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Flexural Properties of Bioactive Restoratives in Cariogenic Environments

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Clinical Relevance

The strength of some bioactive materials can be compromised by cariogenic challenges. This may impact the clinical longevity of restorations, especially in stress-bearing areas.

SUMMARY

This study determined the mechanical performance of bioactive restoratives in cariogenic environments and compared the flexural properties of various bioactive materials. The materials evaluated included a conventional resin-based composite (Filtek Z350 [FZ]) and 3 bioactive restoratives, namely an alkasite (Cention N [CN]), a giomer (Beautifil-bulk Restorative [BB]), and an enhanced resin-modified glass ionomer (Activa Bioactive Restorative [AV]). Beam-shaped specimens (12 x 2 x 2 mm) were produced, randomly allocated to 4 groups (n=10), and conditioned in deionized solution, remineralizing solution, demineralizing solution (DE), or pH cycled for 14 days at 37°C. After conditioning/pH cycling, the specimens were subjected to 3-point flexural testing. Flexural data were subjected to statistical analysis using analysis of variance or Tukey's test (α=0.05). Mean flexural modulus and strength ranged from 3.54 ± 0.33 to 7.44 ± 0.28 GPa, and 87.07 ± 8.99 to 123.54 ± 12.37 MPa, respectively. While the flexural modulus of the bioactive restoratives was not affected by cariogenic/acidic conditions, flexural strength usually decreased, with the exception of CN. The strength of BB was significantly reduced by DE and

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pH cycling, while that of AV was lowered by DE. For all conditioning mediums, AV had a significantly lower modulus than the other materials. Apart from conditioning in DE, where differences in flexural strength was insignificant, FZ and AV were generally significantly stronger than BB and CN. The effect of cariogenic environments on flexural strength was found to be material dependent, and aside from the alkasite material (CN), cariogenic conditions were observed to significantly decrease the strength of bioactive restoratives.

INTRODUCTION

Resin-based composites (RBCs) have become the restorative material of choice in both anterior and posterior teeth, due to their excellent aesthetics, ability to bond to the tooth structure, and conservative cavity preparations.[1] Despite the advances made in photoinitiator, polymer, and filler technology, RBCs are still limited by polymerization shrinkage/contraction stress, incomplete monomer-to-polymer conversion, and limited toughness.[2] Polymerization shrinkage and contraction stress had been associated with several negative clinical outcomes, including cuspal deflection, crack propagation of teeth, decreased bond strength, internal and marginal gap formation, marginal leakage, and secondary caries.[3] Accordingly, systematic reviews of clinical trials have determined secondary caries and composite fracture to be the most common reasons for RBC restoration failure.[4-5] Alvanforoush and others compared the failure rates of direct RBC restorations in posterior teeth over 2 decades and presented some interesting discoveries.[6] Although the failure rates for the years 1995–2005 (10.59%) and 2006–2016 (13.13%) were comparable, the causes of failure differed. In 1995–2005, the reasons for failure were primarily secondary caries (29.47%) and composite fracture (28.84%), with few tooth fractures (3.45%). In 2006–2016, secondary caries (25.68%) decreased slightly, but the incidence of composite fracture (39.07%) and tooth fracture (23.76%) increased considerably. The authors ascribed their findings to the growing use of RBCs for extensive restorations and material variations.[6]

Secondary or recurrent caries (tooth decay at the margins of restorations) accounts for up to 59% of direct restoration replacements.[7] Secondary caries with RBCs can be attributed to patient (caries-risk profile), operator (placement procedures), and material-related factors.[8] More specifically, RBCs appear to favor cariogenic bacterial growth due to their surface characteristics, the components released, and the lack of antibacterial activity.[8] Bioactive restorative materials that replace tooth tissues and possess “therapeutic functions” have been developed to minimize secondary caries.[9] The concept of bioactive restorative materials is not entirely new, and materials were available for several decades in the form of fluoride-releasing materials such as glass ionomer cements (GICs).[10] Commercially available bioactive restorative materials are usually hybrids of RBCs and GICs containing antimicrobials as well as calcium phosphate, silicate, and/or aluminate.[9,10] Their “therapeutic functions” include the potential for suppressing biofilms and acid production, deterring proteins, diminishing secondary caries, and neutralizing acids, as well as replacing lost minerals through ion and other chemical release.[9]

