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New reparative material for repairing failed tooth-restoration complex

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New reparative material for repairing failed tooth-restoration complex

A Thesis submitted for the degree of

Doctor of Philosophy

Lamis Abdul Hameed Al-Taee

King's College London
London, United Kingdom
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Abstract

Objectives: This research aimed to develop a novel reparative material based on resin-modified glass-ionomer cement (RMGIC) chemistry. Two objectives were pursued. Firstly, a systematic study evaluated the ISO properties of 8 commercially available GICs / RMGICs comparing the effect of auto and manual mixing. Secondly, combining the properties of ethylene glycol methacrylate phosphate (EGMP) monomer, a proton-conducting electrolyte with functional groups as an adhesion promoter in RMGICs was explored for its potential application as a reparative material for failed tooth-restoration complexes (TRCs). The reactive polar groups were expected to interact with the metallic cations during setting reactions of the cement and form complexes that might alter its physical properties. Based on this hypothesis a novel class of material with a dynamic interaction with the tooth tissue and restorative material via the inclusion of EGMP as a monomer in commercial RMGIC's were formulated and characterised.

Materials and methods: In the first experiment, the physical properties of eight commercial restorative materials (Fuji IX GP Extra (C&H), Ketac^{TM} Fill Plus Applicap (C&H), Fuji II LC (C&H), Glass Carbomer Cement and Equia® Forte Fil), capsulated versus hand-mixed, were assessed and compared up to four weeks storage in artificial saliva at 37°C. The properties include the compressive strength (CS) and compressive modulus (CM), microhardness (MH), biaxial flexural strength (BFS), fluid uptake and fluoride ion release. In the second experiment, EGMP was incorporated at different proportions (10-40% by weight) to the liquid phase of the commercial RMGIC (Fuji II LC). Optimisation and chemical characterisation of the modified formulations were done to justify the best formulation with optimised physical and adhesion abilities to proceed with, as a step forward developing the new reparative material. The physical properties include working and setting time, CS and CM, MH, BFS, water uptake behaviour, fluoride ion release after different time intervals. Furthermore, the structural and chemical characterisations of the modified formulations were accomplished using FTIR and SEM-EDX analysis. The 30% by weight EGMP-RMGIC (pRMGIC) was selected for in vitro testing to evaluate the shear bond strength (SBS) after 24 h and 3 months' storage to different tooth surfaces (sound enamel, demineralised enamel, sound dentine and carious affected dentine (CAD)) and restorative interfaces (amalgam, composite, RMGIC and GIC). The results were compared to three different commercial restorative materials RMGIC (Fuji II LC), GIC (Fuji XI GP), and universal composite resin (Filtek™ Supreme). All comparisons were considered statistically significant if $p<0.05$. Failure modes and SEM images were analysed.

Results: The encapsulated systems showed superior performance than their equivalent manually-mixed cements due to no variation in powder/liquid ratio, the reduced porosity, uniform wetting of the powder particles during mixing and reduced operator-induced variability. The experimental EGMP-contained cements exhibited higher CS and CM, MH and a two-fold increase in the BFS compared to the control cement post-ageing. The microstructure exhibited an integrated structure that accounted for the increased stiffness and BFS with increasing the content of EGMP. The phosphate groups accounted for the hydrophilicity that it is beneficial in term of adhesion with tooth structure whilst the interaction with the matrix decreased the solubility and fluoride release. pRMGIC showed a robust and durable bond strength to different dental substrates (healthy and diseased). Ageing has no significant effect on further enhancement of the bond strength, except to sound dentine, however, there was a shift from adhesive to mixed/cohesive modes in most groups after three months' storage. This may indicate the potential of augmented chemical integrations of pRMGIC via the phosphate groups with the remaining tooth structure. pRMGIC can effectively repair the conditioned amalgam surfaces when used with adhesive. It demonstrated an effective repair strength to RMGIC and resin composite substrates after three months' storage whether applied with or without an adhesive. In GIC repair, the repair strength of pRMGIC was comparable to the control, however, the adhesion strength was higher than the cohesive strength of the substrate.

Conclusion: This thesis demonstrates the successful inclusion of EGMP monomer into the RMGICs as an effective and innovative material, specifically as a reparative material for failing TRCs and also as a restorative-grade GIC. Results lay the foundation to develop further encapsulated system with the scope of incorporation of remineralising bioactive glasses.

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List of abbreviations

ISO: International Organization for Standardization

TRCs: tooth-restoration complexes.

RMGIC: resin-modified glass-ionomer cement.

EGMP: ethylene glycol methacrylate phosphate

pRMGIC: phosphorylated resin-modified glass ionomer cement.

GIC: glass-ionomer cement.

CGIC: conventional glass-ionomer cement.

CS: compressive strength.

CM: compressive modulus.

MH: microhardness.

BFS: biaxial flexural strength.

SBS: shear bond strength.

CARS: caries associated with restorations and sealants.

CAD: carious affected dentine.

GC: glass carbomer cement.

RBC: resin-based composite.

BAG: bioactive glass.

β-TCP: β-tricalcium phosphate.

SEM: scanning electron microscopy.

EDX: energy dispersive X-ray spectroscopy.

HAp: hydroxyapatite.

FAp: fluoroapatite.

FTIR: Fourier transform infrared spectroscopy.

KHN: Knoop hardness number.

MI: minimally intervention.

MID: minimally invasive dentistry.

XPS: X-ray photoelectron spectroscopy.

AFM: atomic force microscopy.

TEM: Transmission electron microscopy.

F-e SEM: Field-emission Scanning Electron Microscope.

TISAB: total ionic strength adjustment buffer.

ppm: parts per million.

OCT: optical coherence tomography.

PAA: polyacrylic acid.

SBF: simulated body fluid.

10-MDB: 10-methacryloyloxydecyl dihydrogen phosphate.

WL: white light.

WSL: white spot lesion.

PTFs: pretest failures.

µTBS: microtensile bond strength.

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Introduction

Overview

Dental caries is a prevalent bacterially-mediated chronic disease with individuals being susceptible to this disease throughout their lifetime (Selwitz et al., 2007). It is regarded as the primary cause of oral pain and tooth loss in all age groups globally (Selwitz et al., 2007; Costa et al., 2012). The treatment still involves the surgical excision of necrotic tooth tissue followed by restoration of the cavity with artificial dental restorations. There are many dental restorative materials which provide adequate function and aesthetics, however, maintaining the functional integrity of the tooth-restoration complex (TRC) remains a challenge in clinical dentistry. Often restorations are removed, while the replacement of the existing restorations promotes acceleration of the "restoration death spiral" since more tooth tissue is lost each time, while minimally invasive management increases TRC longevity (Blum et al., 2014). The most common reasons for TRC failures include **c**aries **a**ssociated with **r**estorations and **s**ealants (secondary caries, CARS) (Green et al., 2015, Jokstad, 2016), tooth/restoration fractures and the loss of retention/deficient marginal adaptation (Dobloug et al., 2015). Resealing such marginal discrepancies helps in the limitation of the traditional, complex and more destructive restorative therapy involved in replacing restorations. Re-sealing allows preservation of tooth structure and consequently increases the longevity of the tooth-restoration complex (Blum et al., 2014; Mjör and Gordan, 2002). A significant proportion of dental health service budgets are dedicated to the placement and replacement of restorations, all of which have a limited lifespan. Hence lowering the burden of replacing failed restorations through repair is of importance because with each intervention, the likelihood of further unnecessary tooth tissue loss increases. This further weakens the tooth and increases the likelihood of adverse sequelae such as pain, root canal treatments or extraction can be avoided (Kanzow et al., 2016; Kanzow et al., 2017).

The ideal reparative materials must provide a close affinity, both physically and chemically, to both sound and carious tooth margins in a way that minimises the risk for further tooth damage, preventing the ingress of bacteria and at the same time have the ability to adhere to variety of restorative materials, with appropriate physio-mechanical properties. However, there is no dedicated reparative dental biomaterial, and the existing materials often result in inadequate clinical outcomes (Eltahlah et al., 2018).

GIC/RMGIC systems provide long-term chemical adhesion with tooth tissue with an ability to release fluoride ions which potentially reduce the incidence of CARS (Mayanagi et al., 2014). However, they exhibited a limited ability to adhere to different restorative interfaces (Maneenut et al., 2010; Zhang et al., 2011) added to the inherent brittleness, low mechanical strength and wear resistance which limit their use for long-term repair in high stress-bearing areas (Tyas, 2003).

Phosphate functional monomers have been used widely in dental adhesive systems (Perdigão and Swift, 2015; Van Meerbeek et al., 2011). They are mainly incorporated into self-adhesive bonding agents and resin cements as an adhesion promoter. The acidity and reactivity of these functional monomers provide strong and stable bond to dental substrates via chemical interactions with the mineral component of the tooth structure (Yoshida et al., 2000; Münchow et al., 2015). Additionally, they show enhanced bond strength to dental alloys through chemical union via the oxide layer at the alloy surface, and to the other substrates including; composite resin, zirconia, noble and non-precious metals, and silica-based ceramics (Dos Santos et al., 2006; Blum et al., 2012; Balkaya et al., 2018). However, the incorporation of a methacrylated phosphate acidic monomer to RMGIC systems to enhance the bonding properties to different tooth/restorative interfaces have not yet been explored.

Aim of the study

This thesis aimed to develop and characterise a novel reparative material that utilises a resin-modified glass-ionomer cement as a base formulation in conjunction with ethylene glycol methacrylate phosphate (EGMP) resin matrix, a photoreactive acidic monomer with pendant phosphate groups. The EGMP-HEMA allows for the polymerisation, which is hypothesised to not only create a network of covalently linked phosphate groups but additionally improve the adhesion to composite resins, RMGIC/GIC's and amalgams by virtue of the polar phosphate groups. Combining the properties of EGMP monomer as an adhesion promoter within the RMGICs is a unique and interesting concept especially as the pendant phosphate groups are expected to interact with the metallic cations during cement setting to form complexes that might alter the physical and biological properties of the cement itself and in addition improve the adhesion to other dental and restorative substrates.

Description of the thesis

The thesis is divided in to 6 chapters; Chapter 1 provides a critical review of the literature related to the causes of failure of TRCs, and the current treatment modalities with an emphasis on minimally invasive approaches. A brief overview of the different reparative materials is presented, however a more in-depth discussion on the GIC/RMGIC systems is provided since the formulation of the novel reparative material is based on RMGIC cements. This includes details on composition, setting reaction, mechanical properties and previous modifications to enhance the physical and biological properties, as described in Figure1-1.

Chapter 2 describes an in vitro study evaluating the properties of eight commercial GIC/RMGIC systems in accordance to ISO standards. The study investigated two variables that can affect the properties of these cements. Since there is conflicting information in literature on the effect of manual vs. mechanical mixing of the components of GIC/RGIC systems, the effect of mixing mode (mechanical vs. manual) in three selected systems dispensed in both versions were determined under similar testing conditions. Secondly a number of GIC formulations with additives such as ultrafine glass or apatite are being advocated for clinical use hence a change in the GICs' composition is expected to influence the properties, thus two GICs dispensed in encapsulated form, were included as a part of the study to ascertain the effects on the physical properties. This experiment helped in understanding the physical properties of different commercial GIC systems and comparing the properties through changes in their composition and mixing mode. Moreover, this study showed inferior strength properties of the manually-mixed F2LC in comparison to all tested GICs. This led to selecting this commercial product to engineer a synthetic dental biomaterial that may widen its dental applications whilst improving the clinical performance, which is the rationale for developing this new class of material with potential function as a reparative material.

