimal monomer elution (of 2 mm thick specimens following 1 week immersion in a 75% ethanol solution) compared with CQ-based materials irradiated for 20 s [24].

A further example of modern photoinitiator developments in commercial dental products includes the use of benzoyl germanium derivatives. Specifically, the use of benzoyltrimethylgermane and dibenzoyldiethylgermane, amine-free, one-component (Type I) initiator systems, were reported with higher photo-reactivity, increased quantum yield, (the ratio of absorbed photons that deplete the initiator; 0.85 compared with 0.55-0.59 for acylphospine oxides [25] and less than 0.10 for CQ-based systems [26]) and the potential to reduce curing time and increase curing depth [27]. Synthesis of an optimal derivative, bis-(4-methoxybenzoyl)diethylgermane was patented in 2009 under the commercial name, Ivocerin™ (Ivoclar-Vivadent, Liechtenstein) [28].

2.2.2 Photoinitiator concentration

The photoinitiator must absorb light for any photochemical reaction to occur (Grotthuss-Draper Law) and therefore its concentration is a critical consideration for photopolymerization through depth. Previous research has reported the complexities of curing depth dependence on photoinitiator

concentration amongst other parameters (Equation 2) and presented theoretical models to predict the maximum depth of cure [14, 15, 29-34]. Higher concentrations provide increased rates of initiation, which increases conversion, however, light transmission is decreased at greater depths, which reduces cure. Although light transmission will increase with less photoinitiator molecules, a concentration that is too low will result in decreased radical concentration and lower conversion since less photons are absorbed [14]. Indeed, radical formation is not constant and is significantly affected by photoinitiator concentration, exposure time and depth. A spatially resolved rate of free radical production can be derived from the profiles obtained by Equations 1 and 2 and calculated by;

$$R_i(z,t) = 2\phi\varepsilon_i C_i(z,t) I(z,t) \quad (3)$$

where $R_i(z,t)$ is the local rate of radical formation at depth z and time t and ϕ is the quantum yield of the photoinitiator [15]. At early stages of photopolymerization (t<5 s), the rate of radical formation is highest towards the irradiated surface since depletion of the initiator remains low and decreases with depth at a rate that approximates the local light irradiance (Figure 7a; adapted from Miller et al., 2002 [15]). As more photons are delivered (t=25 s and 50 s) the rate of radical formation decreases at distances approaching the irradiated surface since there is progressively more initiator consumption in thinner layers, however, past the maximum rate, radical formation decreases as light irradiance falls (Figures 7b, c). This is commonly known as a 'polymerization wave front' and numerous models have been reported that describe the remarkably complex 'frontal photopolymerization' (classified as a type of frontal polymerization [35]) processes affected by photoinitiator concentration, molar absorptivity (the breadth of the wave front increases with less C_i and smaller ε_i), polymerization reaction kinetics and heat and mass transfer effects [15, 29, 30, 33]. Again, higher photoinitiator concentration limits light transmission and therefore rate of radical formation, whereas a lower concentration permits more efficient light transport at the detriment of a lower rate of radical formation (Figure 6b). Regardless of initiator concentration, a maximum rate of radical formation is expected at the specimen surface due to the time dependent differences in light irradiance acting at the surface and within deeper layers of the

material. Although the highest initiator concentrations and molar absorptivity will result in increased rates of radical formation and rate of polymerization, light transmittance and propagation of the wave front through depth are limited, restricting conversion in deeper layers. Previous fundamental studies of photopolymerization have identified the advantage of using low initiator concentrations for curing thick material sections [32, 36, 37], although a compromise must exist as appropriate irradiation times are required for dental clinical settings (perhaps ideally, >20 s, but with current standards, more likely ~10-20 s).