Information pertaining to the mechanical characteristics of contemporary bioactive restorative materials is still limited. Findings have been equivocal, with studies reporting both comparable and lower strengths when compared with conventional RBCs.[11-13] The disparity can be attributed to differences in the materials evaluated, testing methodology, and the conditioning mediums used.[13] Ion release from bioactive restorative materials intensifies under acidic conditions and was associated with significant degradation of calcium-based glasses.[14] Cariogenic environments may well compromise the strength of bioactive restorative materials, leading to material fracture and early restoration failure. Moreover, some bioactive materials are capable of forming apatite-like phases in saliva that might enhance their mechanical properties.[14,15]

The effect of cariogenic challenges on the mechanical properties of bioactive restorative materials has not been explored. This is clinically pertinent, as these materials are advocated in patients with a high caries risk and are often utilized in large cavities.[16] Thus, the objectives of this study were to determine the mechanical performance of bioactive restorative materials in cariogenic environments and compare the flexural properties of various bioactive materials. The null hypotheses were as follows: (1) cariogenic challenges do not degrade the flexural modulus and strength of bioactive restorative materials, (2) environmental calcium phosphate does not enhance flexural properties, and (3) there is no difference in the flexural properties between the various bioactive materials.

METHODS AND MATERIALS

Specimen Preparation and Conditioning

The materials evaluated included a conventional RBC (Filtek Z350 [FZ]; 3M Oral Care, St. Paul, MN, USA) and 3 bioactive restorative materials, namely an alkasite (Cention N [CN]; Ivoclar Vivadent Inc, Amherst, NY, USA), a giomer (Beautifil-bulk Restorative [BB]; Shofu, Kyoto, Japan), and an enhanced resin-modified glass ionomer (RMGIC; Activa Bioactive Restorative [AV]; Pulpdent, Watertown, MA, USA). The manufacturers and technical profiles of the materials are listed in Table 1. While the alkasite and giomer materials are essentially resin-based composites with alkaline and prereacted glass ionomer fillers, respectively, the "enhanced" RMGIC was augmented with other bioactive glasses. The minimum sample size was determined using the G*power software (version 3.1.9.4)[17] based on the analysis of variance test with an effect size of 0.5,13 alpha error of 0.05, and power of 80% for 16 material-medium combinations. Forty bar-shaped specimens (12 x 2 x 2 mm) of the various materials were fabricated for the mini-flexural test (MFT) using custom-made stainless-steel molds.[13] The materials were mixed according to the manufacturers' instructions (where applicable) and/or placed into the molds in a single increment. The material-filled molds were compacted between 2 polyester strips with glass slides to remove excess material. All materials were light polymerized with 2 overlapping 10-second irradiations from the top and bottom surfaces using a light-emitting diode curing light (Demi Plus; Kerr, Brea, CA, USA) and were left undisturbed in their molds for 5 minutes. The curing light had an output irradiance of 1330 mW/cm^2, a wavelength of 450–470 nm, and an exit window of 8 mm. The curing light was re-charged after every 10 specimens and a radiometer (Bluephase Meter II; Ivoclar Vivadent, Schaan, Liechtenstein) was used to verify the consistency of its performance. The polyester strips or glass slides were subsequently removed, and the materials were further polymerized with another 2 overlapping 10-second irradiations from the top surfaces. Specimens were then separated from their