Chapter 3 describes the development of a new cement (pRMGIC) via incorporating different proportions of EGMP monomer to a RMGIC to enhance the physical, adhesion and biological properties of the cement to be used for repairing failed TRCs. The physical properties of the cements were determined and are described in detail and the optimal formulation for further assessment was identified.

Chapter 4 pertains to the evaluation of the interfacial adhesion of the optimised formulation (pRMGIC) to healthy and diseased tooth surfaces using shear bond strength test and SEM. Chapter 5 details the investigation of the quality and durability of the adhesive bond between pRMGIC and different restorative material surfaces and compared to different commercial products using SBS test.

The objectives of the study are the following:

1. To evaluate and compare the physical properties of eight commercial encapsulated and hand-mixed GIC/RMGICs.
2. To develop a new reparative material by formulating a series of liquid phase via the combination of EGMP and polyacrylic acid using the liquid phase of a commercial formulation (Fuji II LC). The characterisation and optimisation include assessing the setting kinetics, mechanical properties (CS and CM, MH, and BFS), fluid uptake behaviour, fluoride release, bonding to sound dentine, FTIR and SEM-EDX analyses.
3. To evaluate and assess the interfacial integrity of the optimised EGMP-contained cement (pRMGIC) to sound enamel, demineralised enamel, sound dentine and caries affected dentine (CAD) after 24 h and three months' storage using shear bond strength test (SBS) and SEM.
4. To evaluate the interfacial adhesion strength of pRMGIC to conditioned restorative interfaces (amalgam, resin composite, RMGIC and GIC) with and without adhesive after 24 h and three months' storage using SBS test.

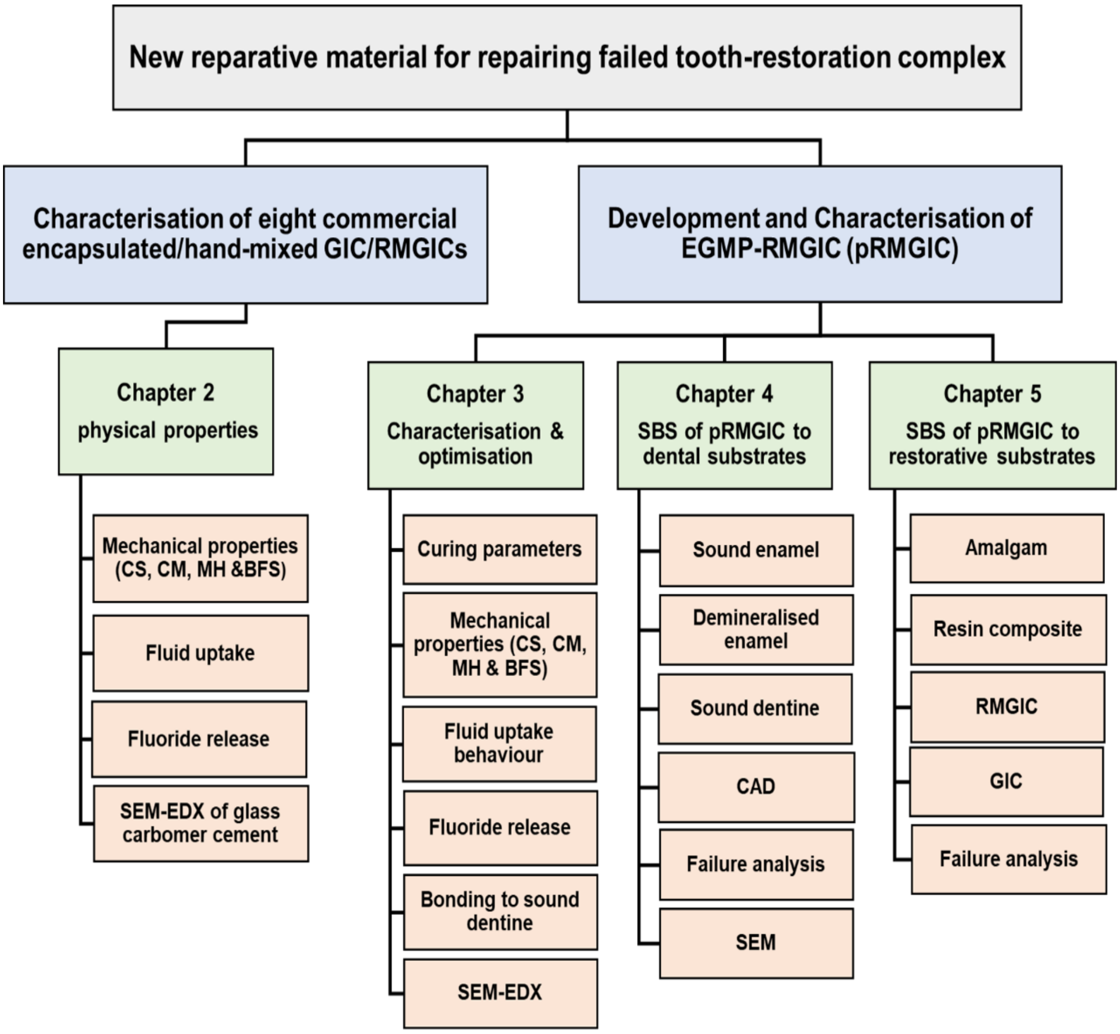


Figure i-1: Organisational flowchart of the experiments conducted in this study

Chapter one

1 Literature Review

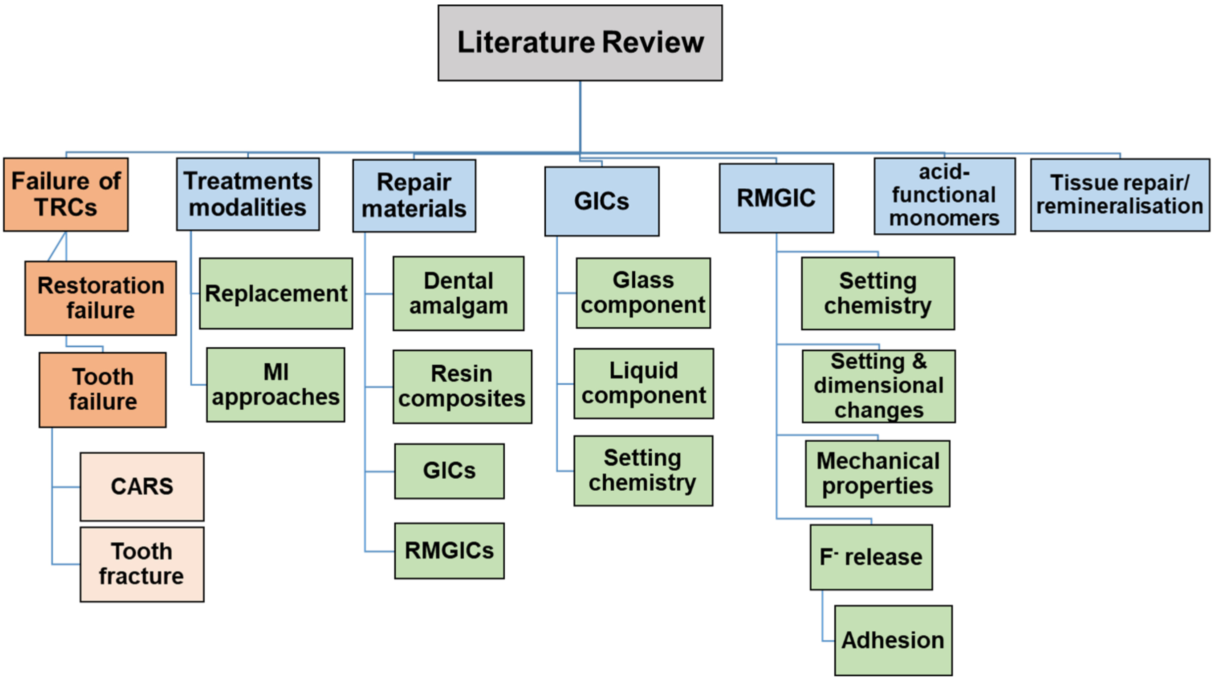


Figure 1-1 A flowchart of the literature review.

1.1 Failure of tooth-restoration complexes (TRCs)

Restorative therapy encompasses managing damage caused by dental caries, trauma, wear or erosion for prevention from further tooth loss with aesthetic and functional considerations. The success of a restoration is governed by several factors ranging from properties of the restorative material, clinical placement, and patient compliance. The ultimate goal is to improve the integrity by simulating tooth tissues both initially and over time. A restorative material should ideally possess similar mechanical and optical properties to tooth tissues, technically easy to place by the clinician, have low sensitivity to operator’s skills. However, presently there is no single dental material that satisfies these ideal requirements which at present no material fulfils. Furthermore, most restorations are significantly dependent the outcome regarding the restoration adaptation, form and function is not related to the material factor alone but also depends on the operator's clinical skills and patient factor who have a significant influence on deteriorating the restoration's technical excellence.

A failed restoration is defined as a biomechanical defect or damage resulting in immediate or subsequent detrimental clinical consequences to the patient. This may affect the restoration alone (bulk fracture, staining, etc.), the supporting tooth structure (fractured cusps, new caries at the tooth-restoration surface (CARS) etc.) or, more commonly, both, affecting the collective tooth-restoration complex. Such failure can present as apparent fractures of this complex, possibly detectable active caries associated with restoration/sealant surface (CARS, previously described as secondary or recurrent caries) or can be subtler, such as marginal discoloration of an anterior aesthetic resin composite restoration or marginal ditching of a posterior restoration (Green et al., 2015).

The quality of dental restorations is assessed by two main clinical indices: USPHS (United States Public Health Service) and CDA (standards of quality of dental care' used by the California Dental Association). Both systems evaluate colour, anatomic form and marginal characteristics (adaptation, discolouration, and caries). However, they describe only the degrees of deviation from an 'ideal' state. This means that only degree of technical excellence is addressed with operational consequences that cannot be applied with the validity in different patients, like those with high caries activity, which influence the judgement of the degree of failure and the necessity for operative intervention. Another recent assessment tool to evaluate and standardise direct and indirect restorations produced by Hickel et al. (2010) addresses it through inclusion of clinical criteria in three separate groups, namely aesthetics, function and biology. This classification is more sensitive than the previous indices with the ability to determine whether a restoration requires repair or replacement since the new clinical criteria and scoring system is a flexible method to reduce the risk of clinically unnecessary restoration replacement.

The aetiology of failure of the tooth-restoration complex can be divided into mechanical or biological, with the most common biological cause cited as CARS (summarised in Table 1-1). Generally, failures are multifactorial aetiology, which are divided into two main categories; restorative failure, and tooth failure. Failures in restorations are mainly mechanically in origin, as they are correlated to the physico-mechanical properties and biocompatibility of the restorative materials. In contrast, tooth failure is associated with mechanical, structural and biological reasons. These failures can occur independently or combined added to the clinician/patient-related factors (Hickel and Manhart, 2001).

