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Abstract. This study aimed to determine the feasibility of using optical coherence elastography to measure internal displacements during the curing phase of a light-activated, resin-based composite material. Displacement vectors were spatially mapped over time within a commercial dental composite. Measurements revealed that the orientation of cure-induced displacement vectors varied spatially in a complex manner; however, each vector showed a characteristic evolution with time. Precise real-time individual displacements were estimated to be ~1 to 2 μm, enabling micrometer time-varying displacements to be detected. © The Authors. Published by SPIE under a Creative Commons Attribution 3.0 Unported License. Distribution or reproduction of this work in whole or in part requires full attribution to the original publication, including its DOI. [DOI: 10.1117/1.JBO.21.4.046001]

Keywords: optical coherence tomography; optical coherence elastography; resin-based composite; dental composite; polymerization shrinkage.

1 Introduction

Light-activated, resin-based composites (RBCs) are widely used in both industry and medicine. In dentistry, they have largely replaced mercury-based dental.

Light cure RBCs are typically comprised of a monomer (i.e., bisGMA, UDMA, TEGDMA), a filler material (silica and various bioglasses), and a photoinitiator system (Camphorquinone/dimethylaminoethyl methacrylate). Illumination of the RBC with light, corresponding to the photoinitiator absorption wavelength, initiates polymerization and consequent hardening of the resin into a solid material.

Optimization of the spatial cure characteristics of RBCs has been the subject of considerable research driven by certain shortcomings of RBCs. For example, they are well known to exhibit shrinkage during cure. In dentistry, shrinkage presents a clinical and technological challenge resulting in stress at the tooth-restoration interface, separation fracture and secondary caries. Furthermore, varying degrees of polymerization are expected to result in a spatially varying elastic modulus.

The dynamic development of RBC elastic modulus is of importance because it necessarily affects shrinkage stress at the tooth-restoration boundary and thus impacts upon the likelihood of failure. However, there are currently no widely available tools to visualize and measure internal physical characteristics of RBCs in real-time, throughout cure. Therefore, finite element computer modeling has been used to estimate three-dimensional distributions of shrinkage stress and movements within restorations although curing dynamics have not been investigated.

Experimentsally, curing has previously been investigated using low-coherence interferometry (LCI) to measure the refractive index change and shrinkage of RBCs. The consequence internal defects of RBC restorations have been imaged using optical coherence tomography (OCT), an LCI-based imaging technique. Outside of dentistry, OCT has been used to evaluate the structure of composite materials and applied to engineering applications.

Optical coherence elastography (OCE) uses OCT to measure spatial variations in the elastic modulus of materials. Under uniaxial loading conditions, the elastic modulus \( E = \sigma / \epsilon \), where the stress \( \sigma = F / A \) is defined by a force \( F \) applied over an area \( A \), and the resulting strain \( \epsilon = \Delta l / l \) is the ratio of length change \( \Delta l \) to the original length \( l \). OCT measures localized deformation in response to a known applied stress. However, for optically cured dental composite, the applied stress is spatiotemporally varying in multiple directions, thus making the direct determination of the spatial elastic modulus nontrivial. Consequently, this work has been restricted to the measurement of local deformation vectors. This is a central step in OCE, and therefore, this term is used for consistency with other literature. To the authors’ knowledge, OCT/OCE has not previously been applied to measuring internal deformation within curing dental composite.

Therefore, this article sets out to determine the feasibility of using OCT/OCE to dynamically measure the internal deformations within light-cured RBCs due to polymerization shrinkage.

2 Materials and Methods

A commercial RBC was investigated (Enamel shade A2, IPS Empress Direct, Ivoclar Vivadent). Following the manufacturer’s instructions, it was cured for 20 s with a dental curing lamp (Optima 10, BA International Ltd). The RBC was measured before, during, and throughout cure for a total duration of 60 s using a custom-built fiber-optic OCT system. The light source was a superluminescent light-emitting diode (SLD1325, Thorlabs Ltd) operating nominally in the 1250 to 1400 nm wavelength and was detected by a custom spectrometer. The spectrometer comprised a 92 kHz line scan camera (SU1024-LDH2, Sensors Unlimited) and a reflective diffraction grating.
grid of 84 regions of interest (ROIs), each measuring $\mu = 9.5$ of acquired over 500 A-scans, yielding a nominal lateral pixel size within 50 ms, followed by a delay of 550 ms between B-scans. The total acquired B-scan image width was nominally 3 mm.