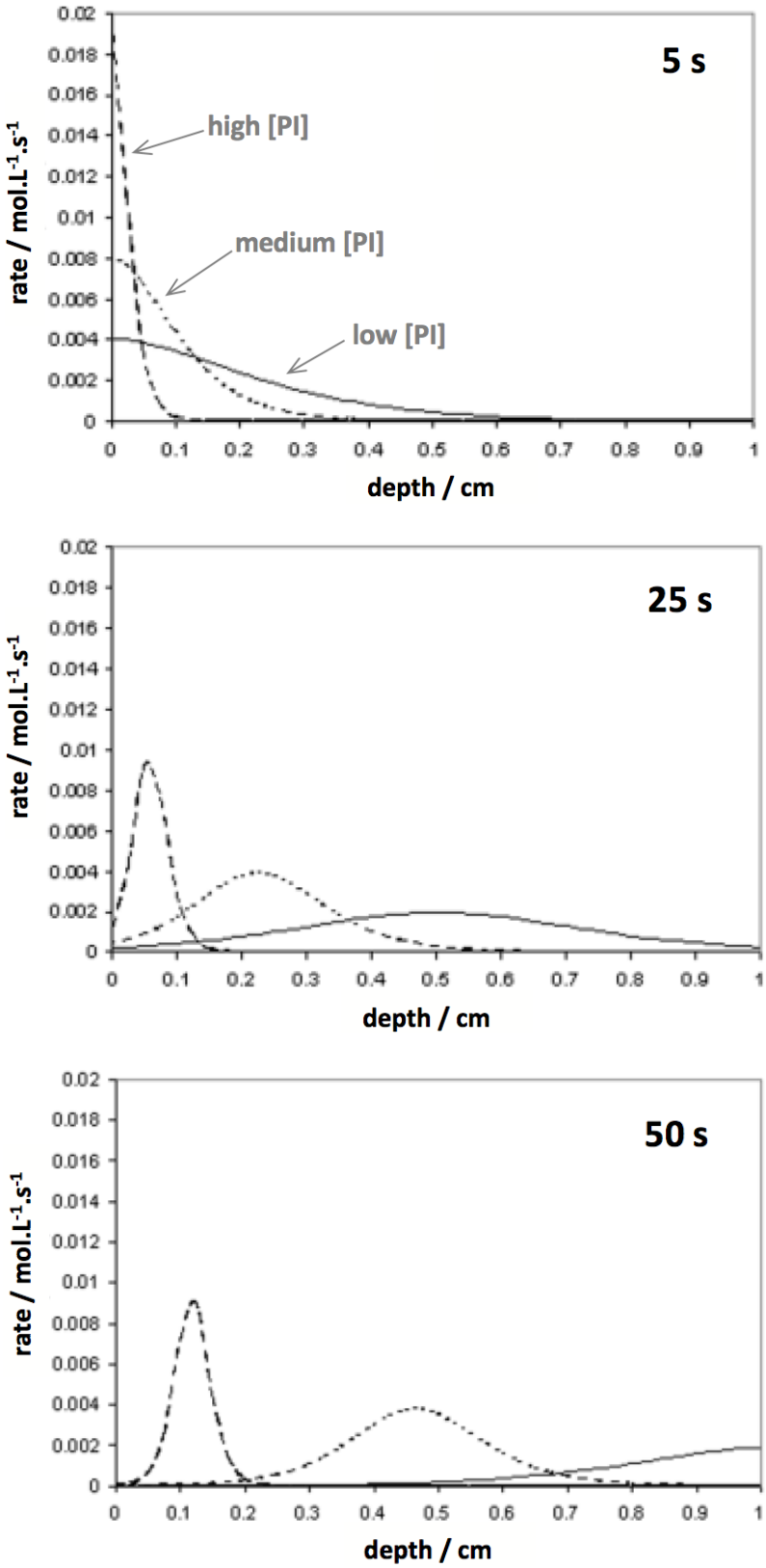


Figure 7: Theoretical radical formation rates with depth and time for three photoinitiator concentrations (high: 0.1340 M, 2.0 wt%; intermediate: 0.0537 M, 0.8 wt%; low: 0.0268 M, 0.4 wt% (assumed to completely bleach, without any photolysis product formation) at; (a) 5 s, (b) 25 s, and (c) 50 s irradiation time. Light irradiance, 10 $mW.cm^{-2}$; quantum yield, 0.2; absorptivity, 97.7 $L.mol^{-1}.cm^{-1}$ (adapted from Miller et al., 2002[15])

Several studies applied to the curing depth of experimental dental resin composite materials have also observed a significant decrease in degree of conversion and surface hardness with depth using photoinitiator and co-initiator concentrations above a critical level [38-41].