1.1.1 Restoration failure

Presently there is no consensus regarding the longevity of restorations from clinical studies. This might be attributed to numerous uncontrollable variables related to the operator/patient and the assessment criteria for designating failures. Clinical studies that report the correlation among variables are either based on experimental or observational parameters. Experimental designs are preferred as it indicates the cause-effect relationship between different factors with a certain degree of uncertainty without limitations through bias or confounding results (Jokstad et al., 2001). Many controlled clinical studies have focused on variations in composition and physical characteristics of the restorative materials that lead to failure. Others examined the influence of other factors such as; dentist's clinical experience (cavity design and size variables, material handling and technical procedures, isolation of the working field and finishing), patient factor (gender, age, frequency of attendance), and oral environment (bite force, caries activity and microflora). In these experiments, the emphasis was related to differences without analysing the reasons, such as perceptual variations, treatment philosophies, decision making and technical skill. Additionally, only few clinical studies have sufficient sample sizes that supply good evidence of strong statistical correlations between the quality and clinical variables (Altman, 1991, Jokstad et al., 2001).

Laboratory research can only provide the indications of the possible technical excellence, while the clinical studies can, under controlled conditions, provide the indications of the potential restoration quality. Unfortunately, there is a weak correlation between laboratory and clinical findings concerning the longevity of restorations (Wilson, 1990; Tyas, 1992). An example to illustrate is the outcome of a study that correlated the failures of low copper dental amalgam with poor occlusal margins due to corrosion. However, clinical studies suggested that failure occurs primarily due to secondary caries, not the poor margins since the observation does not correlate to the problem. Furthermore, high-copper dental amalgam may fail by secondary caries and later by bulk fracture if they survive over a long time. This apparently is a different set of processes that are dependent on the intraoral conditions of the patients and linked to age, caries risk while the type of amalgam alloy had no association with the restoration survival (Kreulen et al., 1998).

Creep predicts corrosion levels but does not predict proximal caries or bulk fracture (Kreulen et al., 1998). Static mechanical tests are generally used to predict bulk fracture instead of fatigue, which represents a more realistic pathway of failure in the oral environment. One of the limitations of determining fatigue is the use of specimens with simple geometries, which do not mimic the actual shapes of clinical restorations (Anusavice et al., 2007). Tests are also run with stand-alone samples and not with those that are interfaced to tooth structure. Other variations are the differing levels of thermal cycling that range from 500-5000 cycles which are not often considered an accurate representation of the environment that a restoration experiences since it is believed that there may be no significant heat transfer during short-term thermal cycling, leading to less than meaningful data (Dunand and Derby,1993). In a similar vein, some solubility tests of dental cements are determined to be of little scientific value (Wilson, 1976). Despite the concerns, all these tools are still being used as screening laboratory tests for the longevity of dental restorations which are valid for material properties but may not be directly correlated to clinical performance (Jokstad et al., 2001).

Restorative failures are related to the weakness in the mechanical properties of dental materials (poor edge strength, compressive strength, wear, and water sorption), or problems in the technical application of the restorative materials for specific clinical situations. The longevity of restorative materials are linked to their physical properties, e.g. the long-term success rate of amalgam is related to the high compressive strength and wear resistance in comparison to the resin composites, while both exhibit superior performance than the GICs that possess low cohesive strength. Burke et al. (1999) examined the reasons for replacement and the median age of 4,608 restorations reported by 73 vocational dental practitioners and their trainers. The median age of the amalgam restorations ranked from Class I>V>II (7.4 to 6.6 yrs.), for composite resin restorations Class III> II and V>IV>I (5 to 3.3 yrs.), and for glass-ionomer cement restorations Class

III>IV> V (4.8 to 3.2 yrs.). Whilst, Manhart et al. (2004) found that the annual failure rates of posterior stress-bearing amalgam restorations are higher than equivalent resin composites (3.0%, 2.2%, respectively). The failures reported are due to secondary caries, fracture, marginal deficiencies, wear and post-operative sensitivity. However, variations in composition and physical properties within specific type of material have a minor effect on their failure rate. This is presented in two clinical studies; five-year prospective study (Van Noort and Davis, 1993) measured the survival of 2,399 Class III and 1,093 Class V chemically-activated anterior composite resin restorations in 26 general dental practices, and another cross-sectional study (Allander et al., 1989) among 75 private practitioners evaluated 1,147 old anterior restorations of 25 different materials for 2-4 years according to the CDA system. Both studies confirmed that there are no apparent differences in the quality of the dental restorations among tested materials. Nevertheless, the rapid development in the materials' science related to the mechanical properties and placement techniques will improve the longevity of these restorative materials over time. However, selection of the appropriate material for each clinical condition is beneficial, which is mainly related to the clinical skill of the operator.

Table 1-1 Restoration failure criteria (taken from Pickard's guide to minimally invasive operative dentistry, OUP Oxford, 10th edition, Banerjee and Watson, 2015)

Restoration failure criteria	**Causes/Comments**
Colour match (aesthetics)	• Underlying discolouration from stained dentine • Superficial discolouration from margin/surface staining • Underlying discolouration from corrosion products (amalgam) • Aged tooth-coloured restorative materials become stained and discoloured due to water absorption leading to a gradual change in optical properties
Marginal integrity	• Loss of marginal integrity (causing plaque stagnation) caused by: o Long-term creep/corrosion/ditching of amalgams o Margin shrinkage of resin composites/bonding agent o Margin dissolution/shrinkage of GICs o Margin chipping under occlusal loading due to poor edge strength o Presence of margin ledges/overhangs, poor contour • If patient can keep the failed margin plaque and recurrent caries-free and it is not of aesthetic/functional concern, then this partial loss of integrity may not be a sole cause to repair/replace the restoration
Marginal discolouration	• Micro-/macro-defects at the tooth-restoration interface will permit exogenous stain • penetration along the outer perimeter of the restoration as well as towards the pulp • Poor aesthetics • Is an indication of margin integrity failure • Not necessarily an indication for recurrent caries
Loss of bulk integrity	• Restorations may be bulk fractured/partially or completely lost due to: o Heavy occlusal loading-lack of occlusal analysis before restoring the tooth o Poor cavity design leading to weakened, thin-section restorations (especially for amalgams) o Poor bonding technique/contamination leading to an adhesive bond failure and lack of retention • Inadequate condensation technique/curing causing intrinsic material structural weaknesses (voids, ‘soggy bottom’) • Patients will often complain of a ‘hole in the tooth’ where food debris is trapped-high caries risk. • Bulk loss of restoration or occlusal wear may affect the bite/occlusal scheme

1.1.2 Tooth failure

Recurrent caries, marginal defects and tooth fracture remain the highly prevalent forms of tooth failures in clinical service.

1.1.2.1 *Caries associated with restorations and sealants (CARS)*

CARS is not a universal attack along the entire interface between the tooth and restoration, rather a new lesion on the surface due to local conditions. It does not differ from the primary caries as it is defined as a localised disease caused by local accumulation of mechanically undisturbed bacterial biomass with cariogenic potential (Thylstrup et al., 1994). Several facts should be taken into consideration in this regard. First, even when there is a close adaptation of the restoration to the tooth surface, there is still enough space for the bacterial ingrowth. Second, there is a little evidence of 'undetectable microleakage' causing CARS (Jokstad, 2016). Third, most papers have reported weak evidence of a correlation between the marginal discrepancies and CARS (Söderholm et al., 1989; Foster, 1994). Fourth, the ground sections of the restored teeth with secondary caries often reveal subsurface lesions unrelated to the cavity wall (Özer and Thylstrup, 1995). Thus, CARS will never develop without a cariogenic biofilm regardless of the technical quality of the restorations (Jokstad et al., 2016). Accordingly, the patient's oral hygiene habits would determine if caries develops, rather than the quality of restorations.

For ethical reasons, it is not feasible to conduct clinical trials to monitor the progress of initial secondary caries adjacent to the restoration margins, study the etiopathogenesis and/or identify the potential prognostic factors. These factors are likely to be associated with the patients, operators and the restorative materials, including the structure of the tooth-restoration interface following optimal as well as suboptimal handling and placement of the restorative material (Demarco et al., 2012).

However, the recent laboratory studies using advanced materials, preparation techniques and caries detection technology predicted a correlation between the marginal sealing of adhesive systems and the progress of demineralisation at tooth-restoration margins (Kuper et al., 2015; Turkistani et al., 2015), and the fluoride release decreases the rate of the progression.

Fluoride release from restorative materials is considered to play a part in prevention of secondary caries, however there are some conflicting reports. In a longitudinal study (Van Dijken et al., 1999) of 274 large Class II open-sandwich RMGIC restorations over three years, no secondary caries was noted, despite a large number of participating patients with high caries risk. However, the anticariogenic properties of the GIC restorations are not strongly substantiated by other clinical investigations (Randall and Wilson, 1999), or even supported by recent laboratory studies, which suggested that the fluoride-releasing activity of the GICs is inadequate for effective antibacterial conservation (Kuhn et al., 2016). As a consequence recent studies have aimed to enhance the antibacterial activity of GICs using innovative strategies. However higher amount of additives in the cements although increases the bacteriostatic effect but compromises their mechanical properties (Hafshejani et al., 2017).

Furthermore, there is no clinical evidence that the polymerisation shrinkage, cavity design (Söderholm et al., 1998) or the presence of corrosion products close to cavity walls are correlated to the development of CARS (Foster, 1994). Secondary carious lesions on proximal surfaces are difficult to detect unless the lesion is relatively advanced with considerable loss of tooth structure (Boston, 2003). Some authors reported a link between marginal fracture and CARS (Hodges et al., 1995), others do not (Kidd and O'Hara, 1990). Laboratory experiments also did not support a correlation between the size of the crevice and CARS (Söderholm et al., 1998), but describe a link to extremely cariogenic environments (Derand et al., 1991).

