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# Initial fracture resistance and curing temperature rise of ten contemporary resin-based composites with increasing radiant exposure

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ABSTRACT

*Objectives:* The principal objective of this study was to determine whether the bulk fracture resistance of ten light activated composites varied over a clinically realistic range of radiant exposures between 5 and 40 J/cm$^2$.
*Methods:* Ten operators were tested for clinically simulated radiant exposure delivery from a Bluephase® (Ivoclar Vivadent, Schaan, Liechtenstein) LED light to an occlusal cavity floor in tooth 27 in a mannequin head using a MARC®-Patient Simulator (Bluelight Analytics Inc., Halifax, NS) device. Notch disc test samples were prepared to determine the torque resistance to fracture (*T*) of the composites. Samples were irradiated with the same monowave Bluephase® light for 10 s, 20 s or 40 s at distances of 0 mm or 7 mm. After 24 h, storage samples were fractured in a universal testing machine and torque to failure was derived.
*Results:* Radiant exposure delivered in the clinical simulation ranged from 14.3% to 69.4% of maximum mean radiant exposure deliverable at 0 mm in a MARC®-Resin Calibrator (Bluelight Analytics Inc., Halifax, NS) test device. Mean torque to failure increased significantly ($P < 0.05$) with radiant exposure for 8 out of 10 products. The micro-fine hybrid composite Gradia Direct anterior (GC) had the lowest mean (S.D.) *T* between 10.3 (1.8) N/mm and 13.7 (2.2) N/mm over the tested radiant exposure range. Three heavily filled materials Majesty Posterior, Clearfil APX and Clearfil Photo-Posterior (Kuraray) had mean *T* values in excess of 25 N/mm following 40 J/cm$^2$ radiant exposure. Mean *T* for Z100 (3MESPE) and Esthet-X (Dentsply) increased by 10% and 91% respectively over the tested range of radiant exposures.
*Conclusions:* Individual products require different levels of radiant exposure to optimize their fracture resistance. Light activated composites vary in the rate at which they attain optimal fracture resistance.
*Clinical significance:* Unless the clinician accurately controls all the variables associated with energy delivery, there is no way of predicting that acceptable fracture resistance will be achieved intra-orally.



## 1. Introduction

Light-activated resin-based composites dominate the market for direct restorations because of increased patient demand for affordable aesthetic treatments. A long service life is possible for posterior composite restorations if patient, operator and materials factors are all controlled.[1] However the median longevity of direct posterior composites placed in

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dental offices is only 6 years.[2,3] Evidence from recent clinical studies identifies bulk fracture as a common cause of failure of large posterior composite restorations.[4,5] Fracture toughness is related to the ability of a material to resist the propagation of a crack from a critical flaw. The relatively low fracture toughness of dental composites makes them susceptible to bulk failure and marginal fracture or chipping.[6,7] Bulk fracture of composites has been correlated to fracture toughness from in vivo and in vitro investigations.[8,9] Fracture resistance is determined by material composition and test method.[10] Many investigations have reported on the fracture properties of dental composites. However it is difficult to make conclusions about the relative fracture resistance of different materials due to differences in test methods and experimental protocols.[11] Adequate polymerization is a fundamental requirement for predictable clinical service of composite restorations. A radiant exposure requirement, that is the product of irradiance and exposure time, may differ with material. Manufacturers recommend minimum radiant exposures for different shades of their products ranging between extremes of 5 J/cm$^2$ up to 40 J/cm$^2$. Manufacturers report only test data for products cured under ideal laboratory conditions. This does not account for myriad clinical variables such as underperforming light sources, light dispersion with distance, inadequate access or poor operator technique. The irradiance of commercially available dental curing lights ranges from below 300 mW/cm$^2$ to above 5000 mW/cm$^2$. Numerous surveys have shown that many practitioners' lights have inadequate output intensity (defined variously as irradiance $<$200 to $<$400 mW/cm$^2$). The percentage of inadequate units in these studies ranges from 12% to 95% with a median of 46%. Whilst the dental radiometers used in all but 2 of these surveys are inaccurate in absolute irradiance terms they give us an overall picture of clinical practice.[12,13] A South African survey reported a 100% satisfaction level by dentists with the performance of their light curing units even though nearly half of the units had inadequate output.[14] Prolonged irradiation time may compensate for low irradiance. However many practitioners use short radiation times. Predictable irradiance close to the specimen surface is readily achieved in the laboratory. Clinically, distances of up to 1 cm may occur between the resin and the light source.[15] Correct light alignment and stabilization may be difficult in posterior intra-oral locations. There is up to a tenfold difference in the

