

Femtosecond laser ablation of dental restorative materials toward digitally enabled precision dentistry

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Abstract

Replacing a dental restoration requires removing restorative material while preserving sound tooth structure. Conventional rotary instruments are effective but offer limited control and produce smearing, debris and mechanical damage. Although indirect restorations are designed and fabricated under digital control, their removal remains an operator-dependent procedure. This exploratory study quantified femtosecond-laser ablation of 13 direct and indirect restorative materials, assessing thresholds, efficiency, surface morphology and ex vivo intrapulpal temperature. Thresholds ranged from laser fluence $0.6\text{--}2.0\text{ J cm}^{-2}$ for direct and $0.9\text{--}5.5\text{ J cm}^{-2}$ for indirect materials, indicating overlapping rather than fully separable processing windows. Ablation efficiency rose steeply with fluence and approached a near-plateau by $6\text{--}8\text{ J cm}^{-2}$ for both groups. Across $500\times\text{--}20,000\times$ magnifications, laser-processed surfaces showed no obvious cracking or melting, whereas rotary-cut surfaces displayed greater smearing, debris and grooving. Because maintaining pulp vitality is important, ex vivo intrapulpal temperature measurements were used to define thermally safe operating windows under defined irradiation and cooling conditions. Finally, a proof-of-concept experiment in a human tooth demonstrated femtosecond-laser cavity preparation, conventional composite placement and subsequent laser removal. These findings establish quantitative processing windows and provide a basis for precise, digitally controlled restoration removal.

Introduction

Removal of dental hard tissues and restorative materials is conventionally performed using rotary instruments. Although these instruments are effective and widely used, mechanical cutting involves direct contact with the treated surface and can generate vibration, noise, debris, and heat. Furthermore, the cutting characteristics of rotary instruments depend on bur geometry, wear, and the mechanical properties of the material being removed.

Laser ablation provides an alternative, non-contact approach in which tissue can be removed through controlled deposition of optical energy.[1–3] Several laser systems have been investigated for dental applications, particularly for soft-tissue procedures and dental hard-tissue ablation. Erbium lasers, including Er:YAG at $2.94\text{ }\mu\text{m}$ and Er,Cr:YSGG at $2.78\text{ }\mu\text{m}$, are effective for ablating enamel and dentine because their wavelengths are strongly absorbed by water and hydroxyapatite.[4] Diode and CO_2 lasers are predominantly used for soft-tissue applications, while Nd:YAG lasers have also been investigated for dental procedures. However, the relatively long pulse durations of conventional dental laser systems can result in heat transfer beyond the irradiated region. When applied to dental hard tissues or restorative materials, this may produce melting, cracking, surface discolouration, compositional changes or a heat-affected zone.[4, 5] Their effectiveness can also vary substantially between materials because absorption depends on wavelength and composition.

Femtosecond lasers represent a distinct approach to laser material removal. They generate ultrashort optical pulses[6] with durations on the order of 10^{-15} s . During each pulse, energy is deposited into the

material over a timescale substantially shorter than that required for significant heat diffusion into the surrounding region. This highly localised energy deposition can therefore remove material while limiting thermal effects outside the irradiated volume. The nonlinear absorption processes associated with the high peak intensities of femtosecond pulses may also permit the ablation of materials that absorb weakly at the laser wavelength under lower-intensity irradiation. These characteristics make femtosecond lasers attractive for precise processing of heterogeneous dental tissues and restorative materials.[1–3, 7–12]

Previous studies have demonstrated femtosecond laser ablation of enamel and dentine. In comparison with longer-pulsed laser systems, femtosecond irradiation can produce well-defined ablation with reduced cracking, melting, and thermal alteration. However, dental restorative materials differ considerably in their composition and microstructure. Resin-based composites contain polymer matrices and inorganic fillers, glass ionomer cements consist of chemically distinct glass and polymeric phases, and dental ceramics have compositions and structures that differ markedly from both. Femtosecond-laser interactions with dental restorative materials have also been reported, but principally as surface modification for adhesion rather than as bulk removal: studies of resin–zirconia bonding, zirconia implant microstructuring, cell–surface interaction, and the surface morphology of lithium disilicate and highly translucent zirconia have each characterised a single material or a small group under parameters chosen for texturing rather than for material removal.[13–16] Complementary optical-penetration models developed for dental hard tissue show that removal behaviour becomes predictable once wavelength-specific absorption is known,[17] but equivalent optical data are not available for most commercial restorative products. Consequently, a single set of laser parameters cannot be assumed to produce equivalent ablation behaviour across different restorative materials. Establishing the ablation threshold, removal efficiency, and resulting surface characteristics for individual material classes is therefore necessary before femtosecond laser processing can be considered for controlled removal of dental restorations.

Dental restorations are broadly classified as direct or indirect.[18, 19] Direct restorative materials, such as resin composites and glass ionomer cements, are placed and set intraorally. Indirect restorations, including ceramic and other laboratory- or CAD/CAM-fabricated restorations, are manufactured outside the prepared cavity before being bonded or cemented to the tooth.[18, 19] Restorations may subsequently require partial or complete removal because of recurrent caries, marginal failure, fracture, wear, discolouration, replacement of the restoration, or the need to gain access to underlying tooth structure.[20, 21] During such procedures, distinguishing and selectively removing restorative material while preserving sound enamel and dentine can be challenging with conventional rotary instruments.[22] Indirect restorations are now predominantly designed and milled under digital control, so the fabrication side of the restorative workflow is already parameterised by material, whereas the removal side is not. A removal process whose parameters can be specified per material, and applied without operator-dependent load or bur wear, is therefore a natural complement to an otherwise digital workflow, provided the required parameters are known.

For femtosecond lasers to provide a useful alternative for restorative-material removal, their interaction with commonly used dental materials must therefore be characterised systematically. Of particular importance are the laser fluence required to initiate ablation, the conditions that provide efficient material removal, the morphology and quality of the processed surface, and the amount of heat transmitted to the dental pulp. These parameters determine whether restorative materials can be removed efficiently without introducing unacceptable thermal or structural damage to the pulpal tissue.

This study investigated femtosecond-laser ablation of commonly used direct and indirect restorative materials. Ablation threshold and efficiency were determined for each material, and optical profilometry and scanning electron microscopy were used to characterise surface morphology. Laser-processed specimens were compared descriptively with specimens prepared using a rotary diamond bur. In addition, intrapulpal temperature changes were measured ex vivo under different processing and cooling conditions. Finally, cavity preparation, composite placement and subsequent restoration removal were demonstrated in a single extracted tooth. The study therefore evaluates material-dependent processing behaviour and ex vivo thermal response; it does not establish clinical safety or autonomous interface-selective removal. Glass ionomer and resin-modified glass ionomer cements and dental amalgam were outside the scope of the present material set.