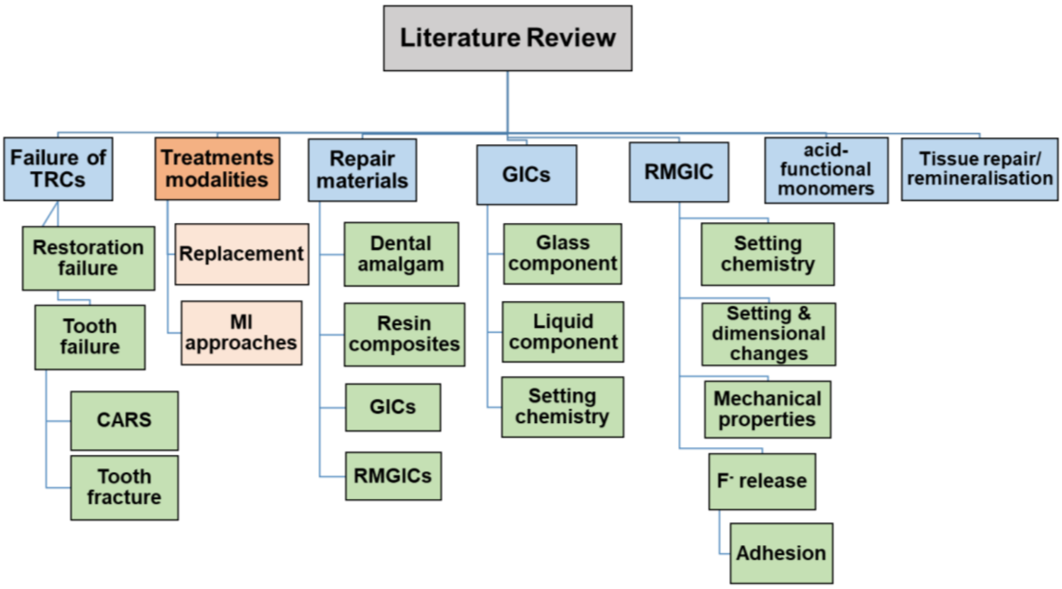
1.1.2.2 *Tooth fracture*

Tooth fracture includes the cusp fracture and cracked tooth syndrome (cracks in enamel or dentine). There is a controversy concerning the correlation between the strength of the tooth and the restorations’ material and adaptation. This is due to the relatively low incidence of tooth fractures, which impedes the execution of clinical studies. Accordingly, the relationship between the clinical factors and tooth fracture is based on the extrapolation of case description and laboratory findings (Geurtsen, 1992; Bader et al., 1995). However, it has been reported that dimensional thermal stability, hygroscopic expansion and setting/polymerisation shrinkage of restorative materials, as well as excessive loading, might induce stresses on tooth tissues (Wilson, 1990, Sindel et al., 1999) which could be tolerated by dentine due to its resiliency, but cause infractions in enamel. Furthermore, several studies postulated that the thermal expansion of amalgam, chemical reactions in the alloy and corrosion might cause enamel infractions and cusp fractures. However, there is no clinical documentation or standardised tests to screen materials for this alleged expansion (Plasmans et al., 1998). Nevertheless, stress-inducing restorations should not exceed the limits according to material test standards, Table 1-2.

Table 1-2 The mechanisms of tooth failure of direct restorations (taken from Pickard's guide to minimally invasive operative dentistry, 10th edition, Banerjee and Watson, 2015)

Tooth failure		**Comments**
Mechanical	Enamel margin	• Poor cavity design can leave weak, unsupported/undermined enamel margins which fracture under load • Cavity preparation techniques (burs) cause sub-surface micro-cracks within the grain of enamel prisms which weaken the surface ultrastructure • Adhesive shrinkage stresses on prisms at enamel surface can cause them to be pulled apart causing cohesive marginal failure in tooth structure and leading to a micro-leakage risk
	Dentine margin	• Adhesive bond to hydrophilic dentine results in a poorer quality bond which hydrolyses over time leading to increased risk of micro-leakage • Deep proximal cavities often have exposed margins on dentine • Poor moisture control leads to compromised bonding technique, in turn, increased risk of micro-leakage
	Bulk coronal/ cusp fracture	• Large restorations will weaken coronal strength of remaining hard tissue • Loss of marginal ridges/peripheral enamel will weaken the tooth crown • Cusps absorb oblique loading stresses and are prone to leverage/fracture • Can cause symptoms of food-packing and sensitivity
	Root fracture	• Often root-filled, heavily restored teeth (with post-core-crown) under heavy occlusal/lateral loads

		• Traumatic injury • Symptoms variable (pain, mobility, tenderness on biting) and radiographic assessment useful
Biological	CARS	• New caries at a tooth-restoration gap with plaque accumulation • Detected clinically or with radiographs • Marginal stain is not an indicator of recurrent caries • Can affect a section of margin and not the whole restoration
	Pulp status	• Heavily restored teeth that liable to pulp inflammation • Iatrogenic damage or ongoing disease that cause pulp necrosis
	Periodontal disease	• Examination of the periodontium required for loss of attachment, pocket depths, bone levels • Can be exacerbated by poor marginal adaptation of restorations (causing plaque and debris stagnation)/margins encroaching into the periodontal biologic width



1.2 Treatments modalities for failed restorations

The treatment of failed restorations with detectable defects must be performed without any detrimental consequence to the patients, considering the clinical and biological factors before deciding to intervene operatively. For example, ‘failing’ restorations in a patient with a low caries risk and low aesthetic demands should be treated differently from a patient with high caries risk and high aesthetic demands (Hickel et al., 2010). Furthermore, the large failures can be managed differently from a minimal defect with a clear detrimental biological consequence (active CARS). Accordingly, the decision-making process must take into account the patients' expectations and their attitude to take the responsibility for maintaining their oral health (Green et al., 2015).

The proper assessment and diagnosis of the restoration failures by trained clinicians increase the accuracy and predictability of decision-making to minimise further failures and promote the clinical longevity. This can be achieved by preserving the quantity and integrity of the healthy tooth tissue via wide range of clinical steps including non-operative reviewing, refurbishment, resealing and repair, before considering restoration replacement, as described in Table 1-3. The understanding of MI operative techniques with an appreciation of the histological properties of the tooth substrates combined with the properties of the restorative materials is critical in the long-term success of minimally invasive approaches (Banerjee, 2013, Banerjee, 2017).

Table 1-3 The minimally invasive '5 Rs' concept to manage failing tooth-restoration complexes, (Green et al., 2015; Banerjee and Watson, 2015)

Reviewing	The monitoring of minor defects, where there would be no clinical advantage to undertaking treatment
Refurbishment	The treatment of small defects present in the restoration which require intervention to prevent further deterioration
Resealing	The application of sealant into a non-carious, defective marginal gap
Repair	The placement of additional restorative material to an existing restoration
Replacement	The removal and replacement of an entire restoration

When defective restorations require intervention, clinicians are often challenged to replace or repair the existing restorations. Based on traditional teaching approaches, complete removal is required if the restorations do not satisfy the strict quality requirements (Blum et al., 2003). However, in recent years, there is an increased demand towards repair rather than complete removal. There are many biological and financial reasons to retain sound parts of the old restoration in place. These include a reduction in costs, unnecessary removal of the tooth structure and avoidance of repetitive trauma from the restorative procedures (Sharif et al., 2010a). The decision regarding the replacement or repair is based on the clinician’s knowledge and clinical experience. Currently, the dental practices shifted towards conservative treatment approaches, and consequently, many dental schools throughout the world have embarked on elaborating the concept of repair of restorations at the undergraduate level either in preclinical or clinical years (Brunton et al., 2017).

1.2.1 Replacement of failed restorations

Dental restorations have limited service life and might be prone to failure due to biological, mechanical or aesthetic reasons generating the need for replacement. The criteria for the replacement of restorations are ill-defined and subjective due to variations among clinicians that lead to diversity in clinical judgment regarding the causes of failure, added to the scarce calibration of these clinical judgements. Large variations in the diagnosis have been noted in many studies (Kidd et al., 1995, Mjör et al., 2000) and despite these disparities, clinically diagnosed CARS is the predominant cause for replacement. The CARS with a histopathological entity may not be related to the crevices at the tooth/restoration interface (Kidd et al., 1995, Özer and Thylstrup, 1995), however, it can be developed when these crevices are located at the gingival part of Class II, III, IV, and V restorations (Mjör, 1998) where it is there is a difficulty to gain proper margins either during restoration’s placement or after restoration setting, or to maintain a proper oral hygiene..

The replacement of glass-ionomer cements in multi-surface ART restorations are mainly due to gross marginal defects induced by occlusal forces or insufficient wear resistance of the restorative material, loss of retention and bulk fracture (Yip et al., 2001; Kopperud et al., 2012). CARS have also been reported as a frequent cause for replacing GIC restorations (Mjör, 1996) supported by a study where more than half of 662 glass-ionomer restorations are replaced due to CARS (Mjör et al., 2000). There are limited clinical studies confirm the caries inhibitory effect of the fluoride-releasing materials (Horsted-Bindslev, 1994; Arends et al., 1995) with a possibility to be applied in patients or specific sites that are prone to caries development than others. Fluoride release from GICs can prevent in vitro induced secondary caries, but its inhibitory effect in vivo is not proven (Jokstad, 2016).

The main clinical concerns regarding replacement of resin composite restorations are resin degradation, wear, marginal staining, cusp deflection, gap formation, dentine sensitivity and bulk failures when used in stress-bearing areas (Mjor, 1997; González-López et al., 2007; Baracco et al., 2012). The polymerisation of conventional methacrylate resin-based composites and associated deleterious effects of the residual shrinkage stress on the adhesive junction lead to debonding or cohesive fracture within the restoration and/ or tooth structure (Loguercio et al., 2004). Particularly in the deep proximal margins of the Class II restorations in which stresses affect the interfacial adaptation (Loguercio et al., 2004) increasing the susceptibility to CARS (Ferrari and Davidson, 1996). Recently, a TEGDMA/HEMA-free resin composite system was introduced based on the silorane monomers with traditional filler particles. This system eliminates the adverse cytotoxic effect of the methacrylate monomers TEGDMA (Triethyleneglycol-dimethacrylate) and HEMA (2-hydroxyethyl-methacrylate) (Geurtsen and Leyhausen, 2001) with reduced volumetric shrinkage via a ring opening polymerisation process. A meta-analysis of 11 clinical studies showed acceptable performance for this system compared to the conventional composite with traditional monomers (Baraúna Magno et al., 2016; Van Dijken and Pallesen, 2017). They exhibited reduced water sorption, solubility and diffusion coefficient, which may potentially improve the hydrolytic stability of RBC restorations (Palin et al., 2005a). However, the biomechanical properties did not show better performance than the hybrid or nano-filled composite systems unless being used with HEMA/TEGDMA free adhesive (Ilie, and Hickel, 2009). The main reasons of failure in this system are bulk fracture followed by recurrent caries. Gap formation was also recognised which seems due to underperforming bonding approaches rather than differences in the composition of resin composites (D'alpino et al., 2011). Additionally, the low-shrinking composite Filtek Silorane showed a significantly lower µTBS to dentine compared to the conventional composite (Filtek Z100), suggesting that factors other than the polymerisation shrinkage might influence the adhesion to tooth structure (Van Ende et al., 2010).

The most frequent reasons for amalgam replacement are CARS (Mjör, 1997), bulk/ cusps fractures (Burke et al., 2001), ditched margins as well as the demand for more aesthetic treatment (Forss and Wildstrom, 2004). Total replacement is the most common treatment for defective amalgam restorations even with the potential loss in tooth structure. However, amalgam replacement with new amalgams might be less common nowadays due to controversies regarding its safety and national as well as global efforts to limit or even ban its use in dentistry (Kopperud et al., 2012).

Complete replacement of partially defective restorations results in unnecessary removal of healthy dental hard tissue by adding an additional restoration surface and increases the risk of restorative failure and subsequently re-treatment leads to further tooth destruction. There is also an increased risk of pulpal complications over time (Kanzow et al., 2017) which subsequently may need an endodontic treatment or lead to tooth extraction.