**Table 1 – Summary of the constituents and quantities/ratios of components contained in the RBCs.**

| Composite/batch number(s) | Classification | Matrix | Filler type | Filler load | |
|---|---|---|---|---|---|
| | | | | wt% | vol% |
| Filtek Z100 (Z100) 8YR & 7YP | Microfill | BisGMA TEGDMA | Zirconia/silica; 0.01–3.5 μm (84.5 wt%) | 84.5 | 66 |
| Filtek Z250 (FZ) 7MB & 8MB | Microhybrid | BisGMA UDMA BisEMA TEGDMA | Zirconia/silica; 0.01–3.5 μm (84.5 wt%) | 84.5 | 66 |
| Filtek Supreme Body (SuB) 7JH & AY | Nanofill | BisGMA UDMA BisEMA TEGDMA | Silica; 5–20 nm nanoparticle (8 wt%)<br>Zirconia/silica; 0.6–1.4 μm nanocluster (71 wt%) | 79 | 59.5 |
| Filtek Supreme Translucent (SuT) 7CT & 7EA | Nanofill | BisGMA UDMA BisEMA TEGDMA | Silica; 75 nm nanoparticle (40 wt%)<br>Zirconia/silica; 0.6–1.4 μm nanocluster (30 wt%) | 70 | 57.5 |
| Gradia Direct (GD) (anterior) 001969 | Microfill/hybrid | UDMA | Silica and pre-polymerized fillers (avg. Particle size 0.85 μm) fluoro-alumino-silicate glass | 73 | 64 |
| Esthet-X (EX) 60701102 | Micro-hybrid | BisGMA BisEMA TEGDMA | Barium alumino fluorosilicate glass (BAFG) < 1 μm. BAFG from 0.02 to 2.5 μm (with an average of from 0.6 to 0.8 μm)<br>Nano-sized silicon dioxide particles (10–20 nm) | 77 | 60 |
| Clearfil Majesty Aesthetic (ME) 010CA | Nanofill | Bis-GMA Hydrophobic aromatic dimethacrylate | Silanated barium glass filler (average; 0.7 μm)<br>Prepolymerized organic filler including nanofiller | 78 | 66 |
| Clearfil Majesty Posterior (MP) 008BB | Nanofill | Bis-GMA Hydrophobic aromatic dimethacrylate TEGDMA | Glass ceramic filler (average: 1.5 μm)<br>Surface treated alumina microfiller (average: 20 nm) | 92 | 82 |
| Clearfil AP-X (APX) 1222AA | Micro-hybrid | BisGMA TEGDMA | Barium glass particles (0.04 μm), silica, colloidal silica, silicon dioxide (0.1–15 μm) average = 3 μm | 85 | 71 |
| Clearfil Photo-Posterior (PP) 222BA | Micro-hybrid | BisGMA TEGDMA UDMA | Silanated silica, barium glass, colloidal silica<br>Particle size (0.04–20 μm) average = 4 μm | 86 | |

ability of different operators to deliver adequate radiant exposure even with the same light source.[16,17] A recent clinical simulation study assessed the ability of 20 operators to deliver energy to 2 posterior cavity locations in a mannequin head.[17] The energy delivered by the operators from the high power LED units ranged from 2.6 J/cm$^2$ to 20.4 J/cm$^2$. This represented <20% to >80% of attainable energy delivery.

An aim of this work was to see what influence clinically realistic extremes of radiant exposure would have on the fracture resistance of contemporary commercial composites. A secondary aim was to assess the temperature rise at the base of 2 mm thick samples of the test products cured at (a) 0 mm and (b) 7 mm light source distances. This mimicked (a) commonly reported in vitro test protocols and (b) the influence of a clinically relevant distance for the initial increment of composite in a deep proximal box.

This work was undertaken to test the following two null hypotheses:

1. Bulk fracture resistance or torque to failure (*T*) of a range of contemporary commercial composites would not vary with blue LED light radiant exposure between 5 J/cm$^2$ and 40 J/cm$^2$.
2. The rank order of *T* for individual test products would not vary with radiant exposure.