The OCT system was configured to acquire B-scans at a rate of one every 600 ms. The spectrometer of 100 logarithmically scaled OCT B-scan intensity images was configured as depicted in Fig. 1 and was repeated three times.

The RBC was placed into a circular washer (diameter = 4 mm, thickness = 0.75 mm). A glass microscope slide (nominal thickness 1 mm) provided a transparent substrate through the curing lamp illuminated the sample. A series of 100 logarithmically scaled OCT B-scan intensity images were obtained at a rate of one every 600 ms. The spectrometer integration time was 100 $\mu$s, enabling single B-scan acquisition within 50 ms, followed by a delay of 550 ms between B-scans. The total acquired B-scan image width was nominally 3 mm acquired over 500 A-scans, yielding a nominal lateral pixel size of 6.9 $\mu$m. For comparison, the axial pixel size was nominally 9.5 $\mu$m.

The OCT images were analyzed numerically using a custom MATLAB script (Mathworks). Each B-scan was divided into a grid of 84 regions of interest (ROIs), each measuring $M = 25 \times N = 25$ pixels ($\sim 173 \times 238 \, \mu m^2$) in the lateral and axial directions, respectively. The cross-correlation of each ROI, $C_i$, was calculated between corresponding ROIs in the first OCT image in the time series, ROI$_0$, and each subsequent B-scan, ROI$_t$, at time $t$ using Eq. (1).

$$C_i(p, q) = \sum_{m=0}^{M-1} \sum_{n=0}^{N-1} ROI_0(m, n)ROI_t(m - p, n - q). \tag{1}$$

The local displacement vector, $\mathbf{dr} = [p_0, q_0]^T$, at each time-point and within each ROI was estimated by calculating the centre of mass, $\mathbf{R}$, of the cross-correlation, Eq. (2), and subtracting from it the ROI centre, $[p_0, q_0]^T$.

$$\mathbf{R} = \frac{1}{\sum_p \sum_q C_i(p, q) \sum_p \sum_q C_i(p, q) \begin{bmatrix} p \n q \end{bmatrix}}.$$

For comparison with the internal displacement measurements, the bulk optical specimens thickness was estimated for each B-scan. The average specimen A-scan profile along the axial (z) dimension resembles a rectangular function of the form $\Phi(z) = 0$ if $|z| > 0.5$, 0.5 if $|z| = 0.5$, 1 if $|z| < 0.5$. A smooth approximation to the rectangular function can be constructed from the product of two error functions, Eq. (3), which is well behaved under parameter optimization.

$$\Phi(z) = A \text{erf} \left\{ \gamma (z - (z_0 - w/2)) \right\} \text{erf} \left\{ \gamma (z_0 + w/2) - z \right\} + k. \tag{3}$$

Parameters $A$, $z_0$, $k$, and $w$ represent the maximum intensity, z-position, noise floor, and specimen thickness, respectively. A constant edge steepness parameter $\gamma$ was manually determined to approximately match the A-scan signal intensity gradient at the material surface. The specimen thickness $w$ was obtained at each time point by minimization of the sum of the squared difference between Eq. (3) and the mean A-scan intensity profile.

### 3 Results

The mean thickness change for all three experiments is plotted as a function of time in Fig. 2. The error bars represent the 95% CI.

![Fig. 2 Mean optical thickness change for the three dental composite specimens measured over a 60 s period. Error bars represent 95% CIs for the mean thickness change over three independent experiments.](image-url)
Displacement vectors were calculated at each time point for each ROI. These are shown as white arrows in Fig. 3 for time points $t = 0.6$ s, $t = 3.6$ s, $t = 5.4$ s, and $t = 7.8$ s. At a given point in time, the direction of the displacement vectors does not follow a smooth spatial distribution, but instead exhibits a complex arrangement. However, each vector was observed to systematically evolve over time both in direction and magnitude.