Table 1: *Technical Profiles and Manufacturers of the Materials Evaluated*

Material	**Manufacturer**	**Type (Curing methods)**	**Resin**	**Fillers**	**Filler content % by weight/ volume**
Filtek Z350XT	3M Oral Care, St Paul, MN, USA	Nano-filled Composite (light cured)	Bis-GMA, Bis-EMA, UDMA, TEGDMA, PEGDMA	Prepolymer fillers containing barium glass, ytterbium fluoride, and spherical mixed oxides. Nonaggregated silica and zirconia Aggregated silica/ zirconia clusters	78.5/59.5
Cention N	Ivoclar, Vivadent Inc, NY, USA	Alkasite (self-cured with optional light curing)	UDMA, DCP, aromatic aliphatic-UDMA, PEG-400 DMA	Br-Al-Si glass, ytterbium and trifluoride Isofiller Calcium barium aluminium fluorosilicate and calcium fluorosilicate glass fillers	75/61
Beautifil-bulk Restorative	Shofu, Kyoto, Japan	Giomer (light cured)	Bis-GMA, UDMA, Bis-MPEPP, TEGDMA	S-PRG (based on fluoroboroaluminosilicate glass) and nano fillers	83.3/69
Activa Bioactive Restorative	Pulpdent, Watertown, MA, USA	Enhanced resin-modified GIC (dual-cured/light-cured)	Blend of UDMA and other methacrylates with modified polyacrylic acid	Bioactive glass, sodium fluoride	55.4 wt

Abbreviations: Aromatic aliphatic-UDMA, tetramethyl-xylylendiurethane dimethacrylate; Bis-EMA, ethoxylated bisphenol-A-glycidyl methacrylate; Bis-GMA, bisphenol-A glycidyl methacrylate; Bis-MPEPP, 2,2-Bis (4-methacryloxypolyethoxyphenyl) propane; DCP, tricyclodecan-dimethanol dimethacrylate; PEG-400 DMA, polyethylene glycol 400 dimethacrylate; PEGDMA, polyethylene glycol dimethacrylate; TEGDMA, triethylene glycol dimethacrylate; UDMA, urethane dimethacrylate; wt, weight.

molds, and any minor material flash was eliminated with fine polishing discs (Sof-Lex; 3M Oral Care). The finished specimens were inspected for defects and a digital vernier caliper (Mitutoyo Corporation, Kawaski, Japan) was used to establish the parallelism, as well as the final dimensions, of the specimens.

The measured specimens were randomly allocated into 4 groups (n=10) and conditioned in deionized solution (DI), remineralizing solution (RE), demineralizing solution (DE), or pH cycled (PC) for 14 days at 37°C in an incubator (IN-450, Memmert, Schwabach, Germany). The composition of the RE and DE solutions and their pH are displayed in Table 2.18 Each pH cycle involved exposure to DE (pH 4.8) for 8 hours and RE (pH 7.0) for 16 hours per day. The conditioning solutions (10 ml) were changed daily, and pH was checked with a digital pH meter (pH2700, Eutech, Singapore). Air exposure and evaporation of the solutions in the incubator was minimized with the use of sealed containers. Following conditioning/pH cycling, the specimens were rinsed with DI, gently air-dried, and subjected to 3-point flexural testing.

Flexural Testing

The conditioned/PC specimens were loaded in a universal testing machine (Shimadzu Corporation, Kyoto, Japan) with a load cell of 5 KN and crosshead speed of 0.5 mm/minute until fracture occurred based on the MFT. The distance between the supports for the 3-point bending set-up was fixed at 10 mm. Flexural strength, σ, in megapascals (MPa) was calculated using the following formula:

$$\sigma = \frac{3PL}{2BH^2},$$

where P is the maximum load exerted on the specimen in newtons; L is the support span in 10 millimeters; B is the width of the specimen in mm; and H is the height of the specimen in mm.

Flexural modulus, E', in MPa was calculated using the following equation:

$$E' = \left(\frac{F}{D}\right)\left(\frac{L^3}{4BH^3}\right),$$

where F/D is the slope, in newtons per millimeter, measured in the straight-line portion of the load-deflection graph. L, B, and H had been defined in the flexural strength equation. Flexural modulus was converted to gigapascals (GPa).