Methods

Teeth and Ethics approval

Ethical approval

for the use of extracted human teeth in this study was granted through the Human Ethics Office of The Australian National University (H/2024/0941). All methods were performed in accordance with the relevant guidelines and regulations. Informed consent was obtained from all participants and/or their legal guardians.

Intact, healthy human permanent teeth were selected from a collection of extracted teeth and used to study the effects of the femtosecond laser. Remnants of dental pulp tissue were removed. The teeth were sterilized by gamma irradiation (25 kGy; Steritech, Narangba, Qld, Australia) and stored in a dry state. Before laser processing, they were rehydrated in distilled water for 24 h to restore their water content to a fully hydrated state.

Dental materials and preparation of samples

A total of 13 restorative materials were selected. Three independent specimens were prepared for each material ($n = 3$ per material). The individual specimen was treated as the experimental unit. Materials were obtained from their respective manufacturers, used within the expiry date and prepared according to the manufacturers' instructions. Light-curable materials were polymerised with a Radium Plus curing light (SDI, Bayswater, Australia) for the manufacturer-recommended duration. Specimens were prepared

as flat pellets or 2-mm-thick sections using a precision saw (TechCut 4, Allied High Tech Products, USA). Surfaces were polished sequentially with 6, 3 and 1 μm glycol-based polycrystalline diamond suspensions using an AD-5 polishing machine (Allied High Tech Products) and were cleaned before laser processing. Product codes, lot numbers, and shade and translucency designations for every material are listed in Table 1.

Table 1
Restorative materials and manufacturer-reported compositions.

Material	Manufacturer	Composition supplied by manufacturer
Direct restorative materials		
G-ænial Universal Injectable (shade A2)	GC Tokyo, Japan	69% filler, 31% methacrylate monomer, trace pigments, trace photoinitiator
G-ænial Flo X	GC Tokyo, Japan	69% filler by weight (mean 700 nm), 10–25% Bis-MEPP, 5–10% UDMA, 2.5–5% TEGDMA, trace oxides/pigments
G-ænial A'CHORD (shade A2)	GC Tokyo, Japan	Barium glass, dimethacrylate, silicon dioxide, initiator, pigment, stabilizer
everX Flow (shade B)	GC Tokyo, Japan	42–52% Barium glass, trace silicon dioxide, 15–25% Bis-MEPP, 1–10% TEGDMA, 1–10% UDMA, E-glass fibres
SDI Aura	SDI Bayswater, VIC, Australia	75% inorganic fillers, 25% cross-linked dimethacrylate resins (UDMA, Bis-GMA, TEGDMA blend)
SDI Conseal-F	SDI Bayswater, VIC, Australia	UDMA-based multifunctional methacrylate resin system containing 7% submicron fluoride-releasing filler (particle size: 0.04 µm).
SDI Wave	SDI Bayswater, VIC, Australia	56% inorganic fillers, 43% cross-linked dimethacrylate resins (UDMA, TEGDMA blend)
Indirect restorative materials		
Zirconia	Kuraray Noritake Dental Tokyo, Japan	87–92% zirconium dioxide (ZrO ₂) & hafnium oxide (HfO ₂), 8–11% yttrium oxide (Y ₂ O ₃), 0–2% other oxides and pigments
EMP LTBL1	Ivoclar Vivadent Schaan, Liechtenstein	60–65% SiO ₂ , 16–20% Al ₂ O ₃ , 10–14% K ₂ O, 3.5–6.5% Na ₂ O, 0.5–7.0% other oxides, 0.2–1.0 pigments
EMP C14L	Ivoclar Vivadent Schaan, Liechtenstein	60–65% SiO ₂ , 16–20% Al ₂ O ₃ , 10–14% K ₂ O, 3.5–6.5% Na ₂ O, 0.5–7.0% other oxides, 0.2–1.0 pigments
IPS e.max CAD	Ivoclar Vivadent Schaan, Liechtenstein	Lithium disilicate SiO ₂ >57%, Li ₂ O, K ₂ O, P ₂ O ₅ , ZrO ₂ , ZnO, Al ₂ O ₃ , MgO, pigments

Material	Manufacturer	Composition supplied by manufacturer
ProCAD E100	Ivoclar Vivadent Schaan, Liechtenstein	SiO ₂ 60.0% – 64.0%, Al ₂ O ₃ 16.0% – 20.0%, K ₂ O 10.0% – 14.0% and Na ₂ O 3.5% – 6.5%
VITA CAD-Temp	VITA Zahnfabrik, Bad Sackingen, Germany	80–90% poly(methyl methacrylate), 10–20% filler particles, 1–3% Colour pigments and Stabilizers, < 1% cross-linking agents

Femtosecond laser experimental setup

The laser operated at a wavelength of 1030 nm with a pulse duration of 275 fs, a repetition rate of 100 kHz, and a maximum pulse energy of 400 µJ. For ablation rate testing, the laser pulses were focused on the sample using a quasi-telecentric f-Theta scanning lens with a 540 mm working distance (S4LFT1420/449, Sill Optics GmbH, Germany). The raw beam had a focal spot diameter of ~ 96 µm (FWHM), which was reduced to 48 µm (FWHM) using a GBE02-C 2X Achromatic Galilean Beam Expander (Thor Labs Inc.) to achieve fluences up to 20 J/cm². A 10-facet polygon mirror (Precision Laser Scanning Inc., USA) and a galvanometer scanner (Cambridge Technology Inc., USA) were used to scan the beam across the dental materials in a single-pulse scanning regime, ensuring well-separated laser spots on the target surface. The fast (vertical) raster scanning was achieved by rotating the polygon scanner at 1,000 rpm, resulting in a beam velocity of ~ 86 m/s at the focal point. The galvo scanner provided lateral (horizontal) stepping with a beam velocity of ~ 20 m/s and a step size of 5 µm on the surface. Laser fluence was adjusted by attenuating the laser power. Custom LabVIEW software controlled key scanning parameters, including speed, inter-shot and inter-scan line distances, and 2D scanning patterns. All experiments were conducted in ambient air and at room temperature. A schematic of the experimental setup is shown in Fig. 1a.

Determination of ablation rates and efficiencies

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Femtosecond laser ablation was used to create grooves in dental materials. A series of grooves was ablated on each sample, each approximately 600 µm wide. Each groove was subjected to a different fluence within the range of 0–20 J/cm². To maintain consistency, equal energy deposition across different experimental conditions was ensured.

Rotary-instrument comparison

For qualitative comparison, curved grooves were produced in separate specimens using a high-speed handpiece (NSK S-Max M900L, NSK, Japan) with an ISO 010 straight diamond bur under integrated air-water spray. Because operator load, cutting duration and bur wear were not instrument-controlled, this experiment was interpreted as a descriptive morphological comparison rather than a quantitative performance comparison.