2.2.3 Competitive light absorbers

The use of photopolymers that exhibit low or zero absorption of the initiation wavelength, along with photobleaching initiators, provide better opportunity for greater light transmission and higher initiation rates through depth. Indeed, methacrylate resins and amine co-initiators commonly used in light curable dental materials do not absorb significantly between 420 and 520 nm [42], meaning that the component, A_m in Equation 2 is largely insignificant. However, if photobleaching is reduced or the formation of chemical species by photolysis competes with absorption by the initiation wavelength (i.e. ε_p and C_p in Equation 2), inevitably, radical formation and polymerization rates through depth will be negatively affected. Previous studies have reported specific photoinitiator systems related to dental use that produce light absorbing products [17,43], which may reduce the beneficial effects of total photoinitiator bleaching and decrease polymerization rates through the thickness of the material [44].

In order to mimic the colour properties and natural appearance of teeth, dental resin composites also contain pigments, usually iron oxides (red and yellow), opacifiers such as titanium dioxide and other colourants (blue) for opalescent effects (Figure 8a). These additives may absorb strongly around the absorption maxima of camphorquinone ($\lambda_{max} \sim 470$ nm; Figure 8b) and will therefore have remarkable effects on light transmission (Figure 8c). Light attenuation and curing depths are significantly affected by the material 'shade' (and pigment concentration) in commercial products [45] and a strong correlation exists between transmittance and depth of cure (Figure 8d). The particle size of inorganic pigments must be small enough to allow effective dispersion, although the high surface area may affect stability within the composite mixture [46] and it might also be reasonable to expect increased light scattering and further attenuation (Section 2.4). Organic pigments have been developed by synthesis of chromophoric sol-gel hybrid materials, which provide improved stability by immobilization within

silica materials. A further advantage is the increased colour intensity compared with conventional inorganic pigments that allows reduced pigment concentration and therefore improved curing depths. Specific colours of the organic chromophores also exhibit shorter wavelength absorption that compete less with that of camphorquinone [47].