## 2. Materials and methods

Ten commercial resin based composites from four manufacturers were tested. All products were universal or A3 shade except for Filtek Supreme Yellow Translucent (3MESPE). The products, their classification, resin constituents and filler morphology (provided by the manufacturer where listed) and code designations are listed in Table 1.

### 2.1. *Operator radiant exposure delivery – MARC®-PS*

Ten operators (5 staff and five students; *n* = 3 replications each) were tested for clinically simulated radiant exposure delivery from a Bluephase® (Ivoclar Vivadent, Schaan, Liechtenstein) LED light (s.n. 2033440) to the floor of an occlusal cavity in tooth 27 in a mannequin head using a MARC®-PS (Bluelight Analytics Inc., Halifax, NS) unit. Inter-incisal opening was set at 43 mm.

### 2.2. *Torque to failure*

Notch disc test samples of 5 mm diameter and 2 mm thickness (±0.1 mm) were prepared in a 90° angle central notch white Delrin mould to determine the torque resistance to fracture (*T*) of the test materials. The geometry of the specimen is a circle with a central "V" section. Stress concentration is achieved at the apex of the notch by the application of a force applied by a cylindrical roller (attached to the testing machine). Fracture is accomplished with a two point mode I tensile load (Fig. 1).[18,19] After packing the test material into the mould, a Mylar matrix strip (E I du Pont, De Neymours & Co., Wilmington, DE 19898, U.S.A.) was laid over the material and a glass plate pressed on top to exude excess material. The glass plate was then removed and the exit window of a monowave Bluephase® blue LED light curing unit was positioned either directly onto the specimen surface at zero distance or held 7 mm away with a purpose made alignment jig. Ten specimens per test group were irradiated for 10, 20 or 40 s at 0 mm or 7 mm. The Bluephase® LCU had peak blue light output at 450 nm. Based on its active light delivering tip diameter (7.50 mm) it produced an irradiance of 1013 ± 2.5 mW/cm$^2$ with a laboratory grade thermopile power meter (PM10-19C, sn:1369C10R; Coherent Inc.). Irradiance output was also recorded with a MARC®-RC unit (Bluelight Analytics Inc., Halifax, NS) at 0 mm and 7 mm distances from the 4 mm diameter light sensor.

After polymerization, specimens were stored dry in lightproof containers for 24 h at room temperature (23 ± 1 °C). Surface peripheral flash was removed and specimen dimensions (thickness and height from notch apex to specimen base) were recorded. Derivation of *T* was accomplished as illustrated in Fig. 1 and as described by Uctasli et al.[18] Roller diameter was 3 mm. Fracture was accomplished using a universal testing machine (Instron 5544, High Wycombe, UK) in a two point opening (Mode 1) tensile load at a cross head speed of 1 mm/min.

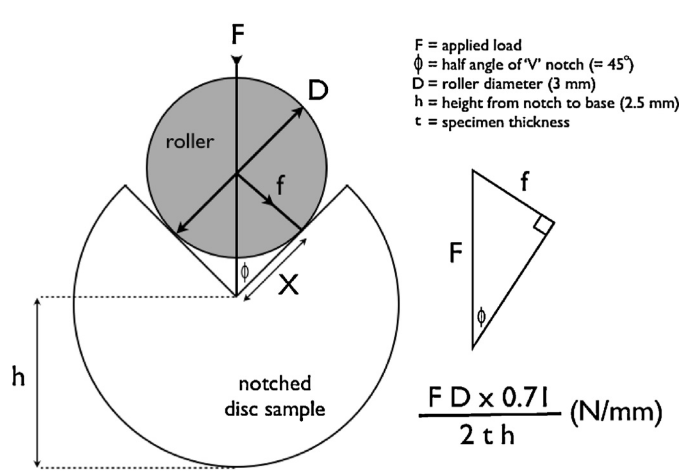



**Fig. 1 – Schematic representation of how the torque to initiate failure (*T*) is derived. The specimen has a 90° V notch, Ø = 45° and sin 45° = 0.71.**