Surface characterisation

Optical microscopy and profilometry. Surface features were assessed for structural changes with a reflection mode optical microscope (Olympus BX-60 with bright and dark field) at 5, 10, 20, 50, and 100x magnifications and surface texture, roughness and depth of the ablated surface were recorded by using an optical profiler (Veeco Wyko NT9100, Bruker, USA) with a 5x and 20x objective and a 0.55x field-of-view multiplier. The vertical scanning interferometry (VSI) mode was used.

Scanning electron microscopy. Specimens were sputter-coated with platinum at 15 mA for 3 min. Untreated, laser-ablated and rotary-cut surfaces were examined using a Zeiss UltraPlus field-emission scanning electron microscope at 500x, 1,000x, 10,000x, 15,000x and 20,000x magnification, with an accelerating voltage of 3 kV and either the in-lens or SE2 detector. The SEM assessment was qualitative and limited to observable surface morphology.

Simulation of cavity preparation and removal of restoration

A sound extracted human canine was sterilised by gamma irradiation (25 kGy; Steritech, Australia). A nominal $2 \times 2 \times 2 \text{ mm}^3$ buccal cavity was prepared using a fluence of 6 J cm^{-2} , wavelength of 1030 nm, pulse duration of 275 fs, repetition rate of 100 kHz, focal-spot diameter of $45.5 \text{ }\mu\text{m}$ (FWHM) and 2,000 scans.[2] G-ænial A'CHORD A2 composite was placed using the manufacturer-recommended primer, adhesive and light-curing procedures. The restoration was then removed with the femtosecond laser. The final cavity measured approximately $2.1 \times 2.1 \times 2.1 \text{ mm}^3$, representing a nominal increase of 0.1 mm in each overall dimension (approximately 0.05 mm per opposing wall), intended to remove residual adhesive at the cavity interface.

Ex vivo intrapulpal temperature measurements

Three gamma-irradiated, caries-free extracted human molars were prepared for each of the direct- and indirect-material groups (Fig. 1d). For the direct-material group, standardised $4 \times 4 \times 4 \text{ mm}$ cavities were prepared with an NSK S-Max M900L handpiece and ISO 010 straight diamond bur. Cavities were etched with phosphoric acid for 30 s, rinsed and gently air-dried. Scotchbond Universal Plus Adhesive (3M) was applied, air-thinned and light-cured according to the manufacturer's instructions. Cavities were restored incrementally with G-ænial A'CHORD A2 composite, with each increment light-cured before placement of the next.

For the indirect-material group, tooth preparations were produced using the same handpiece with a crown-preparation bur. Corresponding IPS e.max CAD restorations were milled, crystallised and fired according to the manufacturer's instructions, then cemented with Panavia V5 resin cement.

To verify access to the pulp chamber and confirm thermocouple position, two size 35 endodontic hand files were inserted through access openings prepared below the cemento-enamel junction, and periapical radiographs were obtained before thermocouple placement (Fig. 1e,f).

The apparatus for the thermal experiments (Fig. 1b) included a custom-built isothermal water bath with a PID temperature controller (Red Lion PXU-24V), an 8-channel temperature data logger (Picolog TC08) and a pressure-regulated cooling air supply with a 2 mm diameter nozzle positioned 35 mm from the top of the tooth. Thermal experiments were done with the laser spot size set at 48 μm (FWHM), and pulse energy set at 140 μJ for average optical powers 3W, 5W, 7W, and 15W, while maintaining a near optimal fluence ($7\sim 8\text{ J/cm}^2$) for material ablation. Optical power on the tooth was adjusted by changing the repetition rate whilst keeping the spot size and pulse energy fixed.

After filling the pulp chamber of the tooth with water using a square tip 28-gauge disposable needle, a 0.5 mm diameter K-type stainless-steel sheathed thermocouple (RS PRO) was inserted into one of the 0.55 mm diameter access holes. To reduce absorption of infrared light from the laser, the thermocouple sheath was electroplated with a thin layer of gold. Additional thermocouples were used to monitor the ambient air temperature and temperature of the water bath. The tooth was mounted with roots in the bath using 1 mm thick translucent silicon rubber tubing stretched over a 1.5 mm-thick acrylic frame which was clamped onto the top of the water bath using four M3 studs with nylon nuts. A 5 mm diameter hole was punched into the silicon rubber, into which the tooth was inserted, holding the tooth in position and forming a seal around the tooth which prevented the cooling air from blowing water out of the isothermal bath (see Fig. 1c).

For each thermal test, the galvo scanner was set to ablate a rectangle on the dental material attached to the tooth. Temperature data sampling was done at a rate of one sample per second. The water bath was set a constant temperature of 37°C. The ambient air temperature was 24°C. Each experiment was started after the temperature inside the tooth had stabilised. For tests where cooling air was used, both the laser and cooling air were switched on and off simultaneously (except for one set of tests where the cooling air was left on after switching the laser off to examine its effect on the cooling rate of the tooth).

Results

Femtosecond-laser ablation of restorative materials

Grooves were created using the femtosecond laser across the range of direct and indirect dental materials by varying the laser fluence and following the procedure explained above. Representative examples, including scale markers, are presented in Fig. 2a. The results demonstrate that femtosecond laser processing effectively produces well-defined, sharp, and smooth grooves on the dental material samples, highlighting the precision of this technique. Optical visual examination confirmed that there was no apparent colour change in the materials after ablation, indicating minimal thermal effects and preserving the original material properties. Additionally, no structural damage, such as microcracks or delamination, was observed at the bottom of the cavities, further supporting the gentle and controlled nature of femtosecond laser ablation.

Ablation thresholds, rates, and efficiencies

The ablation thresholds, rates, and efficiencies of femtosecond laser processing were systematically measured, calculated, and analysed to provide valuable insights into the interaction between fs laser pulses and the selected dental materials. These findings support the optimisation of laser settings for precise and efficient ablation.

For both indirect and direct dental materials, the relationships between ablation depth per pulse ($\mu\text{m}/\text{pulse}$), ablation efficiency ($\text{mm}^3/\text{min}/\text{W}$), and volumetric removal rate (mm^3/min) as functions of laser fluence (energy per unit area) were examined. The resulting data, presented in Table 2 and Fig. 3, reveal distinct trends between material types, with average values for indirect and direct groups indicated by green and orange lines, respectively. TempCAD, an indirect material with a composition resembling direct materials, is represented separately in blue for clarity.

The full range of ablation thresholds among tested materials extended from $0.6 \text{ J}/\text{cm}^2$ (SDI Conseal F) to $5.5 \text{ J}/\text{cm}^2$ (ProCAD). Specific optimal fluence ranges for each material studied are detailed in Table 2.