Table 2: List of Conditioning Mediums, Their Composition and pH

Group	Conditioning Medium(s)	Constituents	pH
DI (Control)	Deionized water	H_2O	7.0
RE	Remineralizing solution	0.9 mM NaH_2PO_4, 1.5 mM $CaCl_2$, 0.15 mM KCl	7.0
DE	Demineralizing solution	2.2 mM NaH_2PO_4, 2.2 mM $CaCl_2$, 50 mM acetic acid	4.8
PC (pH cycled)	Demineralizing solution for 8 hours	2.2 mM NaH_2PO_4, 2.2 mM $CaCl_2$, 50 mM acetic acid	4.8
	Remineralizing solution for 16 hours	0.9 mM NaH_2PO_4, 1.5 mM $CaCl_2$, 0.15 mM KCl	7.0

Abbreviations: $CaCl_2$, calcium chloride; DI, deionized solution; DE, demineralizing solution; H_2O, water; KCl, potassium chloride; NaH_2PO_4, monosodium phosphate; PC, pH cycled; RE, remineralizing solution.

Statistical Analysis

Statistical analyses were carried out using SPSS software (Version 23; SPSS Inc, Chicago, IL, USA) and parametric statistical methods were employed, as data was found to be normally distributed (Shapiro-Wilk's test). Interactions between the independent variables (materials and mediums) for the 2 dependent variables (flexural modulus and strength) were assessed using two-way analysis of variance (ANOVA). One-way ANOVA and Tukey's post hoc test was performed at significance level α=0.05 for intergroup comparisons.

RESULTS

Mean flexural modulus (GPa) and strength (MPa) of the various materials after conditioning/pH cycling are shown in Table 3. As two-way ANOVA showed significant interactions between the mediums and materials, additional statistical analysis was performed based on individual materials and mediums. The results of intermedium and intermaterial comparisons are reflected in Tables 4 and 5, respectively.

The mean flexural modulus ranged from 3.5 ± 0.3 GPa for AV conditioned in DI to 7.4 ± 0.2 GPa for BB exposed to RE. While the difference in flexural modulus was statistically insignificant for CN and BB, significant differences were observed between conditioning mediums FZ and AV (Table 4). For FZ, the modulus after pH cycling was significantly higher

Table 3: *Mean Flexural Modulus (GPa) and Strength (MPa) of the Various Materials*

Materials	Conditioning Mediums	Flexural Modulus Mean (SD)	Flexural Strength Mean (SD)
Filtek Z350	DI	6.5 (0.6)	123.5 (12.3)
	RE	6.6 (0.3)	111.2 (9.5)
	DE	6.2 (0.5)	96.3 (8.7)
	PC	7.3 (0.5)	115.8 (8.8)
Cention N	DI	7.2 (0.5)	98.3 (9.0)
	RE	7.1 (0.7)	98.6 (6.8)
	DE	6.5 (0.6)	98.1 (7.6)
	PC	6.5 (0.6)	97.7 (9.7)
Beautifil-bulk restorative	DI	7.3 (0.5)	110.5 (6.5)
	RE	7.4 (0.2)	99.5 (6.8)
	DE	7.4 (0.2)	93.0 (7.8)
	PC	7.0 (0.5)	87.0 (8.9)
Activa bioactive restorative	DI	3.5 (0.3)	113.6 (4.3)
	RE	3.9 (0.2)	114.3 (7.4)
	DE	3.6 (0.3)	97.5 (9.6)
	PC	3.8 (0.3)	112.0 (7.3)

Abbreviations: DI, deionized solution; DE, demineralizing solution; PC, pH cycled; RE, remineralizing solution.

Table 4: *Comparison of Flexural Properties Between Conditioning Mediums for Each Material*

Materials	Flexural Modulus	Flexural Strength[a]
Filtek Z350	PC > DI, DE	DI, PC, RE > DE
Cention N	NS	NS
Beautifil-bulk Restorative	NS	DI > RE, DE, PC RE > PC
Activa Bioactive Restorative	RE > DI	RE, DI, PC > DE

Abbreviations: DE, demineralizing solution; DI, deionized solution; PC, pH cycled; NS, no statistical significance; RE, remineralizing solution.