Reducing the failure rate of restorations is considered as a major goal in dentistry. Repairing instead of replacing partially defective restorations retains the teeth for longer time by reducing the potential loss of tooth substance. However, for the cost-effectiveness, repairing a composite can be recommended while amalgam repair is considered more expensive than complete replacement. When considering additional factors, repair is most suitable in large composite restorations when failed due to secondary caries. In contrast, repairing fractured composite or amalgam restorations does not seem cost-effective compared to complete replacement (Kanzow et al., 2016). In light of the identified uncertainty and bearing in mind the variety of indications for repair versus full replacement, clinical decision-making should consider patients’ and dentists’ preferences.

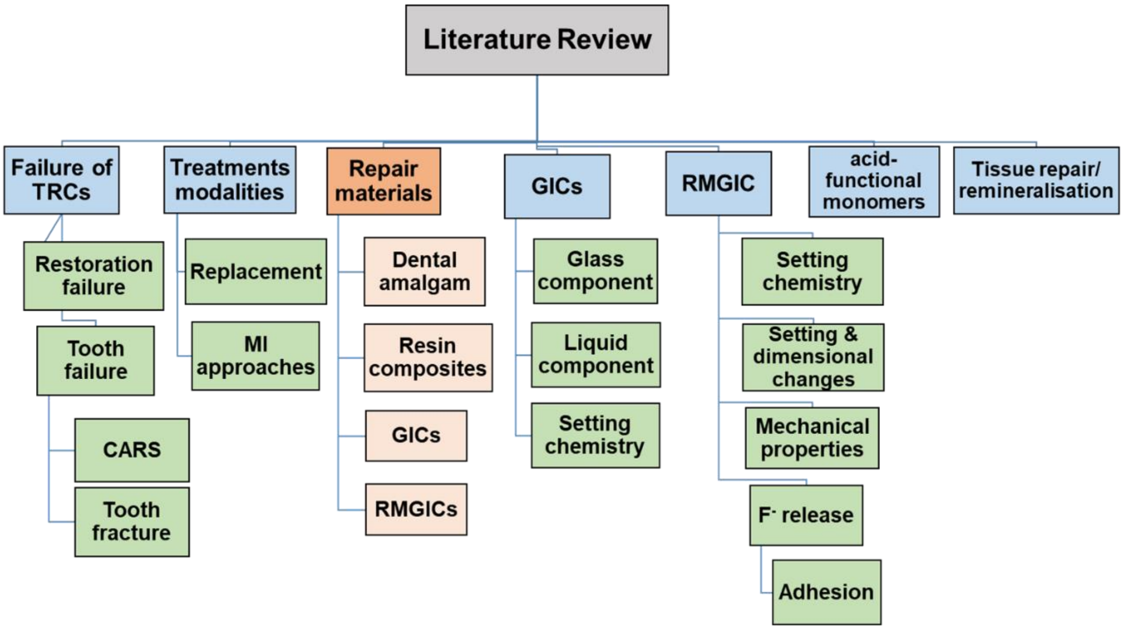
1.2.2 Treatments with minimally invasive approaches

The concept of minimal intervention evolves as a consequence of the increased understanding of caries process and the development of adhesive restorative materials. Demineralised but non cavitated enamel/dentine can be ‘healed’, and the surgical approaches that are previously used to treat caries lesions along with ‘extension for prevention’ as proposed by G.V. Black is no longer tenable whereby the caries lesions are radically removed by a surgical approach requiring the removal of the diseased tissues and the extension to areas that are presumed to be caries resistant. This is attributed to the lack of understanding of the caries process, in particular, the potential for remineralisation combined with the poor physical properties of the available restorative materials. Probably the serious consequence of the surgical approach was the extent of the cavity which had to be prepared to accommodate the principles of ‘extension for prevention, which weakens the tooth structure and leads to a marked increase in ‘replacement dentistry’, wherein there is further loss of tooth structure upon each replacement of a restoration (Tyas et al., 2000; Borges et al., 2011).

The alternative practical minimally invasive approach is the repair of defective areas (Moncada et al., 2009). Repair includes removing of part of the restoration and/or impaired contiguous tissue, and retreatment by a material that can adhere efficiently to the tooth and restorative interfaces. It is considered as state-of-art as it limits the size of restorative intervention, reduces the risk for complications and limits the costs of the intervention (Wilson et al., 2016). It is designed to promote the maximum preservation of the healthy dental structures and subsequently increases the longevity of tooth-restoration complexes. In contrast, replacing restorations tends to drive the restorative cycle towards failure by removing excessive and unnecessary quantities of natural tooth structure leads to further weakening of the TRCs. Clinical studies have reported an enhanced longevity of the repaired restorations failed due to caries compared to restoration fracture (Opdam et al., 2012; Demarco et al., 2012). However, the longevity of repaired restorations is lower compared to the original restoration, depending on the applied material and reason for repair (Opdam et al., 2012).

The proper diagnosis and selection of the minimally invasive approaches for the existing restorations are critical steps, which invariably affect the longevity of the tooth. Repair, refurbishing and sealing are valid alternative treatments than replacement, but there is still a lack of evidence for the longevity of these options. From questionnaires, dentists do repairs in their practices, however, the number of repairs performed by general dental practitioners and the consequence of the restoration survival are not known yet (Casagrande et al., 2017). A systematic review and meta- and qualitative analysis performed through 401 articles and 29 quantitative studies including 7228 dentists and 276 dental schools were surveyed, and treatment data of 30,172 restorations (Kanzow et al., 2018) illustrated that most dental schools teach repairs widely, but not all dentists employ repairs as part of their management of partially defective restorations. Furthermore, data collected from truly treated defective restorations indicate that only a minority of all restoration had been repaired, while most were completely replaced. However, there is a lack of the qualitative elements to yield a deeper understanding of the barriers and facilitators towards repairs. Some interventions like establishing financial incentives, altering the healthcare regulation, or promotion by peers could be used to facilitate repairs in dental practice.

There are in vitro studies that investigated the techniques for repairing various restorations varying from routine adhesive techniques to specific procedures including surface treatments using different mechanical and chemical approaches. For repairing composite and amalgam restorations in vitro, protocols stretch from air-abrasion to etching with hydrofluoric acid, as a pretreatment for the restoration surface, the use of silane coupling agents and/or metal primers to increase the adhesion strength to various substrates. These protocols can explain the consequence of a particular repair protocol in a dental practice on the restoration survival without any clinical relevance since most of them have not been tested in vivo. Even problems with specific procedures when applied on a restored tooth have been described (Loomans et al., 2010; Saracoglu et al., 2011) indicating that the improved longevity of repaired restoration is not merely related to a higher repair bond strength measured *in vitro* (Anusavice, 2012; Opdam et al., 2012). Therefore, early determination of the extension of the exploratory procedure is a key for the clinical decision to repair or to replace the restorations. It is not possible to repair amalgam when secondary caries is not accessible or bulk damage has occurred that would typically necessitate complete replacement. Other contraindications must be considered when patients are reluctant towards repair and prefer replacement due to the prior history of a failed repair.



1.3 Materials used for repairing failed TRCs

1.3.1 Dental amalgam

Dental amalgam is still considered as a restorative treatment option especially as it is regarded as the material of choice for stress bearing dental restorations. Despite the relatively long-term clinical effectiveness and economic viability, it does not fulfil the aesthetic demand, added to the lack of adhesion to tooth structure (Opdam et al., 2007). Amalgams present limited longevity in the oral environment, which has been reported to be between 4.7 and 11.8 years (Kim et al., 2013). Failures are associated with secondary caries, marginal deficiencies, degradation/wear, fracture, or loss of anatomic form (Moncada et al., 2015). Amalgam replacements that may be treated conservatively is preferred among dentists (Gordan et al., 2011), however, alternative treatment approaches including repair, sealing or refurbishing show a similar survival up to 5 years (Smales and Hawthorne, 2004). Amalgam can be used successfully to repair failed amalgam restorations. Longitudinal clinical trials (Gordan et al., 2006; Gordan et al., 2011; Moncada et al., 2015) reveal a broad clinical success for amalgam-amalgam repair in Class I/ Class II restorations based on proper indications. They maintain clinically acceptable characteristics and performance with no evidence of fracture at the repaired interface up to 10 years. This can be achieved through localised and accessible defects with low to medium caries risk. However, variables that are related to individual characteristics including; flexion of the tooth cusp, deep carious lesions, size or design of the restorations, malocclusion, and bruxism are not measured in these studies which influence the repair prognosis.

Laboratory studies support the use of amalgam for repairing defective amalgam restorations (Shen et al., 2006; Roggenkamp et al., 2010). Surface treatments of aged amalgam restorations appear to be a significant factor in achieving high-quality bonds. Other variables that were investigated include an uncontaminated substrate, roughening amalgam surfaces, additional undercut, using different amalgam types for repair (Shen et al., 2006) added to the use of bonding adhesives designed for metallic surfaces (Özer et al., 2002). Depending on the study design, the effect of surface treatment yielded mixed results. For example, the use of some bonding agents appeared to be superior when no mechanical roughening of the surface was performed, while other designs showed no benefit from using bonding agent if adequate roughening had been implemented (Özer et al., 2002). The shear bond strength of repaired amalgam approaches that of unrepaired amalgam and remain unaffected by the age of amalgam undergoing repair, from 24 hours through 7 years (Roggenkamp et al., 2010).

Nevertheless, amalgam is not considered as the preferred repair material due to the questionable adherence ability to different restorative substrates and tooth structure. It requires further modification in the cavity design to enhance the restoration’s retention and stability which sacrifice the tooth structure and negatively affects tooth longevity (Ermis and Aydin, 2004; Green et al., 2015). Aesthetic concerns have added to the drop-in use of dental amalgam, which has further decreased due to the ban in several countries due to environmental considerations (Kopperud et al., 2016).

1.3.2 Composite resins

Adhesive dentistry enables the advent of more conservative treatment approaches for defective TRCs. This is based on reducing the size of the prepared cavities and bonding resin-based composite to tooth/ restorative surfaces which facilitate repairing the existing restorations rather than complete replacement (Junior et al., 2009). Resin composite restoratives have seen a tremendous development over the last decades concerning material strength, handling properties, aesthetic features and longevity. However, CARS and bulk fractures remain the main reasons for restoration failures (Opdam et al., 2014).

Long-term clinical studies (Fernández et al., 2015; Estay et al., 2018) strengthen the concept that minimal intervention using resin composite with adhesive systems can increase tooth/restoration longevity. Repaired restorations following USPHS criteria with localised, marginal, anatomical deficiencies and/or CARS adjacent to the resin composite can survive up to 14 years. However, the continuous development in this field make it difficult to conduct long-term clinical trials using a particular resin composite system that would reflect their actual performance hence suggestions on composite systems for long-term repair is flawed.

Despite the difficulty in interpreting or comparing the results of bond strengths from in vitro studies due to the variety of materials and testing that are employed, there is an agreement that mechanical roughening using diamond burs, sandblasting or acid etching (Bonstein et al., 2005), and/or chemical bonding via silane/adhesive systems enhance the interfacial repair strength of resin composites to the defective restorations (Shahdad and Kennedy, 1998; Yesilyurt et al., 2009).