In terms of ablation rates, the findings confirm the presence of two distinct material categories. As laser fluence increases, ablation depth per pulse follows a generally proportional trend. The ablation threshold, defined as the minimum fluence required to initiate material removal, was found to differ significantly between material types. Indirect materials generally required a fluence above $2.0 \text{ J}/\text{cm}^2$, whereas direct materials exhibited ablation thresholds as low as $0.6 \text{ J}/\text{cm}^2$. This distinction reinforces the categorization of materials into indirect and direct groups based on their ablation behaviour.

Analysis of ablation efficiency revealed optimal energy ranges for each material type. As shown in Fig. 3b, the direct material group demonstrated peak efficiency at a fluence of approximately $6.0\text{--}7.0 \text{ J}/\text{cm}^2$. In contrast, indirect materials reached their optimal ablation efficiency at fluences above $10.0 \text{ J}/\text{cm}^2$, followed by only marginal increases with further fluence. These trends suggest that direct materials favor lower fluence for efficient ablation, whereas indirect materials require higher energy input to reach their optimal performance range.

TempCAD, although classified as an indirect restorative material, demonstrated ablation behaviour closely aligned with that of direct materials. Its trend, marked by a blue line, supports its compositional similarity and highlights its unique positioning between the two material categories.

Table 2

Ablation threshold, optimal ablation efficiencies, and optimal ablation fluence for the studied indirect and direct dental materials under femtosecond laser irradiation at 1030 nm.

Class	Material	Ablation threshold (J cm⁻²)	Maximum efficiency (mm³ min⁻¹ W⁻¹)	Optimal fluence (J cm⁻²)
Indirect	Zirconia	1.2 ± 0.1	0.67 ± 0.01	5.5–11.5
Indirect	EMP LTBL1	0.9 ± 0.1	0.35 ± 0.01	> 13.0
Indirect	EMP C14L	2.1 ± 0.1	0.36 ± 0.01	11.2 ± 0.1
Indirect	IPS e.max CAD	3.5 ± 0.1	0.37 ± 0.01	19.2 ± 0.1
Indirect	ProCAD E100	5.5 ± 0.1	0.30 ± 0.01	10.0–19.2
Indirect	VITA CAD-Temp	1.2 ± 0.1	2.13 ± 0.01	6.6 ± 0.1
Direct	G-ænial Universal Injectable A2	1.0 ± 0.1	3.37 ± 0.01	6.6 ± 0.1
Direct	G-ænial Flo X	1.9 ± 0.1	2.26 ± 0.01	11.2 ± 0.1
Direct	G-ænial A'CHORD A2	1.8 ± 0.1	2.92 ± 0.01	7.8 ± 0.1
Direct	everX Flow	2.0 ± 0.1	2.95 ± 0.01	8.9 ± 0.1
Direct	SDI Conseal-F	0.6 ± 0.1	2.78 ± 0.01	2.0–7.0
Direct	SDI Aura	0.7 ± 0.1	3.03 ± 0.01	2.5–6.5
Direct	SDI Wave	0.7 ± 0.1	3.33 ± 0.01	2.5–7.5

Surface morphology and roughness

The ablation profiles and optimal regimes of the dental materials were assessed using a profilometer to obtain corresponding 3D surface images and contour maps (Fig. 2b,c). The resulting ablation grooves exhibited clean, well-defined profiles with no evidence of cracking or thermal damage, such as melting. The contour lines confirmed precise material removal with minimal surface irregularities. Consistent with the greater depth per pulse for direct materials in Fig. 3a, the depth of the channels shown in Fig. 2b for direct materials was much greater than that for indirect materials for a similar processing time (Fig. 2b,c, centre column). As is also shown in Fig. 2b,c, scanning electron microscopy (SEM) at 15,000X magnification revealed consistent surface roughness and preserved microstructural integrity. This can be seen by comparing the images for the non-lased material surfaces (Fig. 2b,c, left column) with the laser processed surfaces from the bottom of the ablated channel (Fig. 2b,c, right column). In many cases (e.g. zirconia) the ablated surface shows an increased roughness compared to the surface material prior to laser processing. Differences in surface roughness may also be related to final surface finishing processes employed by the manufacturer of the material samples.

Surface features of the dental materials were examined further using high-resolution SEM imaging (Fig. 4). At 20,000× magnification, both direct and indirect dental materials treated by femtosecond laser ablation exhibited material-dependent microstructural patterns, with characteristic feature sizes ranging from approximately 1 to 3 µm. Although the detailed morphology varied among materials, all laser-ablated surfaces displayed relatively regular and well-defined surface structures. Among the materials examined, zirconia exhibited the finest surface features, whereas EverX Flow showed the roughest and most irregular morphology. Importantly, no obvious cracking, melting, or other signs of thermal damage were observed on the femtosecond-laser-ablated surfaces (Fig. 4, middle column).

At 20,000× magnification, laser-ablated surfaces showed material-dependent features of approximately 1–3 µm (Fig. 4). No obvious surface cracking or gross melting was observed in the imaged fields. Zirconia showed comparatively fine surface features, whereas everX Flow showed a rougher and more irregular morphology. These SEM observations are limited to the sampled surface fields and do not exclude subsurface cracks, phase transformation or chemical modification. The rougher and more irregular morphology of everX Flow is consistent with its E-glass fibre reinforcement, since fibres and surrounding matrix ablate at different rates and leave protruding or recessed fibre ends; the same reinforcement plausibly underlies its comparatively high threshold of 2.0 J cm⁻².

Rotary-cut specimens showed greater smearing, grooving and adherent debris in the selected SEM fields (Fig. 4). Because rotary load, cutting time, bur wear and coolant delivery were not instrument-controlled, the images support only a qualitative comparison and not a quantitative conclusion that one method causes less damage.

Among the indirect materials, rotary-cut zirconia showed a debris-covered surface, while ProCAD E100 and EMP C14L showed grooves and torn or fractured-appearing features. VITA CAD-Temp showed localised deformation and debris. Laser-ablated surfaces showed different, generally more regular morphologies, but chemical and phase integrity were not measured.

Overall, femtosecond-laser and rotary processing produced visibly different surface morphologies. Further quantitative roughness, chemical, phase and mechanical analyses are required before concluding that laser processing preserves material integrity.

Proof-of-concept restoration workflow

In this single-tooth proof of concept, a nominal 2 × 2 × 2 mm³ buccal cavity was prepared at 6 J cm⁻², 1030 nm, 275 fs, 100 kHz and a 45.5 µm focal-spot diameter (Fig. 5). G-ænial A'CHORD A2 composite was placed with an adhesive protocol and subsequently removed with the femtosecond laser. The final cavity measured approximately 2.1 × 2.1 × 2.1 mm³.