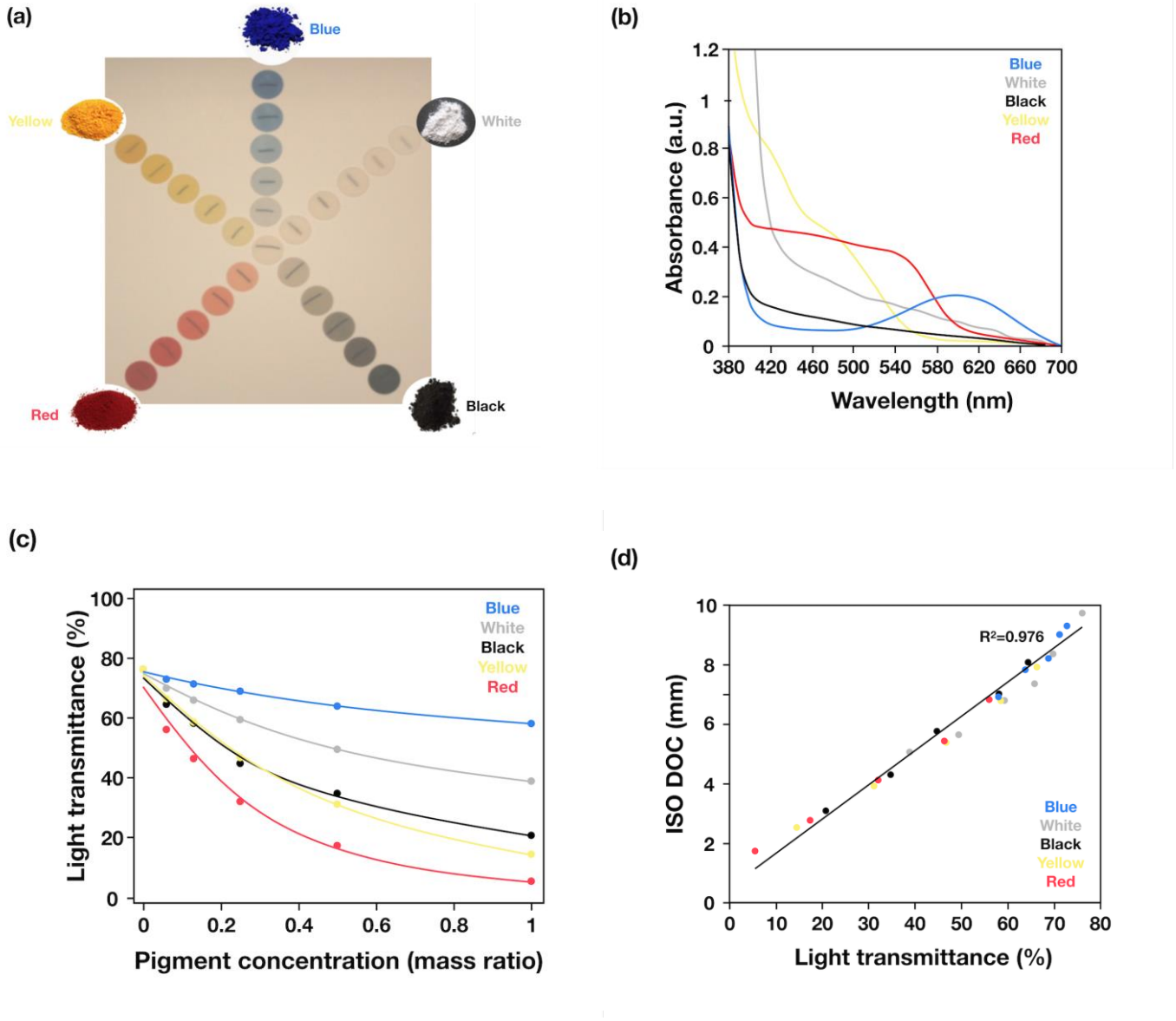


Figure 8: (a) Typical pigments used to adjust shade and optical properties of dental resin composites. (b) The spectral absorption of pigments in comparison with the relevant absorption band of camphorquinone and a phosphine oxide photoinitiator highlighting the potential competition for initiating light absorption. (c) The effect of increasing concentration of typical pigment dispersions used in resin composites on light transmittance (measured at 470 nm), and (d) the strong positive correlation between light transmittance and depth of cure (ISO4049;2009 [48]) using different pigments at varying dispersion concentration [data from 3M, St Paul, USA]

2.2.4 Light refraction and scattering

The required physico-mechanical properties of dental resin composites are achieved by optimization of filler morphology, and although particle loading in modern commercial materials may vary from ~50-85 mass% (in order to adjust viscosity, from low-fill 'flowables' to compact pastes) [49], the increase in light attenuation compared with the unfilled resin alone will be substantial. Previous research has reported the optical effects of filler properties on light transmittance, and maximum scattering is known to occur when the filler diameter approaches approximately one-half the wavelength of incident light, i.e. ~0.2-0.3 μm [50-53]. However, such a simple approximation is complicated by the large variations in filler particle size distribution, where more or less smaller sizes (< 500 nm) and agglomerates of micron-sized particles may exist. Further, as the particle size range widens, much larger fillers (> 10 μm) will lower the total interfacial area between the filler and matrix, reducing scatter. Some manufacturers of modern 'bulk-fill' resin composites have seemingly adopted this approach, presumably to aid higher curing depths [54-55].

Similar to the complex transmission profile of light as a consequence of passing through media that exhibit progressive changes in absorption properties, the dynamic process of light scattering is affected by non-linear optical properties of the resin by polymerization. Refraction (a type of scattering) is a process that affects the momentum of light and its propagation direction following interaction with a different medium (both are wavelength and temperature dependent). Refractive index of the medium describes how light is transmitted through it and compared with the speed of light in a vacuum, e.g. light will travel 1.360 times slower in ethanol than in a vacuum. The extent of scattering as light passes through two material interfaces is governed by Snell's Law of Refraction, where the difference in angle of incident and refracted light is minimized as the mismatch in refractive index between each material is reduced. If the refractive index of the monomer, polymer and filler phases are known (as well as absorption properties of the constituents), light transmittance through the curing material can be predicted. The optical properties of the filler remain unchanged prior to, or immediately following cure, however, the refractive index of the resin increases as polymerization proceeds with higher

material density. Generally, commercial materials are designed to increase translucency as the resin composite cures (Section 1, Table 1), hence the use of glassy fillers with refractive indices higher than the co-monomer mixture. It follows that resin composites can be formulated to increase, decrease or remain at a similar opacity prior to and following polymerization [56], which may have implication for pre-cure tooth shade matching and improvements to otherwise high translucency modern 'bulk-fill' materials.