Much concerns have been raised on the matrix chemistry of the adhesive agent intermediated resin composite and the repaired substrate and its role on bond strength. Adhesives enhance surface wetting and improve the micromechanical retention (Brosh et al., 1997) suggesting a synergistic bonding between the conditioned substrate and enamel/dentine interfaces. Three possible mechanisms accounting for the effects of intermediary adhesives; chemical union to the old resinous matrix, chemical bonding ability to the exposed filler particles, and the micromechanical interlocking caused by monomer penetration into the micro-irregularities within the resin matrix (Staxrud and Dahl, 2015).

Bonding of dental adhesives relies partly on hydrogen bonds and molecular attraction forces, such as Van der Waals forces, more than the stronger covalent or ionic types. These relatively weak bonds are vulnerable to hydrolysis followed by degradation at the interface over time (Malacarne et al., 2006; Breschi et al., 2008). The hydrolytic instability is mainly attributed to the presence of hydrophilic monomers, such as 2-hydroxyethyl methacrylate (HEMA) which attracts water and enhances wettability that is essential for intimate contact with different substrates.

However, the stoichiometric configuration of other molecules like silane, may, on the other hand, prevent water movement and sorption in the area, making an impact on the long-term stability (Lung and Matinlinna, 2012; Staxrud and Dahl, 2015). Moreover, the presence of phosphorylated acidic monomers in the current universal adhesives, such as 10-methacryloyloxydecyl dihydrogen phosphate (10-MDP), enhance quality of the dentine-composite interface through the formation of chemical bonds to dentine and enamel via stable CaP salts (Yoshida et al., 2012). Additionally, MDP can also bond resin composite to dental alloys through chemical union with the oxide layer at the alloy surface, and to other substrates including; zirconia, noble and non-precious metals, and silica-based ceramics without the need for dedicated and separately placed primers such as silane, metal and zirconia primers (Dos Santos et al., 2006; Blum et al., 2012; Balkaya et al., 2018).

Laboratory and clinical studies indicate that MDP-based adhesives effectively and durably bond to dentine (Inoue et al., 2005; Peumans et al., 2010). The underlying mechanism of bonding is based upon submicron micro-mechanical interlocking (Van Meerbeek et al., 2003), supplemented by the primary chemical interaction of the functional monomer with HAp that remains around the partially exposed collagen (Yoshida et al., 2004; Fu et al., 2005). According to the “Adhesion-Decalcification concept” (Yoshida et al., 2001), MDP chemically bonds to HAp, produces highly stable MDP-Ca salts (Kim et al., 2010a) which contribute to the clinical longevity of the hybrid layer that resists interfacial biodegradation and thus improves bond stability (Inoue et al., 2005; Erhardt et al., 2011; Yoshida et al., 2012), Figure 1-2.

Mild self-etching is the preferred approach when dentine is involved (Van Meerbeek et al., 2011) since Ca-salts remain embedded within the hybrid layer and are not rinsed off as in the case of an etch-and-rinse procedure which causes poor adaptation to HAp-denuded collagen (Van Meerbeek et al., 2003; Liu et al., 2011). In contrast, the etch-and-rinse approach remains the preferred choice for enamel that requires sufficient etching (Perdigao et al., 2012). Selectively etching enamel combined with a mild self-etch adhesive can, therefore, be recommended to achieve effective and durable bonding to tooth enamel and dentine (Van Meerbeek et al., 2011).

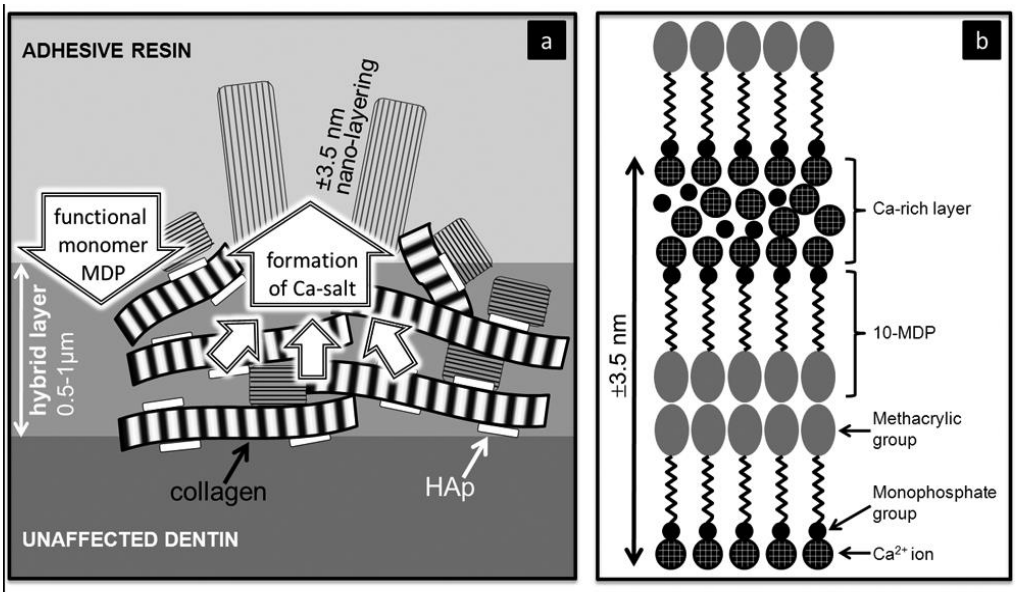


Figure 1-2 A schematic demonstrating the formation of MDP-Ca salt and an interfacial nano-layer. When the MDP-containing adhesive is rubbed on to dentine, the surface is partially demineralized up to a depth of 0.5-1µm. Ca^{2+} ions are released upon partial dissolution of HAp that diffuse within the hybrid layer and assemble MDP molecules into nano-layers, a process that is driven by MDP-Ca salt formation. The measured size of one nano-layer is about 3.5 nm, (Yoshida et al., 2012).

Resin composites have been used as a reparative material however, the longevity of the interfaces remains a challenge. This is attributed to resin degradation over time caused by water diffusion and hydrolytic deterioration of polymer chains resulting in elution of components and plasticisation of the composite (Ferracane, 2006). This compromises surface hardness, wear resistance, and subsequently interferes with the bulk properties such as; strength and fracture toughness (Ferracane and Marker, 1992). Furthermore, an initial volumetric reduction due to polymerisation shrinkage may cause stress on the cavity walls (Atai and Watts, 2006) leading to cuspal movement (Palin et al., 2005b), which may compromise the synergism at the restoration-tooth interface (Davidson et al., 1984). Consequently, de-bonding followed by bacterial micro-leakage through marginal gaps may develop associated with postoperative sensitivity, CARS, enamel fracture or pulpal inflammation/ or even necrosis (Jokstad et al. 2016).

1.3.3 Glass-ionomer cements (GICs)

Polyalkenoate glass-ionomer cement (GIC) is considered as a material of choice for atraumatic restorative treatment (ART) with minimal removal of tooth structure followed by the application of adhesive/therapeutic restorative materials. Two most significant properties in the context of minimal intervention are; adhesion to tooth structure, and release of fluoride and other ions. Adhesion occurs as a result of anion exchange between the tooth structure and the cement (Van Meerbeek et al., 2003; Peumans et al., 2005). The polyalkenoic acid from the glass-ionomer attacks tooth surface releasing calcium and phosphate ions along with calcium, phosphate, strontium and aluminium ions released from the glass, forming a new material which unites the two substrates. There is also a degree of adhesion between the acid carboxylate groups and dentinal collagen. The strength of the union is dependent upon the tensile strength of the cement itself, and any failure is usually cohesive within the cement. Thus, the stronger the cement, the higher measured bond strength (Ngo et al., 1997; Tyas et al., 2000; Croll and Nicholson 2002). In addition to the adhesive properties and biocompatibility, GIC can be applied in less accessible areas where polymerisation shrinkage of light curing RBCs cause problems, or moisture control is difficult. They provide an adequate seal against microleakage as compared to resin composite-composite restorations (Welsh and Hembree, 1985), reducing the incidence of CARS (Randall and Wilson 1999), fluoride release (Sidhu, 1993) with a possibility to remineralise CAD (Smales et al., 2005).

The commonly assumed anticariogenic properties of GICs are related to fluoride release based on laboratory studies, clinical models and retrospective clinical assessment (Knight, 1984; Mount, 1986). However, the clinical effect is not supported by the literature (Tyas et al., 1991a; Papagiannoulis et al., 2002; Alirezaei et al., 2018).

The physical properties of the glass-ionomers are essential in the context of minimal intervention techniques. However, like all water-based cements, GICs are relatively brittle with low flexural and tensile strengths, fracture toughness and a higher rate of wear compared to other restorative materials. These major drawbacks influence the survival rates when placed in load bearing areas and limit their use for long-term repair (Scholtanus and Huysmans, 2007). Following maturation, they show better performance in low-stress areas (Knight, 1992; Croll and Nicholson 2002). Even though, in larger cavities or case of failure, glass-ionomer restorations can be easily repaired or modified to a base under indirect restorations or as part of a sandwich restoration.

The ‘high viscosity’ GICs (HVGICs) were developed with smaller glass particle size and an increased powder: liquid ratio, compared to ‘normal viscosity’ GICs. Greater cross-linking in the high-viscosity GIC’s matrix enhance the mechanical properties including compressive strength, flexural strengths, surface hardness, wear resistance and solubility compared to conventional GICs (Pereira et al., 2002; Sidhu, 2011; Zanata et al., 2011). They are used in direct load-bearing restorations in both occlusal and approximal cavities in permanent and primary teeth using conventional rotary cavity preparation techniques or ART. The application of nanofilled resin coating over the HVGIC (Equia Fil) reduces moisture contamination during setting and can infiltrate the surface and seal defects thus retarding crack propagation (Diem et al., 2014). It provides slightly lower annual failure rates than other GICs and similar to that of resin composite in single surface occlusal restorations as shown in a study over six years (Gurgan et al., 2017). Nevertheless, the effectiveness of HVGIC in single-surface ART restorations is well supported (Frencken et al., 2007), while for multisurface ART restorations, the survival rates are less satisfactory, and a variable performance has been reported (Van Gemert-Schriks et al., 2007; Cefaly et al., 2007; Scholtanus and Huysmans, 2007; Ruengrungsom et al., 2018).

Metal reinforced GIC (Cermet) has been used to repair marginal defects, fractured cusps or fractured restorations. However, the survival rates did not exceed 3.5 years and failure occur mainly due to wear, fractures, defects or even a complete loss in occlusal areas (Hickel and Voss, 1988).

In vitro research indicated that surface pre-treatment enhances the repair strength of GIC to different substrates. Aboush and Jenkins, (1989) reported that the adhesion strength of GICs to mechanically-roughened amalgam is comparable to that with enamel and significantly higher than those of dentine. Surface roughening followed by phosphoric acid application promotes GIC-GIC repair (Pearson et al., 1989; Jamaluddin & Pearson, 1994) with a potential chemical bonding to the exposed glass components by free polyacrylic acid from freshly mixed cement (Pearson et al., 1989). In contrast, bonding to resin composite is micromechanical mediated by adhesive systems which ligate resin composite to the GIC via chemical and micromechanical bonding (Mount 1989; Hinoura et al., 1989; Wooford and Grieve, 1993) in which bond strength is limited by the low cohesive strength of the GIC (Sneed and Looper, 1985; Hinoura et al., 1989).