The increase from 2.0 to 2.1 mm in each overall dimension corresponds to approximately 0.05 mm per opposing wall and an additional nominal volume of 1.261 mm³. The experiment demonstrates technical

feasibility in one extracted tooth; it does not establish interface-selective removal, complete adhesive elimination or preservation of subsurface tooth structure.

Ex vivo intrapulpal temperature response

Figure 6. Ex vivo intrapulpal temperature change during and after 30 s of femtosecond-laser irradiation at different incident powers, with and without air cooling. (a) G-ænial A'CHORD A2, (b) IPS e.max CAD and (c) human enamel. All panels are drawn at the same size; the ordinate range differs between rows. The red region denotes an intrapulpal temperature increase above $\Delta T = 5.5^{\circ}\text{C}$; individual curves represent tooth-level measurements.

To evaluate potential thermal effects on pulpal tissue in this ex vivo model, the 5.5°C temperature-rise value reported by Zach and Cohen, and commonly used as a thermal safety reference in dental research, was adopted as the operational threshold for the present study.[23–25] Accordingly, test conditions producing a peak intrapulpal temperature rise (ΔT_{max}) of $\leq 5.5^{\circ}\text{C}$ were classified as remaining within the adopted threshold, whereas those producing a ΔT_{max} of $> 5.5^{\circ}\text{C}$ were classified as exceeding it.

Figure 6 shows tooth-level temperature traces during 30 s of irradiation and subsequent cooling. Orange curves denote no additional cooling, light-blue curves denote air cooling during irradiation, and dark-blue curves denote air cooling continued after irradiation. Replication varied by condition, and the curves should therefore be interpreted descriptively.

In the direct material group (Fig. 6a), air cooling reduced the intrapulpal temperature rise at all tested laser powers. Maintaining air flow after laser cessation further accelerated the cooling process, suggesting that continued air cooling may help reduce tooth temperature before subsequent irradiation cycles and thereby improve procedural safety, as shown by the dark blue curves in Fig. 6a. Although 7 W was the highest power that remained acceptable for direct material removal under air-cooling conditions, incidental enamel removal may occur during clinical composite removal. Since 5 W remained within the safe thermal range for enamel, as shown in Fig. 6c, this setting appears to be the most clinically relevant and conservative parameter under the present experimental conditions. At 5 W, the ablation rate was approximately $14 \text{ mm}^3/\text{min}$. Therefore, an occlusal restoration measuring $2 \times 2 \times 2 \text{ mm}$ could be largely removed within a single 30 s irradiation cycle, allowing the clinician to reassess the amount of residual material before further treatment.

For the indirect material group, greater variation was observed among tooth samples compared with the direct material and enamel groups. This variation is likely attributable to multiple factors, including differences in the distance between the irradiated surface and the pulpal chamber, variations in tooth anatomy, and differences in the thickness of the resin cement layer. Under the present experimental conditions, 5 W was the highest power that remained within the accepted thermal safety range for vital teeth. This suggests that 5 W may be suitable for the slow and controlled removal of residual ceramic from restorations such as veneers, crowns, or onlays in intraoral applications. A combined approach may also be clinically advantageous, in which conventional rotary instruments are used for bulk material

removal, followed by femtosecond laser irradiation for precise removal of the final residual layer. Such a strategy could improve efficiency while reducing the risk of damaging the underlying tooth structure.

Because enamel may be exposed either unintentionally or intentionally during the removal of dental materials, the thermal response of enamel was also investigated at 5 W and 7 W, as shown in Fig. 6c. Both power settings produced intrapulpal temperature rises that were within the accepted thermal safety range under the present experimental conditions. The maximum temperature increases observed with and without air cooling were similar in magnitude and remained mostly below the critical threshold. These results further support the safe use of 5 W irradiation on the tooth surface for a processing time of 30 s. Although 7 W also appeared thermally acceptable for enamel in this study, 5 W remains the more conservative and clinically relevant setting when vital teeth are involved.

Overall, under the present experimental conditions, 5 W appears to be the most clinically relevant and conservative operating power for intraoral applications involving vital teeth. This setting enabled effective material removal while maintaining the intrapulpal temperature rise within the accepted safety threshold. As 5 W represents the thermally safe power limit of the current cooling system, the maximum ablation speeds achievable at this power were evaluated to further assess the material-removal capability of femtosecond (fs) laser ablation. As summarised in Table 2, direct restorative materials exhibited a mean ablation rate of 14.74 mm³/min, whereas indirect restorative materials exhibited a mean rate of 2.05 mm³/min.

At the mean rate observed for direct materials, removing a 1 mm³ restoration (e.g., 1 mm × 1 mm × 1 mm) would require approximately 4 seconds. These findings provide a practical reference for selecting material-specific laser parameters in clinical dental applications.

For indirect restorations, fs lasers may be particularly suitable for ablating the final thin layer overlying a vital tooth because of their high precision. Additional air cooling, particularly when continued after laser irradiation has ceased, further improved thermal control and may enhance procedural safety during repeated irradiation cycles.

Direct comparison with earlier ultrashort-pulsed-laser studies is limited. Ablation thresholds and rates have been reported for oral hard tissue together with a small number of restorative materials,[26] ablation efficiency has been examined for dental hard tissue and restorative materials,[27] post-irradiation properties have been characterised for zirconia, lithium disilicate glass ceramic and a hybrid ceramic,[28] and laser-assisted direct restoration of enamel has been demonstrated with a 3D-printable biocomposite.[29] Each of these studies covered one or a few materials under its own irradiation conditions, so the values reported are not directly comparable with those obtained here.

For root canal-treated teeth and extraoral applications, where pulpal thermal injury is not the primary limiting factor, power selection may be guided more strongly by procedural efficiency. Higher laser powers may therefore be appropriate for extraoral procedures, such as the adjustment or milling of ceramic crowns. In these applications, fs laser processing may offer a precise, consistent, and

potentially cleaner alternative to conventional rotary milling, particularly when controlled material removal is required. Further studies should evaluate a broader range of dental materials, restoration geometries, and tooth types to better reflect clinical variability.

Optimisation of the cooling system should also be investigated. Increasing the airflow rate, lowering the air temperature, modifying the nozzle geometry, or adjusting the nozzle-to-surface distance may provide more effective cooling and further improve thermal safety during laser treatment.

Conclusion

In this study, femtosecond (fs) laser irradiation enabled the controlled and precise ablation of a broad range of direct and indirect dental restorative materials while producing limited collateral damage. Compared with conventional rotary instrumentation, fs laser processing generated cleaner, more clearly defined surfaces with less cracking, melting, debris accumulation, and smear-layer formation. Ablation performance was strongly material dependent, however, indicating that material-specific operating parameters are required to balance removal efficiency with thermal safety.