3.0 The era of "bulk-fill"?

The ability to effectively photopolymerize resin composite restorations in 4 – 5 mm increments is certainly more convenient and may reduce operatory times for restoration of large cavities compared with a conventional 2 mm thickness limit. However, 'bulk-curing' is not a new phenomenon. Since the late 1990s manufacturers have marketed materials with up to 5 mm curing depth (Table 2). Previous studies have refuted such claims of earlier bulk-fill materials with degree of conversion being severely compromised at 5 mm depth [57-58]. More recently, the plethora of commercial round-robin testing of resin composite materials that advocate thicker incremental placement (≤ 5 mm) provide some evidence that, under ideal laboratory conditions, adequate levels of bulk polymerization are achieved. However, standard testing methodologies such as ISO 4049 depth of cure 'scrape' test (0.5 times the total depth), arbitrary proportional hardness (80%) of the irradiated surface, or proportion of the maximum surface hardness, may overestimate depth of cure [59-61]. Previous 3- and 5-year longitudinal clinical studies of bulk-fill resin composites have demonstrated similar failure rates compared with more conventional materials placed in 2 mm increments [62,63].

As with most types of dental resin composites, modern materials for deeper curing are available as pastes with a higher volume percentage of filler particles or as a lower viscosity base. The latter requires a higher viscosity cap material to allow sculpting and improved wear resistance, although either type may require an opaque layer if translucency of the base (for increased rates of initiation in deeper layers) precludes the desired cosmetic effect. More recently, bulk fill manufacturers have

adjusted the optical properties of the filler particles and resin matrix in order to create a greater mismatch in refractive index between each phase resulting in reduced translucency in surface layers (Table 1; BF 3 and 4).

e.g. Material	Approximate year of introduction	Manufacturers' description; quoted cure depth
SureFil Posterior, Denstply	1999	High viscosity, 60 vol%; **5 mm**
Alert, Pentron	1999	High viscosity, 84 mass%; **5 mm**
QuiXfil, Dentsply	2003	High viscosity, 86 mass% (66 vol%); **4 mm**
X-tra Fil, Voco	2003	High viscosity, core build-up, 86 mass%; **4 mm**
SDR, Dentsply	2009	Low viscosity, 68 mass% (45 vol%); **4 mm**
SonicFill, Kerr	2011	Reduced viscosity by ultrasonic energy, 84wt%; **5 mm**
Tetric Bulk Fill, Ivoclar	2011	Novel photoinitiator, prepolymer, low modulus filler, 61 vol%; **4 mm**
Filtek Bulk Fill, 3M	2014	High mW, low stress resin, 77 mass% (58 vol%); **5 mm**

Table 2: Some examples of commercial resin composite materials introduced to the dental market over the last two decades that claim curing depths of up to 5 mm, presumably tested using either the first incarnation of the ISO 4049; Depth of Cure "scrape test' in 1988 (the total scraped back 'cured' length divided by 2), or the same method in the updated standard in 2000.

Although there is no clinical evidence that establishes a direct correlation between polymerization stress and longevity of dental resin composites, its potential deleterious effects remain clinically meaningful [64] and a reason for continued research and development of existing chemistries and improved placement techniques that claim to reduce shrinkage stress. The general consensus for conventional resin composites has been to implement an incremental layering technique, which not only reduces the volume of material being set, but also reduces the ratio of bonded to free surfaces (decreasing the 'configuration factor'), increasing strain relaxation during the pre-gel phase of a curing material. In addition, an extended pre-gel phase has been reported in lower viscosity flowable resin composites, and with a reduced elastic modulus, may favour strain relaxation compared with higher viscosity materials [65-66]. Whilst modern materials for deeper curing are available as both high or low viscosities (Table 2), the latter providing the potential to alleviate developing stresses during polymerization, the contradiction to the previous consensus may cause some confusion over the placement of bulk fill materials. For example, larger incremental volumes and lower filler content may proportionally increase shrinkage as well as the potential for higher stress as the ratio of bonded to free