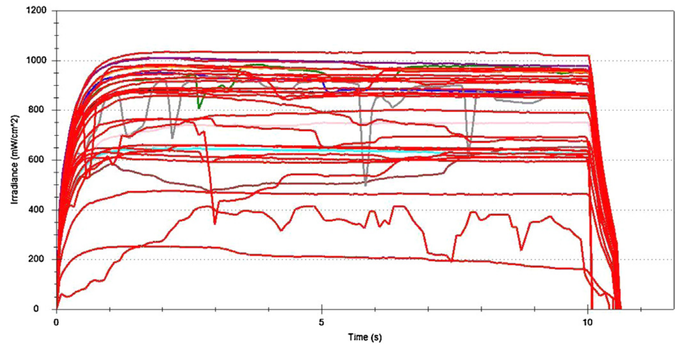



**Fig. 2 – Results – irradiance delivered by 10 operators (5 staff and 5 students; 3 tests per operator) to the occlusal cavity floor of tooth 27 in a MARC-PS® unit (Bluelight Analytics Inc., Halifax, NS) using the Bluephase® light for 10 s at high power setting. The distance between the cusp tips and the sensor in the cavity floor of tooth 27 was 4 mm. Inter-incisal opening of the mannequin head was fixed at 43 mm.**

### 2.3. *Radiation induced temperature changes*

Samples ($n = 3$) of each test composite were cured for 40 s in 2 mm thick black Nylotron moulds, aligned so that a miniature bead thermistor (246-045 RS Components, Corby, England) formed the central part of the cavity base. Spacer rings and an alignment guide ensured that the light guide exit window was maintained concentric and parallel to the sensor surface. The thermistor was set within a hollow brass stage through which water was circulated at $35 \pm 0.5$ °C.[20] During irradiation the combined temperature rise brought about by the light source and polymerizing composite specimen was recorded. Tests were conducted at both 0 mm and 7 mm test distances.

### 2.4. *Data analysis*

Statistical analyses of data from all three test procedures were undertaken with GLM, two- and one-way ANOVA and post hoc Holm–Sidak tests at a 95% confidence level. Independent factors for two-way ANOVA testing were radiation time (3 levels) and light source distance (2 levels). Pearson correlations were performed between *T* and filler loading by weight for the test products.

## 3. Results

### 3.1. *Operator radiant exposure delivery – MARC®-PS*

Irradiance and radiant exposures delivered to the sensor in the cavity floor of tooth 27 varied considerably (Figs. 2 and 3). Radiant energy delivered ranged from 2.15 J/cm$^2$ to 10.42 J/cm$^2$ representing 14.3–69.4% of deliverable mean radiant exposure as determined using the same LCU tested in a MARC®-RC unit at 0 mm distance (Fig. 4). Staff delivered significantly ($P < 0.05$) higher mean energy levels 9.2 (0.7) J/cm$^2$ than student operators 6.8 (2.3) J/cm$^2$.

### 3.2. *Torque to failure*

Boxplots of the *T* for materials irradiated at 0 mm and 7 mm distances are shown in Figs. 5 and 6. Mean values and standard

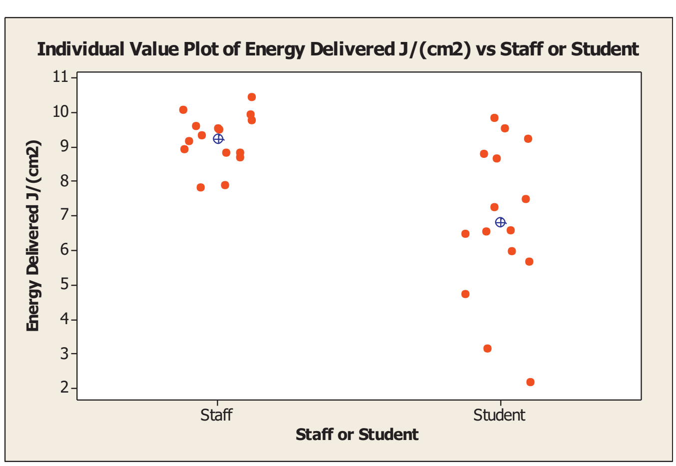



**Fig. 3 – Results – plot of the individual radiant exposure (J/cm$^2$) results for the ten operators. Circled crosshairs show the mean for each of the 2 operator groups. Note the higher scatter and lower mean energy delivered by the less experienced student operators.**