NS denotes no statistical significance while > indicates statistical significance. Results of one-way analysis of variance and post hoc Tukey test (p<0.05).

than conditioning in DI and DE. For AV, storage in RE resulted in significantly higher modulus values than in DI. No significant difference in flexural modulus was observed between DI and DE/PC (acidic environments). For all conditioning mediums, AV had a significantly lower modulus than the other materials (Table 5). When conditioned in DI, RE, and DE, BB had a significantly higher modulus than FZ. In addition, BB was also significantly stiffer than CN after exposure to DE. When PC, FZ had a significantly higher modulus than CN.

The mean flexural strength varied from 87.0 ± 8.9 MPa for BB when PC to 123.5 ± 12.3 MPa for FZ conditioned in DI. The flexural strength of CN was not affected by conditioning mediums and pH cycling (Table 4). For FZ and AV, storage in DE presented significantly lower strength values than all other mediums, including PC. For BB, exposure to RE, DE, and PC resulted in a lower strength than DI. Moreover, conditioning in RE gave higher strength values when compared with PC. No significant difference in flexural strength was observed between materials after conditioning in DE (Table 5). When stored in DI, CN was significantly weaker than all other materials and FZ was stronger than BB. After conditioning in RE and pH cycling, FZ and AV presented higher strength values than BB and CN.

DISCUSSION

Key Findings and Methods

This study examined the effect of cariogenic environments on the flexural properties of bioactive restorative materials and compared the flexural modulus and strength between different bioactive materials. As cariogenic challenges and environmental

Table 5. *Comparison of Flexural Properties Between Materials for Each Conditioning Medium*

Conditioning Medium	Flexural Modulus	Flexural Strength
DI	BB, CN > FZ > AV	FZ, AV, BB > CN FZ > BB
RE	BB, CN, FZ > AV BB > FZ	AV, FZ > BB, CN
DE	BB > CN, FZ > AV	NS
PC	FZ, BB, CN > AV FZ > CN	FZ, AV > CN, BB

Abbreviations: AV = Activa Bioactive Restorative; BB = Beautifil-bulk Restorative; CN = Cention N; FZ = Filtek Z350.

NS denotes no statistical significance while > indicates statistical significance. Results of one-way ANOVA and post-hoc Tukey test (p<0.05).

calcium phosphate influenced the flexural properties of some materials—and significant differences in both modulus and strength were observed between materials—all 3 null hypotheses were duly rejected. A conventional composite (FZ) and DI that is void of all ions served as the controls for the present study. The RE (pH 7.0) mimicked saliva that is saturated with calcium phosphate, while an acidic cariogenic environment (pH 4.8) is represented by the DE. The alternating demineralization and remineralization of teeth was developed by ten Cate and Duijsters to produce artificial caries.[19] This pH cycling model attempts to replicate the periodic alternation of pH that occurs intraorally when sugars are metabolized by cariogenic bacteria, imitating the mineral lost and gained during caries formation.[20] The pH-cycling approach is especially applicable given the hypothetical interactions between bioactive restorative materials with tooth tissues and their surrounding environments.[9,10]

The technical and clinical significance of flexural testing has been detailed in earlier studies.[13,21] Briefly, flexural modulus defines material stiffness/rigidity, while flexural strength characterizes the maximum bending stress that can be applied before the material fractures. More recently, laboratory fracture strength was found to be correlated to material wear in clinical trials.[22] It was opined that for Class I, II, III, and IV cavities, materials with high modulus and strength are necessary to minimize deformation and fracture under occlusal forces during function and parafunction. In Class V cavities, materials with a low modulus that are capable of flexing with teeth are favored, as they are believed to reduce stress and debonding at the adhesive interface of restorations.[23,24] The MFT was selected over the traditional International Standards Organization (ISO) flexural test due to its strong correlation with the latter, greater clinical relevance, and laboratory efficiency (less specimen fabrication flaws and materials are required).[21]