surfaces increases in larger cavities. Contrarily, previous in vitro reports have demonstrated at least similar or even significantly reduced negative effects of polymerization stress of bulk fill compared with conventional resin composites [67-71]. Whether the magnitude of polymerization stress (observed in vitro) correlates directly with in vivo failure has yet to be determined [64], and will always prove difficult given the 'unknowns' in commercial material composition and variations in clinical procedure [72]. However, there is plentiful previous and current innovation that attempts to reduce the polymerization stress of conventional and modern bulk fill materials. For bulk fill materials, a lower shrinkage stress is achieved by using materials that exhibit lower elastic modulus that might dissipate stress, or adapted and novel chemistries that provide greater stress relaxation. Bulk fill materials tend to exhibit lower elastic modulus compared with conventional 'nano-hybrid' types, although a range of stiffness values have been reported regardless of resin composite type [73], which may also result in varied polymerization stress levels. An example of modified chemistries in modern bulk fill materials includes the development of addition-fragmentation chain transfer agents where shrinkage stress is mitigated by continuous re-arrangement of the polymer network connectivity without a concomitant change in network chemistry or crosslink density [74]. Such chemistries are likely to involve the incorporation of allyl sulfide functional groups in ternary thiol-ene methacrylate polymerisations which has previously been suggested to reduce polymerization stress [75, 76]. Whilst the incorporation of chain transfer agents in diacrylate-functional monomers reportedly reduced shrinkage stress through participation in network formation, at higher percentages a significant reduction in polymerization reaction kinetics, onset of the gel-point and final double bond conversion was reported [77].

4.0 Emerging and potential future technologies for deep photocuring

Notwithstanding optimization of both conventional resin chemistry and optical and morphological filler properties to improve light transmittance through depth (Section 2), there have been several fascinating attempts to overcome depth limiting factors of photopolymerization. Garra et al. (2016) [78] have reported photointiation systems that allow photocuring of very thick (unfilled) specimens up to 8.5 cm at ~110 mW/cm2 (405 nm) for 20 mins. In that investigation, the generation of charge

transfer complexes (CTCs) between an iodonium salt, phosphine additives and an amine electron donor substantially extended the depth of photopolymerization since molar absorptivity (at 405 nm) of the starting compounds and their associated CTCs were very low. Some drawbacks were identified including heterogeneity of the polymer due to shrinkage in the long (9 cm) and narrow glass mould, and presumably a consequence of a low aspect ratio and number of bonded surfaces resulting in the generation of high polymerization stresses towards the edge of the mould. Also, a non-homogenous conversion profile was identified throughout depth (ranging from ~70 % to ~50 % conversion at the surface compared with 8 cm depth, respectively). Although a high optical transparency and prolonged exposures are required, which is likely to be problematic in highly filled or pigmented materials for a dental application, the development of this chemistry may allow for new technologies where very deep photocuring is required.

Since Rayleigh scattering is inversely proportional to the fourth power of wavelength, blue light will scatter to a greater extent compared with longer wavelengths (c.f. early UV-curable dental resin composites; Section 1). It follows that the use of longer initiating wavelengths will provide deeper photon transmission and therefore increased cure depths. For example, the use of near infrared (IR) irradiation and two-photon polymerization has been the subject of extensive research for the development of additive (stereolithographic) manufacturing processes. Two photon initiation provides improved resolution compared with conventional UV sources and photoinitiators and the ability to form three dimensional patterns beneath the surface of uncured resin [79]. However, the requirement for a broad absorption range of photoinitiators, high initiating irradiance, as well as a time-consuming spot curing process limits its use for a broad range of processing applications. A further approach for deep curing involves the use of upconversion (UC) phosphor particles that convert near-IR energy to shorter wavelengths (anti-Stoke shift) for UV or visible light initiation processes. For curing resins associated with use in dental applications, its premise relies on weaker Rayleigh scattering of IR to allow activation of UC particle luminescence and emission of shorter wavelength light that coincides with the absorption profile of a conventional visible light photoinitiator, e.g. camphorquinone (CQ;

λ_{max} ~470 nm). Essentially, the UC particles act as an internal light source that re-emit in the blue region for photoinitiation of CQ. The perceived advantage of a shorter activating distance between 2 UC particles and favorable photobleaching properties resulted in conversion of ~60 % up to depths of 13.7 cm using $NaYF_4$:18%Yb,0.5%Tm loaded resins [80]. Stepuk et al., (2012) [81] also reported an increased DC of a photoactive (CQ initiator) bisGMA/tegDMA resin (60/40) loaded with UC particles (also, $NaYF_4$ doped with Yb an Tm) at 10 mm depth irradiated with a 980 nm laser compared with an unfilled resin irradiated using a halogen lamp 400-500 nm. However, the energy requirements for effective photopolymerization were seemingly high: 160 W/cm^2 at 980 nm of IR excitation provided an UC irradiance of 0.05 $\mu W/cm^2$. Further improvements in polymerization rate were also observed using near-IR irradiation through an enamel slab since the penetration depth of 980 nm through enamel was over twice that reported for visible light (6.8 mm and 2.6 mm, respectively).