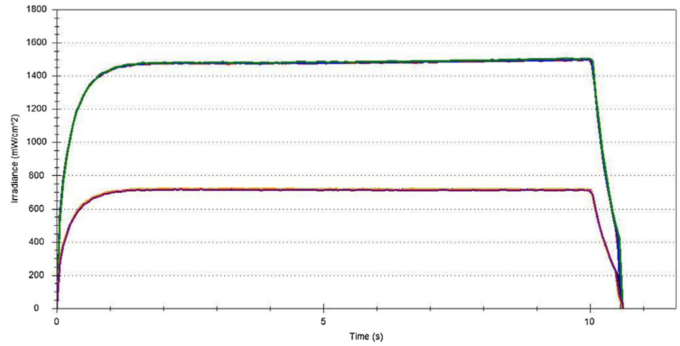


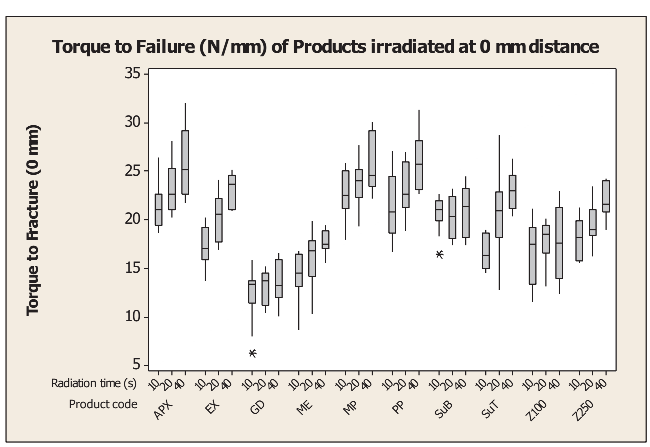

**Fig. 4 – Results – irradiance delivered ($n = 3$ tests at each distance) by the Bluephase® light at used for 10 s at high power setting at 0 mm and 7 mm distances from the light sensor (4 mm diameter) of a MARC-RC® unit (Bluelight Analytics Inc., Halifax, NS).**

deviations (S.D.'s) are given in Table 2 together with a summary of the two-way ANOVA test results. A pattern of increasing mean *T* was found with increasing radiant exposure. Heavily filled materials with larger mean particle size distributions (APX, Majesty Posterior and Clearfil Photo Posterior) yielded higher *T* values. Significant Pearson correlations were found between *T* and filler loading by weight but only at low levels of radiant exposure ($P = 0.004$ at 5 J/cm$^2$; $P = 0.017$ at 10 J/cm$^2$). Two-way ANOVA tests for each product revealed that the variable distance was significant ($P < 0.05$) for all products except for Z100 (Table 2). Radiation time was significant ($P < 0.05$) for all but 2 products – Table 2.

Analysis of the data at matching radiant exposures of 10 J/cm$^2$ (10 s at 0 mm or 20 s at 7 mm) and 20 J/cm$^2$ (20 s at 0 mm or 40 s at 7 mm) revealed no significant difference ($P > 0.05$) in mean *T* across products (Table 2). Mean *T* at the tested extremes of radiant exposure are displayed in Fig. 7 and Table 2. The average increase in *T* was 40% and ranged from 10% for Z100 to 91% for Esthet-X. The rank order of *T* for these latter 2 products reversed ($P < 0.05$) as radiant exposure increased.

### 3.3. *Radiation induced temperature changes*

Mean peak temperature rise ($n = 3$) at the base of 2 mm thick samples during polymerization are shown in Fig. 8 for 0 mm and 7 mm light source to specimen distances. Mean (and S.D.) peak temperature rise values for the light source at these distances were 6.73 (0.03) °C and 2.11 (0.04) °C. Two-way ANOVA established that product and distance were both highly significant as was their interaction ($P < 0.001$). The power of the performed tests with alpha = 0.05 was 1.00 validating the small set size used in this experimental series. Temperature rise during polymerization at the base of the samples was greater for all materials than for the light itself apart from the heavily filled product Majesty Posterior which yielded the lowest peak temperature rise of 4.5 °C. Curing from the 7 mm distance always gave lower temperature rises than at 0 mm distance ($P < 0.05$). The Filtek Supreme Translucent gave the greatest mean peak temperature rise of 14.7 °C at 0 mm distance (Fig. 8).




APX = Clearfil APX, EX = Esthet-X, GD = Gradia Direct (anterior), ME = Majesty Esthetic, MP = Majesty Posterior, PP = Clearfil Photo Posterior, SuB = Supreme Body shade, SuT = Supreme Translucent, Z100 = Z100, Z250 = Z250.