This systematic evolution is shown in Figs. 4(a)–4(d) for four selected displacement vectors. The specific vector locations are...
identified in Fig. 3(d) as corresponding sub-images i–iv. The top
of the image represents the side furthest from the curing light
from which the samples were imaged with OCT. The plots have
been aligned such that \( t = 0 \) s represents the time at which
cure was initiated. Positive axial displacement represents
movement toward the curing light (bottom of the OCT image).

Figure 4(a) shows a vector in the top left of the specimen at
the furthest point from the curing light. Axially, a displacement
of \( \sim 6 \, \mu m \) is observed over the 20 s curing period. Laterally,
there is no statistically discernable change, although the vectors
indicate a movement toward the center. The top-center vector,
represented by Fig. 4(b), exhibits similar behavior, with an axial
displacement of \( \sim 8 \, \mu m \) observed over the entire measurement
period, including postillumination. The central vector, Fig. 4(c),
events \( \sim 3 \) to 4 \( \mu m \) displacement away from the curing light for
the first 10 s of illumination, although this appears to cease
thereafter. Laterally, there is \( \sim 1 \, \mu m \) of movement to the right
that occurs for the entire illumination period with an apparent
reversal during the postillumination period. Figure 4(d) shows
the displacement of a region near to the illuminated surface. An
axial displacement of 6 \( \mu m \) was measured over the entire illu-
mination and postillumination period. Systematic lateral dis-
placement of 2 \( \mu m \) was observed postillumination. Notably,
the top-center vector [Fig. 4(b)] moves toward the light
down) by \( \sim 8 \) to 9 \( \mu m \) compared to the bottom-center vector
[Fig. 4(d)], which shows a displacement of 5 to 6 \( \mu m \). Thus,
the net displacement of the top and bottom surfaces was in
the range of 3 to 4 \( \mu m \), which agrees with the bulk shrinkage
plot, Fig. 2, to within the 95% CI.

To characterize these results, three forms of precision were
calculated in terms of the displacement standard deviation,
\( \sigma \). The ROI precision \( \sigma_{\text{ROI}} \) represents the repeatability of
displacement measurements from a single ROI in a given speci-
men. Each ROI precision was estimated from all 16 displace-
ments measured during the final 10 s of each experiment.
During this time, the displacement change due to material
cure was assumed to be negligible.

The mean intraspecimen precision \( \sigma_s \) is the mean of \( \sigma_{\text{ROI}} \)
over all 84 ROIs in a single specimen. Along with its 95% CI,
it estimates the repeatability of a single displacement meas-
urement from a given specimen regardless of ROI. This provides
a baseline level of measurement noise present in the analysis,
although displacement sensitivity can be improved by averaging
over multiple measurements. Finally, the precision reproducibility
\( \sigma_r \) was assessed across all three specimens as the standard
deviation of \( \sigma_s \). This provides an estimate of how much the over-
all specimen precision varies between experiments.

The mean intraspecimen precision \( \sigma_s \) and associated 95% CI
was determined over all 84 ROIs within the B-scan field of
view. The resulting specimen-dependent lateral and axial dis-
placement precisions are shown in Tables 1 and 2, respectively,
for all three experiments.

Table 1 shows that the intraspecimen lateral precision was
found to range between 1.06 and 1.80 \( \mu m \) across all the
three specimens. Axially, Table 2 shows a greater spread, with
a mean intraspecimen axial precision range of 1.48 to 5.16 \( \mu m \).
Axial precision in sample 1 was less than half that in samples
2 and 3. Similarly, sample 1 also exhibited lower lateral preci-
sion than the other samples.

Nevertheless, experimental reproducibility \( \sigma_r \) of the
precision estimates was assessed from all three repetitions of
the experiment. Laterally, \( \sigma^2_{\text{rep}} = 0.31 \, \mu m \) compared to an axial
reproducibility of \( \sigma^2_{\text{axial}} = 1.70 \, \mu m \). In each experiment, the lat-
eral displacement was found to have a greater precision than the
axial precision. However, over all the experiments, no statistical
difference could be discerned (Mann-Whitney U Test, \( p > 0.05 \)).
This was largely due to the lower precision observed in sample 1
and the relatively low number of specimens. Notwithstanding
this, there is nothing visible in the image data to suggest a prob-
lem with sample 1, and therefore, the result should be considered
representative of the experimental reproducibility.