The proof-of-concept restoration-replacement workflow demonstrated the technical feasibility of sequential fs laser cavity preparation, conventional composite placement, and subsequent laser removal in a single extracted tooth. Under the present experimental conditions, 5 W was identified as the most clinically relevant and conservative operating power for intraoral applications involving vital teeth, providing effective material removal while maintaining the intrapulpal temperature rise within the accepted safety threshold. At this power, direct and indirect restorative materials exhibited mean ablation rates of 14.74 and 2.05 mm³/min, respectively. These findings indicate that fs laser ablation can achieve clinically meaningful removal rates, although both ablation efficiency and thermal response vary substantially among material classes. Air cooling improved thermal control, particularly for direct restorative materials, and its continuation after laser cessation may further reduce heat accumulation during repeated irradiation cycles. Indirect materials generally exhibited narrower thermal safety margins and a weaker response to cooling, highlighting the need for careful parameter selection.

Collectively, these findings establish fs laser ablation as a promising platform for selective and minimally invasive restorative dentistry. For procedures involving vital teeth, further optimisation should prioritise thermal management while maintaining clinically practical removal rates. For root canal-treated teeth and extraoral applications, where pulpal thermal injury is not the principal constraint, higher laser powers may be used to improve processing efficiency. Fs laser processing may therefore also offer a precise, consistent, and potentially cleaner alternative to conventional rotary methods for the extraoral adjustment or milling of ceramic restorations.

Further research should optimise laser power, scanning strategy, and cooling-system design, including airflow rate, air temperature, nozzle geometry, and nozzle-to-surface distance. Validation across a wider range of restorative materials, restoration geometries, and tooth types is also required. Ultimately, the development of clinically practical laser-delivery systems and real-time feedback mechanisms for

material recognition, selective removal, and thermal control will be essential for translating this technology into routine clinical use.

Declarations

Competing interests

O.Z. and L.J.W. are employees of Dentroid Pty Ltd, which provided funding to the project. The remaining authors declare no competing interests.

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Author Contribution

J.P.Y.: conceptualisation, methodology, validation, investigation, resources, formal analysis, writing—original draft, and writing—review and editing. K.M.: investigation, formal analysis, visualisation, and writing—review and editing. P.W.: methodology, investigation, formal analysis, and writing—review and editing. S.M.: software, conceptualisation, and methodology. A.V.R.: conceptualisation, methodology, and writing—review and editing. H.S.: resources, formal analysis and writing—review and editing. L.J.W.: conceptualisation, methodology, resources, and writing—review and editing. O.Z.: resources, writing—review and editing. L.R.: conceptualisation, methodology, validation, investigation, resources, formal analysis, supervision, project administration, funding acquisition, and writing—review and editing.

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Data Availability

The data that support the findings of this study are available from the corresponding author upon the datasets generated during and/or analysed during the current study are available from the corresponding author on reasonable request.

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Figures

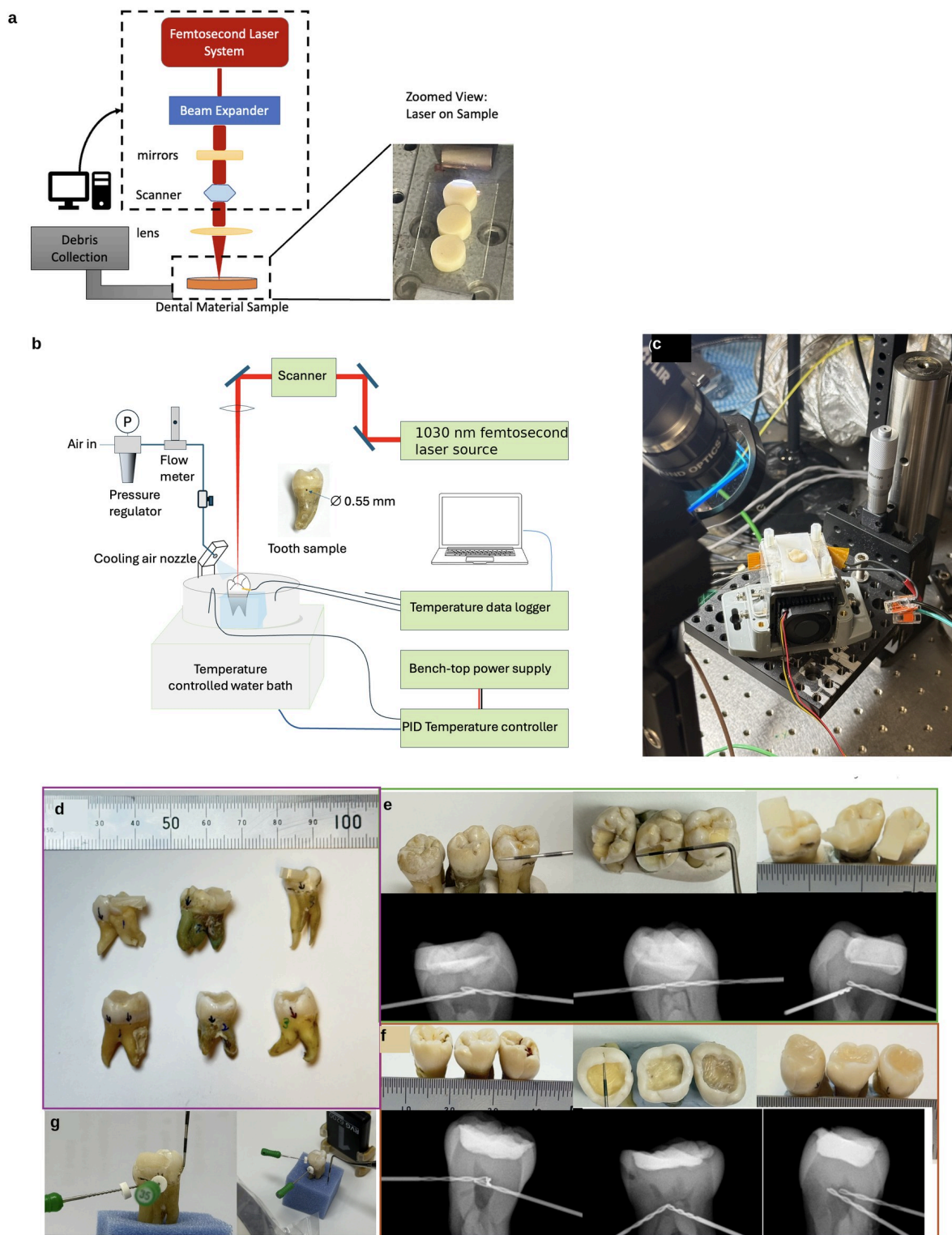


Figure 1

Experimental configurations and specimen preparation. (a) Femtosecond-laser processing and beam-delivery arrangement used for the ablation-rate measurements, with an inset showing the beam incident on a dental-material specimen. (b) Schematic of the ex vivo thermal-test apparatus. (c) Photograph of the assembled thermal-test setup. (d) Extracted molar specimens, with IPS e.max CAD restorations in the upper row and G-ænial A'CHORD A2 restorations in the lower row. (e,f) Clinical photographs and

periapical radiographs showing thermocouple positioning within the pulp chamber. (g) Size 35 endodontic hand file used to verify instrument position.