**Fig. 5 – Results – torque to fracture (N/mm) at 0 mm irradiation distance.**

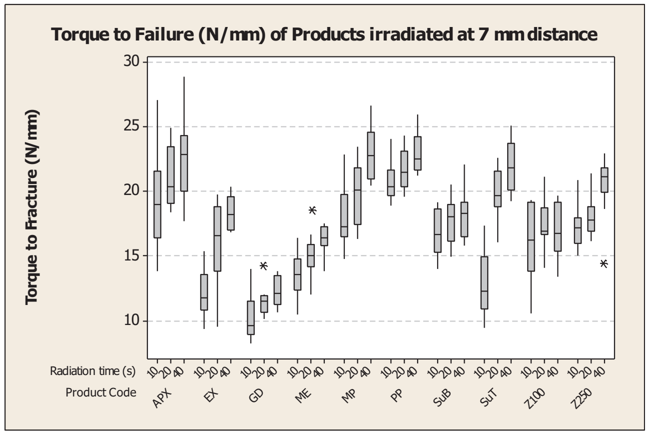



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**Fig. 6 – Results – torque to fracture (N/mm) at 7 mm irradiation distance.**

## 4. Discussion

Measurement of the plain strain fracture toughness of resin-based dental composites has been reported using various test methods and specimen geometries. A major problem in mechanical testing is the production of test samples that adequately reflect the dimensions of clinical restorations. Whilst test methods for assessing plain strain fracture toughness $K_{Ic}$ have been scientifically validated they often involve clinically unrealistic specimen dimensions with volumes up to 840 mm$^3$.[18] Light curing large volumes of composite creates difficulty in achieving adequate polymerization. The small notched disc technique facilitates uniform polymerization of small test pieces (40 mm$^3$) approximating the dimensions of dental restorations which may be irradiated in a single cure cycle.[18,19] It introduced the concept of torque to failure. Stress concentration is accomplished at the notch apex by the application of a force applied via a cylindrical roller which is attached to the testing machine. Fracture is accomplished in a two point Mode I tensile load manner. In the clinical situation cusp fissures of posterior teeth act as stress concentration sites and when load is applied via an antagonist cusp Mode I tensile loading is induced.[19]

The first null hypothesis that the fracture resistance of tested products would not increase with increasing Radiant Exposure over the range 5–40 J/cm$^2$ was rejected. The second null hypothesis that rank order of fracture resistance would not vary between products with increasing radiant exposure over 5–40 J/cm$^2$ was partially accepted as it applied for the majority of tested products.

Mechanical properties of composites are related to their filler content.[21] Miyazaki et al.[21] investigated the influence of three irradiance levels at matching radiant exposures

**Table 2 – Torque to failure results showing means and standard deviations (in parentheses). Also shown are the *P* values for independent variables of two factor ANOVAs for each product and their Interaction terms.**

| Radiation time | 10 s | | 20 s | | 40 s | | Two way ANOVA | | |
|---|---|---|---|---|---|---|---|---|---|
| Distance | 0 mm | 7 mm | 0 mm | 7 mm | 0 mm | 7 mm | Time | Distance | Interaction |
| Radiant exposure (J/cm$^2$) | 10 | 5 | 20 | 10 | 40 | 20 | | | |
| Z100 | 16.8 (3.3) | 16.0 (3.1) | 17.8 (2.4) | 17.5 (2.1) | 17.6 (3.8) | 17.1 (2.1) | 0.35 | 0.47 | 0.95 |
| Z250 | 18.1 (2.1) | 17.3 (1.7) | 19.6 (2.0) | 18.0 (1.5) | 22.0 (1.9) | 20.4 (2.6) | <0.001 | 0.012 | 0.824 |
| SuB | 20.6 (1.9) | 16.8 (1.8) | 20.3 (2.2) | 17.8 (1.8) | 21.1 (2.6) | 18.3 (2.0) | 0.336 | <0.001 | 0.56 |
| SuT | 16.7 (1.7) | 12.9 (2.6) | 21.0 (4.4) | 19.8 (1.9) | 23.1 (1.9) | 22.0 (2.0) | <0.001 | 0.004 | 0.186 |
| GD | 12.4 (2.9) | 10.3 (1.8) | 13.0 (1.8) | 11.6 (1.2) | 13.7 (2.2) | 12.3 (1.2) | 0.033 | 0.002 | 0.827 |
| EX | 17.4 (2.0) | 12.1 (1.8) | 20.3 (2.5) | 16.1 (3.2) | 23.1 (1.8) | 18.3 (1.3) | <0.001 | <0.001 | 0.749 |
| ME | 14.3 (2.5) | 13.6 (1.7) | 16.1 (2.9) | 15.1 (1.8) | 17.7 (1.2) | 16.3 (1.1) | <0/001 | 0.039 | 0.849 |
| MP | 22.7 (2.4) | 18.0 (2.4) | 23.9 (2.3) | 19.9 (2.5) | 25.8 (3.0) | 23.0 (2.0) | <0/001 | <0/001 | 0.499 |
| APX | 21.4 (2.3) | 19.3 (3.7) | 23.2 (2.6) | 21.1 (2.4) | 26.0 (3.7) | 22.5 (3.2) | <0.001 | 0.002 | 0.713 |
| PP | 21.4 (3.5) | 20.8 (1.6) | 23.1 (2.7) | 21.7 (1.6) | 25.9 (3.0) | 23.0 (1.6) | <0.001 | 0.013 | 0.345 |