Changes in repair strength of the GIC over time are a complex phenomenon. In some instances, the improvement in the mechanical strength and wear resistance of the cement due cements' maturation produces higher long-term adhesion strength. However, bond strength can be deteriorated over time associated with the weakening in cements' properties due to erosion or the plasticising effects of water. Most studies reported that GIC gains most of its mechanical strength during the first week and remains relatively stable over several weeks and months then declined after six months (Pearson et al., 1989; Zhang et al., 2011). Generally, the slow maturation of the GIC and sensitivity to clinical conditions, associated with low cohesive strength and wear resistance rendering the long-term viability of GIC repair is unknown (Pearson et al., 1989; Zoergiebel and Ilie, 2013)

1.3.4 Resin-modified glass-ionomer cement

Resin-modified dual-setting glass-ionomer cements contain the essential components of both an aqueous glass ionomer cement and a photocurable resin (hydroxyethyl methacrylate (HEMA) and photoinitiators. The curing of the resin phase is able to reduce moisture sensitivity at the early phases of setting of GICs, hence do not require a varnish protection or isolation from moisture to protect loss of matrix forming ions and maintain their clinical advantages (Sidhu and Watson, 1995). They combine the favourable adhesive and cariostatic properties of the GICs (De Moraes et al., 2016) with longer working time and also enable command setting. The strength and brittleness are reported to be superior to conventional GIC's (McKinney and Antonucci, 1986; Mitra, 1991; Rusz et al., 1992; McCabe, 1998). The polymer matrix also assists in improving the cohesive strength and RMGICs' resistance to debonding forces which influences the adhesion to dental substrates added to the covalent linkage produced by resin components.

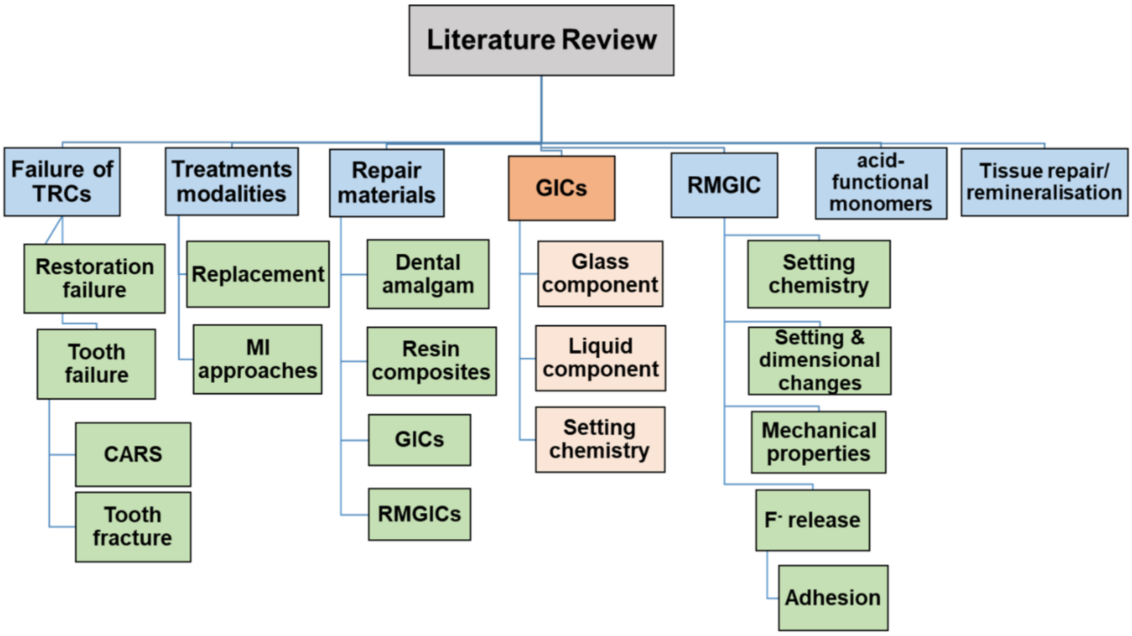
The adhesion mechanism of RMGICs to dental surfaces relies on the ion exchange mechanism, and chemical bonding of the methacrylated polyalkenoic acid to the calcium in HAp added to the micromechanical interlocking via penetration of polymer tags into the microporosities of the conditioned enamel/dentine surfaces (Mitra et al., 2009). Surface pre-treatment using weak polyacrylic acid, citric acid, phosphoric acid, and ethylenediamine tetraacetic acid (EDTA) have been reported to enhance the bonding efficiency of RMGIC to tooth surface (Inoue et al. 2001; Rai et al., 2017). This is achieved via the cleansing and demineralisation effects which increase the surface area and produce microporosities for micromechanical interlocking or hybridisation (Imbery et al., 2013).

RMGICs are used effectively in repairing non-carious cervical restorations with high retention rate (93%) in ART restorations compared to 30% in GICs when placed in Class II primary molars and anterior restorations up to six months (Rodrigues et al., 1998; Yip et al., 2001). A higher incidence of retention of RMGIC’s and lower incidence of CARS have been reported (Yap & Neo 1995; Sidhu 2010), however studies report variable microleakage results with various products. Although some of the physical properties of the RMGI cements show a marked improvement compared to conventional GIC’s long term clinical research is required to establish compelling evidence of their behaviour, particularly regarding retention in carious cavities and their biological effects (Sidhu, 2010).

The repair strength of RMGIC is adequate and satisfactory in laboratory studies, and it is significantly higher than GIC-GIC repair (Maneenut et al., 2010; Welch et al., 2015). The compatibility of the resin component of both RMGICs and resin composites with a possibility of chemical union between substrates produce high bond strength that translated by a predominant cohesive failure within RMGIC (Rusz et al., 1992; Blum et al., 2012). Surface roughening followed by etching using phosphoric acid, polyacrylic acid, or maleic acid also show evidence of improved repair strength up to six months (Yap et al., 2000), which probably is a result of micromechanical interlocking to the repaired surfaces with no obvious effect from acid surface treatment (Yap et al., 2000; Maneenut et al., 2010; Camilleri et al., 2013). The ageing also enhances the quality of repair strength (Shaffer et al., 1998) as the initial strength of the RMGIC is due to the formation of a polymer matrix while the acid-base reaction hardens and strengthens the matrix over time (Wilson, 1994).

Additionally, the chemical interaction between exposed glass particles and the polyacrylic acid produces a stable bond of the new to aged RMGIC (Yap et al., 2000). Chemical conditioning with an adhesive could further enhance the RMGIC-RMGIC bond strength (Shaffer et al., 1998) as adhesive can flow into the surface irregularities promoting the micromechanical attachment to the underlying roughened cement, added to the chemical bonding possibility with the exposed glass particles which expected to further enhance the bond strength after storage. RMGICs can adhere effectively to roughened amalgam without intermediary adhesive between them, as previously stated by Aboush & Jenkins, (1991). Other studies (Fruits et al., 1998; Pilo et al., 2012) suggested that RMGIC can offer better adhesive performance when sandwiched between resin composite and set amalgam, even better than using an adhesive.

RMGICs are still brittle cements which may occasionally lead to fracture or wear. However, the inherent weakness as reflected by the in vitro studies may not reflect the real bond strengths in vivo, even though, a material that exhibits lower bond strength under ideal laboratory test conditions is very likely to fail clinically. Furthermore, incomplete polymerisation may also occur due to resin phase, resulting in the residual monomer component diffusing out of the cement into the oral environment, which might interfere with their biocompatibility (De Munck et al., 2005).



1.4 Glass-ionomer cements

The first glass-ionomer cement developed by Wilson and Kent (1972) was the product of acid-decomposable glass and a water-soluble acid that set by a neutralisation reaction within a clinically acceptable time (McLean, 1994). It is also defined as a water-based material that hardens by acid-base reaction between an ion-leachable fluoroaluminosilicate glass powder and an aqueous solution of poly (acrylic acid). The nature of the set cement comprised an organic/inorganic complex with high molecular weight known as aluminosilicate polyacrylate (ASPA) or glass-ionomer. Since its advent, GIC has received a varied response from clinicians. It is acknowledged for its chemical adhesion with the enamel and dentine and fluoride release (Wilson, 1989; Smith, 1998), but also been a subject of debate due to its lack of physical strength and translucency (Crisp et al., 1976; Mount and Makinson, 1982). Despite this criticism, the material has found its place in a broad spectrum of applications such as luting/lining cement; base, or dentine substitute under composite resin; sealant over an active carious lesion and a restoration. The driving force in all these applications above is indeed the chemical adhesion with tooth and the long-term fluoride release (Mount and Hume, 1998; Anusavice, 2003), Figure 1-3.

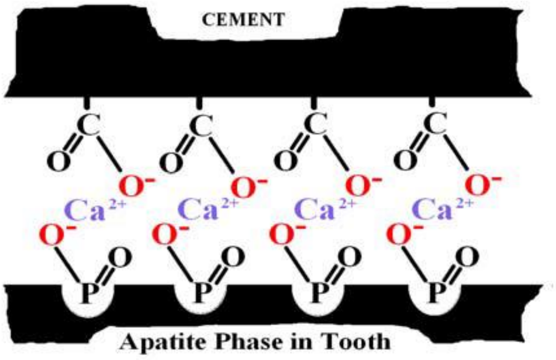


Figure 1-3 Bonding of polyacrylate cements with tooth.

1.4.1 The glass component

The glass powder has been the subject of many changes from the earliest GI formulation (ASPA-I, or G200) up to the most recent commercially available GIs. Efforts were directed to enhance the reactivity of the glasses with the polyacid liquid, the translucency of the set GI (Wilson and McLean, 1988), and control on the rate setting (Wilson and Nicholson, 1993). Structurally, the glass is prepared by sintering mixtures of powdered silica (SiO_2), alumina (Al_2O_3), cryolite (Na_3AlF_6), aluminium trifluoride (AlF_3), fluorite (CaF_2) and aluminium phosphate ($AlPO_4$) at 1100-1500°C depending on the chemical composition of the glasses (Wilson and Nicholson, 1993). The glass melt is shock cooled in water, the resultant course glass frit is ground and sieved to form a powder with a maximum particle size of 45 µm for GI restoratives and 15 µm for GI luting cements. The major concern with these early formulations were poor aesthetics due to high fluoride content, and sluggish set when used clinically, however the inclusion of tartaric acid improved setting (Wilson et al., 1976; Nicholson et al., 1988; Hill and Wilson, 1988). Consequently, a large number of glass formulations were developed as the powder components of the GIs are based on calcium aluminosilicates (SiO_2-Al_2O_3-CaO) or calcium fluoroaluminosilicates (SiO_2-Al_2O_3-CaF_2). The presence of fluoride offers lower fusion temperature, affects the rheological and setting properties of GICs, improves the translucency of the mix (Wood and Hill, 1991), and inhibits caries development when used clinically (Mayanagi et al., 2014).