Flexural Modulus

The flexural modulus of CN and BB was not significantly influenced by conditioning mediums/pH cycling, while that of FZ and AV was. For FZ, modulus after storage in DI and DE was observed to be lower than that for RE and PC. The reduced resistance to elastic deformation of RBCs after exposure to DI and other aqueous solutions is well reported in the literature and has been attributed to water sorption.[23,25] In addition to the plasticizing effect of water on the resin matrix, complete or partial debonding of the fillers also occurs arising from "stress corrosion"; this arises from the diffusion of water and the swelling of the resin matrix that generates radial tensile stresses at the resin-filler interfaces.[26] Nano-filled composites utilizing nanoparticles, and nanocluster fillers like FZ, appear to be more susceptible to degradation by acidic solutions.[27] The modulus of AV, as with the other bioactive restorative materials CN and BB, was not significantly decreased by acidic environments (ie, DE and PC). When conditioned in RE, a significantly higher stiffness was noted when compared with DI. AV is an RMGIC that is enhanced with a resin matrix that releases and recharges calcium, phosphate, and fluoride.[14,28] As with other GICs, they may be capable of surface apatite formation when environmental calcium phosphate is high.[15] This has been shown to increase the physico-mechanical properties of GICs and could likewise increase the flexural modulus of AV in the present study.[15]

Despite the aforementioned potential impact on flexural modulus,, AV had a significantly lower modulus than FZ, BB, and CN regardless of conditioning mediums. This may be attributed to its lower filler content when compared with the other materials (55.4 vs 75.0%–83.3% weight).[29]

The flexural modulus of BB was generally significantly greater than that of the conventional composite FZ. BB is a bulk-fill giomer restorative that employs the use of prereacted glass ionomer (PRG) technology. Fluorosilicate glass is reacted with polyacids, freeze-dried, milled, silanized, and ground to attain the PRG fillers, which are incorporated into nano-filled resin copolymers. Given its greater filler content, the higher modulus of BB when compared with FZ was anticipated. Findings of this study corroborated those of Eweis and others, who ascertained that BB had a similar or greater modulus than conventional and bulk-fill restorative/flowable RBCs.[13] CN is a powder-liquid–based alkasite material. The powder is comprised of a wide array of calcium/barium glasses, ytterbium trifluoride, and isofillers, while the liquid is a mixture of 4 dimethacrylate resins. The lower modulus of CN, when compared with BB and FZ after exposure to DE/pH cycling, could be partly contributed by the degradation of the array of basic glass fillers employed in CN when exposed to acids.[14]

Flexural Strength

Flexural strength, like flexural modulus, is determined by filler characteristics (composition, content, size, and geometry), as well as the constitution of the resin matrix.[24,29,30] Collectively, these factors can also influence the degree of conversion and water sorption that may also have bearing on flexural properties.[31] With the exception of CN, conditioning of all materials in DE significantly lowered flexural strength. Furthermore, BB was also significantly weakened after exposure to RE and PC. The decrease in flexural strength observed with FZ, BB, and AV after storage in DE may be attributed to the susceptibility of their fillers to acid degradation.[14,25,27] Moreover, the polysalt matrix of AV may also be prone to acid attack.[32] The strength of FZ and BB was also found to be reduced after storage in citric acid.[13,25] BB was also weakened after RE/PC, and this may be contributed by its lower degree of conversion and conceivable increased environmental interactions.[33] The alkasite (CN) was the only material that was not affected by both cariogenic challenges or environmental calcium phosphate. This was in spite of its high ion release and may be contributed in part by the formation of an apatite-like phase in neutral artificial saliva.[14] Fluoride ion release and alkalizing potent in acidic conditions was found to be even greater if CN was not light-cured.[34]