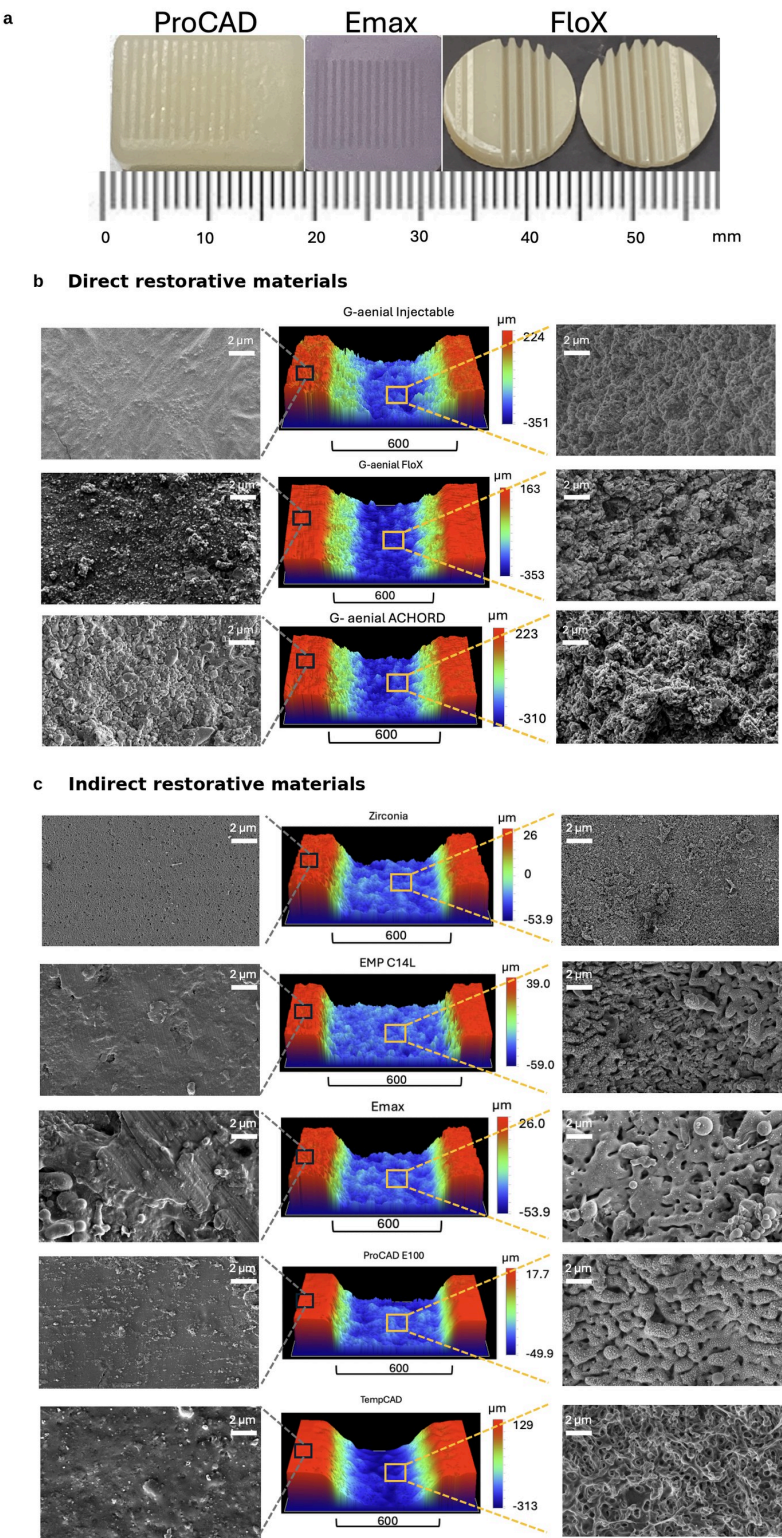


Figure 2

Appearance and topography of femtosecond-laser-processed restorative materials. (a) Representative specimens after processing at a range of fluences (ProCAD E100, IPS e.max CAD and G-aenial Flo X);

scale in mm. (b) Direct and (c) indirect restorative materials. Left: untreated surface; centre: three-dimensional profilometry of the ablated groove floor; right: laser-processed surface.