Z100 = Z100, Z250 = Z250, SuB = Supreme Body shade, SuT = Supreme Translucent, GD = Gradia Direct (anterior), EX = Esthet-X, ME = Majesty Aesthetic, MP = Majesty Posterior, APX = Clearfil APX, PP = Clearfil Photo Posterior.

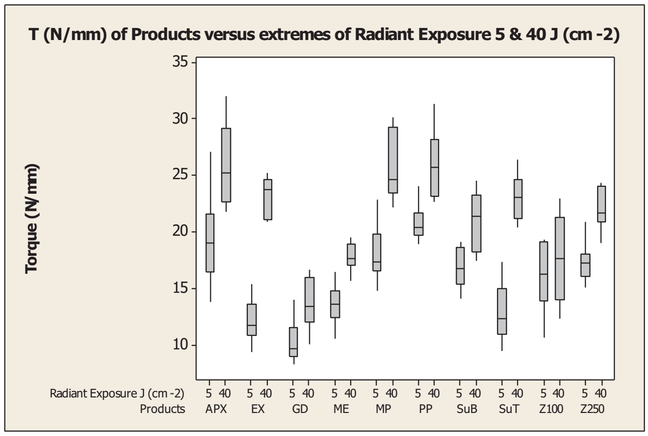


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**Fig. 7 – Results – mean T (N/mm) to failure at 5 and 40 J/cm$^2$ radiant exposure.**

(12 J/cm$^2$) on the fracture toughness and flexural strength of 4 composites. The large particle size composite Clearfil APX yielded higher fracture toughness and flexural strength values than Z100 despite similar filler loading. Kim et al.[22] have reported that high filler loading was not necessarily associated with high fracture toughness. In this study greater *T* values were observed for more heavily filled products. Whilst Majesty Posterior had the highest filler loading of any material under test it did not yield increased *T* values compared with Clearfil APX or Clearfil Photo Posterior. Fracture toughness values of hybrid and nano-particle resin composites are significantly higher than those of micro-filled resin composites.[23] Watanabe et al. reported mean $K_{Ic}$ values of 0.53 and 0.48 for the composites Filtek Z250 and Filtek Supreme and of 0.38 MPa m$^{1/2}$ for the microfilled/hybrid Gradia Direct (anterior) consistent with the current results. Loomans et al.[24] reported results for simulated marginal ridge fracture resistance of Filtek Supreme, Clearfil APX and Majesty Posterior. Clearfil APX had statistically superior fracture resistance. Their test method was closer to an edge load bearing capacity test than a bulk fracture test. Degree of conversion of the resin matrix is also an important co-determinant of the mechanical properties of composite resins.[25–29]

Studies have reported lower conversion and slower cure for the micro-hybrid composite Esthet-X in comparison to other products.[30–32] Assuming that 40 s is the longest practical radiation time Esthet-X required a greater radiant exposure to optimize fracture resistance compared to the other tested products. Aside from in resin matrix chemistry differences manufacturers employ different of photoinitiators, accelerators and inhibitors in their products and these impact on polymerization kinetics. Early hardening is not necessarily