When conditioned in DE, no significant difference in strength was observed between the materials. This could be ascribed to the deterioration of all the materials evaluated, including the conventional composite FZ when exposed to acidic environments.[14,27] When conditioned in DI, CN was significantly weaker than FZ, BB, and even AV despite its high filler content. The high release of ions associated with calcium fluorosilicate glass degradation in neutral pH might be an explanatory factor.[14] PRG filler degradation and ion release may also contribute to the significantly lower strength of BB when compared with FZ in DI. This could be further heightened by BB's relatively low degree of conversion.[33] In addition to fluoride, other elements released by PRG-fillers include aluminum, boron, sodium, silicon, and strontium.[35] Despite its relatively low filler content, no significant difference in strength was observed between FZ and AV when stored in DI. The low modulus of AZ could theoretically offer greater resilience and enhance its ability to withstand higher stress prior to fracture.[13]

After storage in RE and pH cycling, BB and CN were significantly weaker than FZ and AV. Filler content by itself cannot explain this observation, as both BB and CN had comparable or higher filler loading than FZ and AV. Both RE and PC involved exposure to calcium phosphate-rich solutions. Unlike GICs, the calcium glass fillers used in the giomer and alkasite are partially (ie, surface) reacted or unreacted. As they elute ions that play a role in mineral induction,[36] it is conceivable that these glass fillers are more reactive and interact considerably with environmental phosphate, leading to lower strength values. The formation of surface apatite phases on CN after conditioning in artificial saliva provides some evidence for this phenomenon.[14]

Limitations of the Study

Despite its positive findings, the present study had some limitations. First, the materials were exposed to the various mediums or pH cycling continuously for 14 days. While this conditioning period may appear extensive and might exaggerate the effects of DE/PC, it may occur in vivo at the margins and contact areas of restorations where dental plaque accumulates. As bioactive materials are capable of releasing ions over long periods of time,[37] the duration of exposure could also be extended to establish the long-term effect of the mediums or pH cycling on flexural properties. Second, the titratable acidity of the conditioning solutions, including DI, which is known to absorb atmospheric carbon dioxide to produce carbonic acid, should be examined in addition to pH. Titratable acidity was reported to play a more crucial role in surface degradation of RBCs than pH.[38] Third, bioactive materials are not a homogenous group of materials and, as such, more products from different manufacturers must be evaluated before definitive conclusions on the effect of cariogenic challenges on bioactive restorative materials can be made. In the meantime, caution must be exercised when indicating these materials in large cavities of patients with a high caries risk. Additionally, the assessment of flexural properties should be coupled with ionic and polymeric release of the various materials to better explain any observed changes. Finally, the present study utilized a static MFT that does not yield identical results to the ISO flexural test and provide an understanding of material structures. As most contemporary bioactive restorative materials are hybrids of RBCs and GICs, they are anticipated to undergo both viscous and elastic deformation when loaded. Dynamic mechanical analysis (DMA) should be conducted to examine the visco-elastic behaviors of these bioactive materials.[39] Moreover, DMA can be performed with different temperature loading frequencies and displacements to simulate variations in intraoral conditions.

CONCLUSIONS

Within the limitations of this study, the following conclusions can be derived:

1. The effect of conditioning/pH cycling on flexural properties of bioactive restorative materials was material and medium dependent.
2. The flexural modulus of the bioactive materials was not affected by cariogenic conditions. Environment calcium phosphate appeared to increase the rigidity of the enhanced RMGIC (AV).
3. With the exception of the alkasite material (CN), flexural strength of the giomer (BB), enhanced

RMGIC (AV), as well as the conventional RBC (FZ) was negatively impacted by cariogenic conditions.

4. AV had a significantly lower modulus than all the materials evaluated, regardless of conditioning environments.
5. Apart from conditioning in DE where differences in flexural strength was insignificant, the conventional composite (FZ) and enhanced RMGIC (AV) were significantly stronger than the giomer (BB) and alkasite (CN) restoratives.

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Conflict of Interest

The authors of this manuscript certify that they have no proprietary, financial, or other personal interest of any nature or kind in any product, service, and/or company that is presented in this article.

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