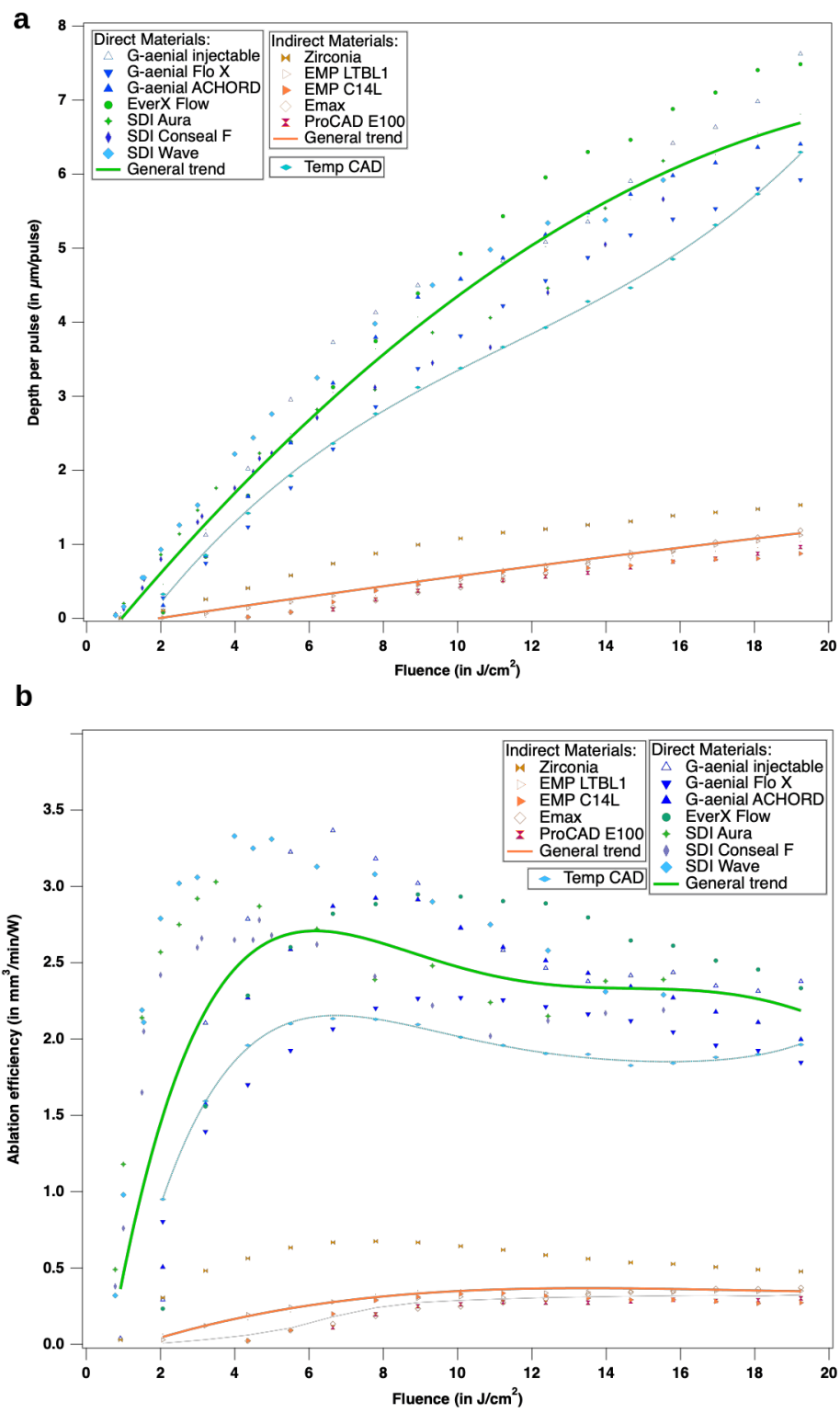


Figure 3

Femtosecond-laser processing of restorative materials: (a) ablation depth per pulse and (b) ablation efficiency as functions of fluence. Curves are descriptive fitted trends; markers show specimen

measurements.

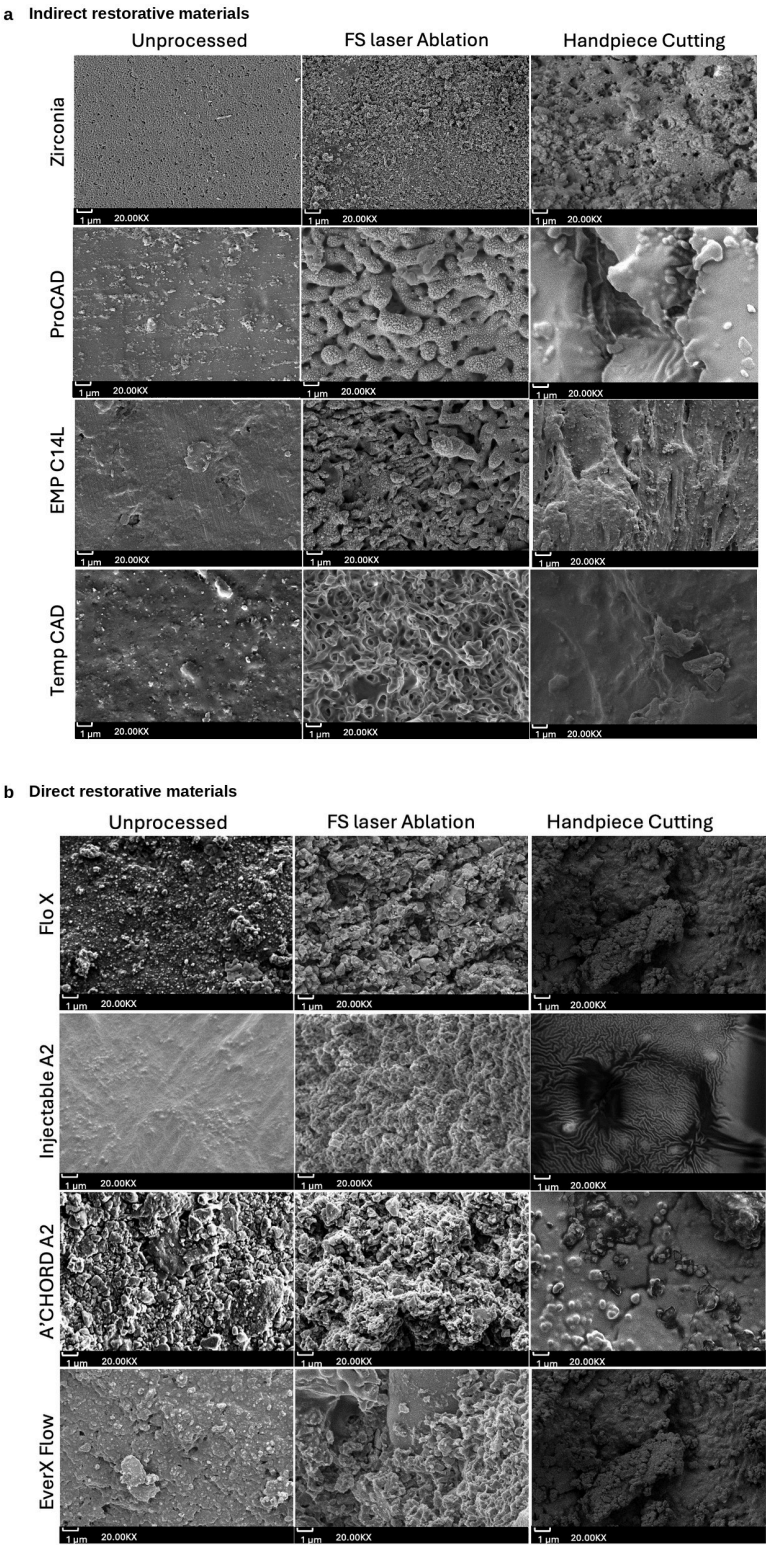


Figure 4

Restorative materials imaged by SEM at 20,000×. (a) Indirect materials: zirconia, ProCAD E100, EMP C14L and VITA CAD-Temp. (b) Direct materials: G-ænial Flo X, G-ænial Universal Injectable A2, G-ænial

An intact Canine Buccal surface

Preparing a cavity into a tooth with FS laser
2 x 2 x 2 mm³ cavity

Performing dental filling
G aenial A'CHORD A2

Removing previous material with FS laser
2.1 x 2.1 x 2.1 mm³ cavity

Bench-top proof-of-concept sequence in an extracted human canine: femtosecond-laser cavity preparation, composite placement and subsequent femtosecond-laser removal



Ex vivo intrapulpal temperature change during and after 30 s of femtosecond-laser irradiation at different incident powers, with and without air cooling. (a) G-ænial A'CHORD A2, (b) IPS e.max CAD and (c) human enamel. All panels are drawn at the same size; the ordinate range differs between rows. The red region denotes an intrapulpal temperature increase above $\Delta T = 5.5\text{ }^{\circ}\text{C}$; individual curves represent tooth-level measurements.

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