In conventional photocuring, and indeed for the potential emerging applications described hitherto, the rate of radical formation and initiation is localized to the penetration depth of photons (either excitation or direct initiation wavelengths), which therefore limits depth of cure since light transmission in opaque and highly scattering materials is reduced. Further, post-irradiation "dark cure" reactions are limited due to radical annihilation and increased rates of termination. In contrast, chemically cured materials are not limited by light curing reactions and offer the ability to synthesize very thick material layers, however, temporal control of the reaction is lost since base and catalyst components require mixing. Other issues with this approach (specifically for dental application) include the potential for non-homogenous mixing since the two components are folded, either by hand (rarely) or Kenics mixer, resulting in non-optimal reaction conditions and the incorporation of air bubbles [82]. The requirement of lower viscosities to enable mixing is usually achieved by reducing filler loading, which may also reduce mechanical properties. So-called "dual-cure" systems that provide photo- and redox chemistries provide some temporal control, although the initiation reactions work independently and similar problems remain in terms of mixing and setting time limitations. Stansbury's group have recently presented a quasi-biomimetic approach where novel photoinitiated

polymerization pathways mimic photosynthesis reactions of the Calvin cycle [83], i.e. light-independent chemical reactions that occur within the stroma of chloroplasts to produce glucose. Translating this natural concept to photoinitiated polymerization involves organic photoredox catalysis reactions to store metastable reactants that sustain radical formation hours following an initial and relatively short, low irradiance exposure. The use of methylene blue (absorption in the visible wavelength range), an amine reducing agent and oxidizing iodonium salt enabled the development of photocurable formulations that exhibited ~80 % conversion (HEMA) after ~120 mins following exposure and where initial conversion during light irradiation was ~8 %. Remarkably, ~1.2 cm thick HEMA specimens were cured using a low irradiance of 3.4 mW/cm^2 for 60 s without significant variation in conversion throughout depth. In contrast, under similar irradiation conditions, the use of a conventional CQ/amine photoinitiator system resulted in a steep reduction in conversion to zero even though optical transmission through the conventional material was 10 times higher compared with the novel organic photoredox system [83]. The authors point out that substantially higher intensities or longer exposures are needed to achieve a similar outcome with conventional photoinitiators (and analogous to the effects of extended cure times of conventional commercial resin composites demonstrated earlier in Section 2.1; Figure 2). Although the initial (methylene blue) dye turns instantly colorless due to the extremely photon efficient light activation step, the material cannot be practically considered for dental application since the latent radical production step of the process reforms the ground state methylene blue leading to an unavoidable colored final polymer. However, the concept of visible light organic photocatalysis is currently under development for use in bulk fill dental resin composites [84]

5.0 Conclusions

Photopolymerization has benefited dentistry since the 1970s and continues to offer unique advantages for controllable and effective curing of materials prepared at the chairside and in the dental laboratory. Although current “bulk-fill” dental resin composites are not an entirely new class of material, their current use in general dental practice has expanded due to ease of placement and time saved with less

incremental steps. Achieving serviceable material properties at the base of a thick resin composite restoration is critical for its longevity in vivo and is dependent on multiple parameters including light intensity and exposure duration, photoinitiator chemistry, morphology and density of the filler, pigments and other opacifiers, and highlighted by the strong correlation of light transmittance and depth of cure. Generally, for modern bulk fill materials there is a trade-off between tooth mimicry and translucency to allow for sufficient radical formation at depth and within clinically acceptable timeframes. Research of materials chemistry and related technologies for the development of photopolymerizable materials continues to advance within and outside the dental field, and there exists tremendous future potential for photocuring at remarkable depths that will not only benefit dentistry, but other medical and industrial fields.

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