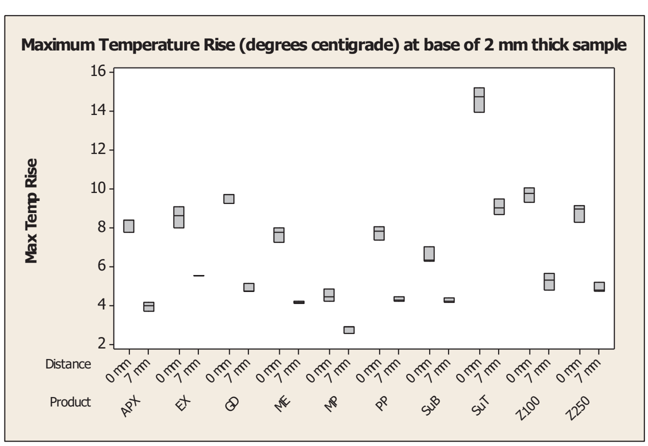


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**Fig. 8 – Results – maximum temperature rise at base of 2 mm thick samples of polymerizing products irradiated at 0 mm or 7 mm distances.**

equivalent to faster conversion.[33] Emami and Soderholm[29] concluded that only relatively low energy levels (5–15 J/$cm^2$) were required to produce high conversion levels in 2 mm thick Z100 and Z250 samples. It may be that the higher filler loading and greater diluent TEGDMA matrix component and/or differences in the photoinitator system of Z100 contribute to the very rapid hardening of this product.

The fact that mean peak temperature rise during polymerization at the base of the 2 mm thick test samples was greater for all but one of the polymerizing materials than for the light source irradiating the empty test cavity is a reflection of two competing influences. The polymerizing material generates heat in addition to the light but the material attenuates light transmission heating in depth. Curing from the 7 mm distance yielded lower temperature rises than at 0 mm suggesting that increased source distance may offer some protection to the dental pulp from thermal challenges when curing resins in deep cavities with high power light sources.[20] Arbitrarily increasing the radiation time is not an acceptable way of guaranteeing high conversion because of the increased risks of thermal insult to soft tissues and polymerization contraction stress. Stansbury has shown how stress for dental polymers rises steeply towards end stage conversion.[34]

Whilst light activated materials may yield optimal mechanical properties when tested following high levels of radiant exposure as is routine in laboratory based tests this ideal is frequently not met in clinical practice. Studies have shown that there is a high degree of variability between individual operators when performing light curing intra-orally[16,17] and as confirmed by the current work. Price et al.[17] found that radiant exposure delivered over a 10 s exposure using optimally performing light units ranged between 2.6 4 J/$cm^2$ and 20.4 J/$cm^2$ depending on light source and cavity location. Access was limited to the occlusal cavity in this work because it was a second molar tooth. Also the light guide hampered access compared to a low profile light source exit window and the sensor in the cavity floor lay 4 mm below the level of the cusp tips.[17] In this study the radiant exposure delivered ranged from 2.2 J/$cm^2$ to 10.4 J/$cm^2$ for a 10 s exposure. Thus even if a 20 s exposure had been used energy delivered may not have reached 5 J/$cm^2$ level in a worst case scenario despite the high unit irradiance at zero distance. Staff delivered significantly higher mean energy levels ($P < 0.05$) than student operators who had not been trained on MARC®-PS before testing. A recent retrospective study from the University of Texas at San Antonio has found a tenfold higher early failure rate for class II composite restorations in comparison to class II amalgam restorations in the dental school's predoctoral clinic.[35] The authors' speculations regarding possible causes of failure included improper position of the curing light. The disappointing results for the less experienced student operators in the current work highlights the need for training of undergraduate student operators in light curing. Seth et al. have reported that training on MARC-PS® immediately after testing resulted in a significant improvement in energy delivery and that this benefit was retained 4 months later.[36] Whilst manufacturers, researchers and standards bodies test optimally cured materials this may not reflect the performance of materials cured under routine clinical practice conditions. In the current work a single LED light source was used. The decline in the irradiance of different LED lights varies significantly with distance.[37–39] To achieve adequate intra-oral polymerization the dentist needs to control all the four core variables of energy delivery accurately.[40]

## 5. Conclusions

Manufacturers, researchers and standards organizations assess products following polymerization with high levels of radiant exposure. Materials testing should also be undertaken at sub-optimal as well as high levels of radiant exposure to reflect the situation as may occur in clinical practice.

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