### 4 Discussion

Displacement precision was greater laterally compared to
axially. This is because the lateral pixel spacing (6.9 \( \mu m \))
was \( \sim (6.9 \, \mu m - 9.1 \, \mu m)/9.1 \, \mu m = 24\% \) smaller than the lat-
eral PSF FWHM (9.1 \( \mu m \), compared to the axial pixel spacing
(9.5 \( \mu m \) that was \( (9.5 \, \mu m - 7.8 \, \mu m)/7.8 \, \mu m = 21\% \)
larger than the corresponding axial PSF FWHM (7.8 \( \mu m \)). Thus,
laterally, the PSF FWHM is sampled more than once due to the
pixel spacing, but axially the PSF FWHM is under-sampled.
This is critical because the speckle size is dependent upon the
PSF. Insufficient sampling of the speckle necessarily reduces
displacement precision. Lateral sampling density can be con-
trolled experimentally, although for this work it was preferred
to maximize the scan width relative to the number of A-scans
permitted by the specific OCT implementation. The axial pixel
spacing, \( \Delta z \), in a spectral domain OCT system is propor-
tional to the reciprocal of the detected spectral width \( \Delta \nu \), i.e.,
\( \Delta z \propto 1/\Delta \nu \). The spectrometer design can be improved to
optimize axial speckle sampling, although eventually this
amounts to zero-padding the spectrum and consequent sinc
interpolation. Alternatively, phase sensitive algorithms can
measure displacement with nanometer precision.\(^{23,24}\) However,
phase-based methods are limited to measuring axial components of displacement vectors. Speckle tracking is advantageous because it has the potential to track three-dimensional deformation directly. This is important for dental composites that have been shown to exhibit complex local displacement. Furthermore, phase sensitive detection requires processing of the raw interferometric data and places stringent requirements on experimental stability. In laboratory systems such as ours, phase information is readily available, although such data are not necessarily accessible in commercial OCT systems.

The precision values reported in this paper represent single-point measurements. However, for continuous time-series measurements as shown in Fig. 4, submicrometer relative displacement changes can be detected in both axial and lateral dimensions. It is these systematic and continuous changes that support the findings, that OCE can detect polymerization-borne displacements occurring within a curing composite material.

Overall, less displacement was observed laterally than axially. It is expected that bonding to the sides of the circular retaining ring resisted lateral displacement, while axially the unconstrained composite surface allowed a higher degree of deformation. At the base of the specimen, the composite becomes fixed as it adheres to the glass slide. However, away from the boundary, the composite is relatively unbound. Thus, polymerization shrinkage results in the unbound material moving toward the bound RBC. Hence, close to the interface, displacement vectors point toward the illumination lamp.

The results show that away from the glass–RBC boundary, the local displacements follow a complex spatial arrangement rather than a continuous deformation as seen in compression experiments with solid phantoms. The observed spatial arrangement of displacement vectors is similar to the turbulent flow of a viscous fluid. However, the RBC is not a homogeneous liquid, but exhibits varying degrees of monomer-polymer conversion and solid filler material. Local displacement occurs when there is a net applied stress. The source of the applied stress is polymerization shrinkage, which acts in combination with adhesion of the resin at its constrained boundaries, adhesion to filler particles, and friction generated by interacting filler particles. Computational models could be used to explore this and define the distribution of internal stress. Experimentally, polarization sensitive OCT may allow direct measurement of the shrinkage stress. The deformation results imply elastic modulus heterogeneity. However, OCT makes optical measurements with an inherent ambiguity between changes in refractive index, \( n \), and physical deformation, \( d \). The bulk optical thickness measurements shown in Fig. 2 therefore represent the product \( w = nd \). Previous work \(^{16}\) resolved this ambiguity, justifying the approximation that physical thickness change dominates for bulk material. Whether this is true for local displacements remains to be fully elucidated. Therefore, future research will focus upon understanding the relationship among optical deformation measurement, shrinkage stress, and clinical outcomes.

The results presented in this paper demonstrate the feasibility of using OCE to dynamically measure and spatially map local polymerization displacement within a light-cured composite material. A baseline level of precision and reproducibility has been estimated based upon speckle tracking from B-scan images. These results have applications in a variety of fields from clinical dentistry to aerospace engineering.

References