Commercial glasses also contain strontium, barium or lanthanum ions to impart radiopacity (Smith, 1998). Strontium was investigated by Deb and Nicholson (1999), as an ideal candidate for replacing calcium in the glass structure without disrupting the glass network or loss of translucency (Shahid et al., 2014). Strontium can diffuse into the hypomineralised tooth tissue replacing the calcium ions in the HAp (Curzon and Losee, 1977; Ngo et al., 2006). Although combing strontium and fluoride may have advantages over either ion alone, when incorporated together, they improved the apatite crystallinity, enhancing the remineralisation process (Featherstone et al., 1983; Thuy et al., 2008). Some commercially available glasses used in GIC cements contain zinc (Chemfil Rock, Dentsply), lanthanum (Ketac Molar, 3M ESPE), or strontium (Fuji IX, GC Corporation and Ionofil Molar, Voco GmbH), and calcium fluoroaluminosilicate glasses. The reactivity of commercial glass powder is driven by chemical composition, glass fusion temperature (Wilson and Nicholson, 1993), surface treatment through acid washing (Schmitt et al., 1983) and the powder particle size (Kaplan et al., 2004).

The addition of reactive glass particles such as bioactive glass (BAG) to the GIC systems were attempted to encourage their remineralisation potential, however they yielded significantly lower compressive strengths (Yli-Urpo et al., 2005a; Yli-Urpo et al., 2005b). However, Osorio et al. (2015) showed that incorporating 2% of biosilicate (crystalline bioactive glass-ceramic) to the commercial RMGICs enhanced their microtensile bond strength (µTBS) to dentine and were demonstrated to encourage remineralisation potential. The incorporation of nano-sized hydroxyapatite and fluoroapatite (8-12% by weight) to the GIC powder enhance their mechanical properties (compressive, diametral tensile and biaxial flexural strength), working time and bond strengths to dentine (Lucas et al., 2003; Moshaverinia et al., 2008; Moshaverinia et al., 2012). Further material developments have included a newly named material: ‘Glass Carbomer®’ (GCP Dental, Mijlweg, Netherlands) which is claimed to contain nano-glass particles, hydroxyapatite/fluorapatite (HAp/FAp) nanoparticles and liquid silica. The nanocrystals of calcium fluorapatite (FAp) may act as nuclei for the remineralisation process and initiate the formation of FAp mineral as well as nanocrystals of hydroxyapatite (HAp). The glass has a much finer particle size compared to conventional GICs, giving properties that are thought to aid its dissolution and ultimate conversion to FAp and HAp. However, using "magic angle spinning" nuclear magnetic resonance spectroscopy, Zainuddin et al. (2012) have shown that the HAp in the powder is consumed during the cement formation process in this material and so may have reduced the availability for bio-mineralization (Watson et al., 2014).

Hong et al. (2008) found that adding 15% nano-β- tricalcium phosphate (β-TCP) to the GIC powder promotes protection against acid demineralisation and enhance enamel remineralisation. Calcium phosphates can induce remineralisation of the mineral-deficient dentine promoting dentine repair (Dickens et al., 2003, Dickens et al., 2004; Dickens and Flaim, 2008). These in vitro studies suggested the use of dental cements containing hydroxyapatite and/or other calcium phosphates where complete removal of the carious tissue is contra-indicated, since these additives can precipitate in the carious tissues, resulting in increased mineral content and enhanced clinical handling with improved strength and adhesion (Dickens et al. 2003; Dickens and Flaim, 2008).

Table 1-4 Components of fluoroaluminosilicate glass and their effects (compiled from Mount and Hume, 1998; Wilson and McLean, 1988)

Component	%	Effects
SiO_2	29.0	These are three essential components of the glass which fuse to form calcium fluoroaluminosilicate glass. Glasses higher in SiO_2 (>40%) are more translucent whereas those high in CaF_2 or Al_2O_3 are more opaque.
Al_2O_3	16.6	
CaF_2	34.2	
Na_3AlF_6	5.0	Complements the fluxing action of CaF_2 i.e. reduce fusion temperature.
$AlPO_4$	9.9	Improves translucency and adds body to the cement paste.
Sr, Ba, La, Salts	-------	Used to replace calcium fully or partially to give a radiopacity to the cement

1.4.2 The liquid component

Acids used in conventional glass-ionomer system (CGICs) are water soluble polymeric acids. These polyacids include homopolymers and copolymers of unsaturated mono-, di-, or tri- carboxylic acids. Of these, the most important used to date have been polyacrylic acids, copolymers of acrylic-itaconic acids and copolymers of acrylic-maleic acids (Nicholson, 1998; Smith, 1998). The number of carboxylate groups directly affects the reactivity and strength of each copolymer in GIC systems (Wilson et al., 1976; Culbertson, 2006). The structures of these acids are shown in Figure 1-4.

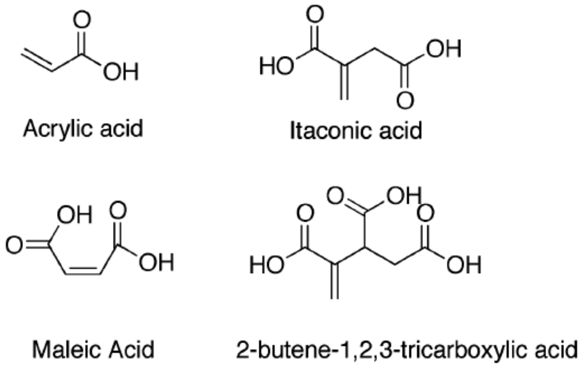


Figure 1-4 Structures of different carboxylic acids that take part in GIC liquid (Wilson and Mclean, 1988)

The molecular weight of the polyacids range between 40,000 and 60,000 for adequate mechanical properties, the higher molecular weight, the better mechanical properties, however, the molecular weight is limited by the viscosity, and some balance has to be achieved between concentration, viscosity and molecular weight (Wilson, 1989). Tartaric acid addition (5-10%) improves the compressive strength and handling properties of the GIC system by extending the workability and sharpening the setting time (Crisp et al., 1975). It reacts preferentially with the glass forming complexes to prevent the early binding of cations to the polyacid chains (Nicholson et al., 1988) and delays the formation of calcium carboxylate.

There are two major concerns associated with the GICs' liquid, firstly, the direct or very close attachment of the carboxylic acid groups (COOH) to the acrylic backbone. This interferes with the complete conversion of the carboxylic acids to carboxylate during setting reaction and affects salt-bridge formation associated with the presence of unreacted carboxylic acids that creates firmly bonded hydrogen ions via electrostatic forces. This impedes the movement of the metal ions that remain static and aren't able to migrate to carboxyl sites. The steric hindrance and the reduction in Al^{+3} di- and tri-carboxylate salt-bridges compromise cement strength and fracture toughness (Wilson and McLean, 1988; Ouyang et al., 1999). Secondly, the molecular weight of the polyacids in which the higher concentration and molecular weight enhance the flexural and compressive strengths of the resultant cement. However, this would complicate mixing, limit the shelf time, and minimise the non-structural water content which plays a critical role in the setting reaction of the GICs (Crisp and Wilson, 1976; McLean, 1991). Accordingly, higher density of the carboxylate groups can be achieved by using either low concentration of long chains polymer or high concentration of short polyacid chains. However, beyond a certain limit, flexural strength decreases since higher concentration impedes the dissolution of the acid components which deteriorate the resultant cement. It also affects the handling properties that compromise the quality of the set cement. Therefore, balance must be achieved. An optimum concentration and molecular weight of the polyacid should be selected to achieve the minimum viscosity of the liquid that leads to a higher powder/liquid ratio as well as higher mechanical properties with appropriate handling characteristics concurrently (Craig et al., 2004; Powers and Sakaguchi, 2006).

The incorporation of amino acids and other organic space maintaining chemicals likes; N-vinylpyrrolidone (NVP) and N-Vinylcaprolactam (NVC) allow the movement of the acid groups from the rigid polymer backbone. These polymeric formulations contain wider mix of pendants such as -CH2COOH, -CH-COOH and -C-COOH. They permit a higher degree of freedom for the carboxylate anions and adding -COOH groups to the matrix which encourage more salt-bridges formation which in turn enhances the mechanical and adhesive characteristics of the modified GICs (Culbertson, 2006). NVP enhances the wettability of the polymeric matrix via the presence of more hydrogen bonds. It also decreases the contact angles with dentine (Culbertson and Kao, 1994), added to the enhanced polyacrylate interactions which improves the adhesion to this substrate (Moshaverinia et al., 2009). Current research are focused towards functionalisation of the acrylic acid copolymers using organic space maintaining chemicals or functional monomers to enhance blending with the inorganic glasses for greater salt-bridge formation along with improved mechanical and adhesive properties for the GICs.

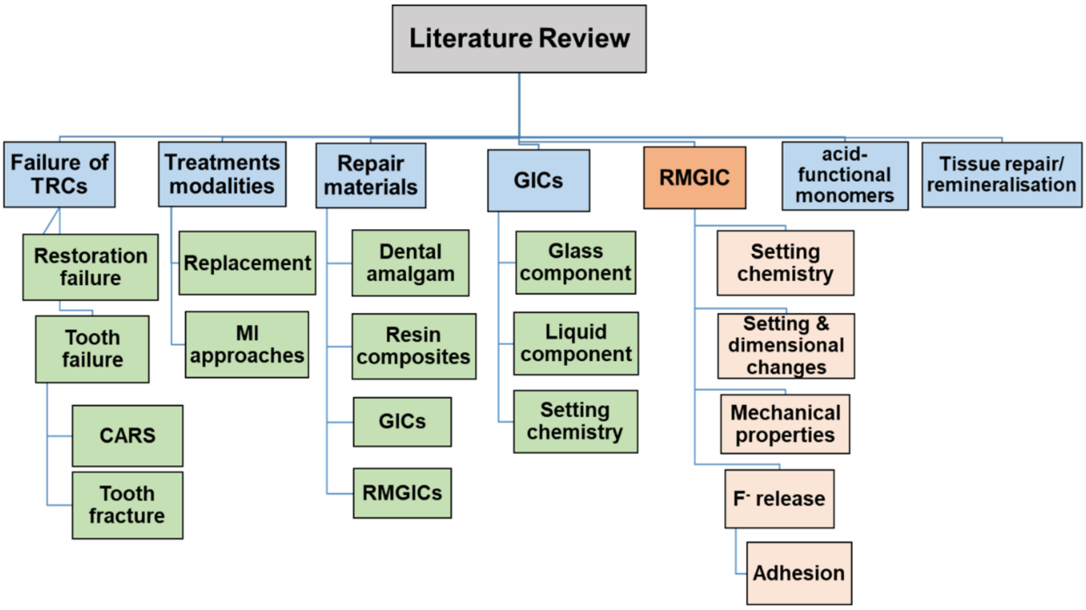
1.4.3 Setting chemistry of the GIC systems

The setting reaction of the GIC systems is complex and may vary according to their composition. Generally, it is an acid-base reaction between the polyacid liquid and the glass in which Ca^{2+} and Al^{3+} ions are liberated by the acid attack on the surface of the glass particles and ultimately cross-link the polyacid chains into a network (Wilson and McLean, 1988). The summary of the setting stages and GIC characteristics are illustrated in table 1-5. Initial setting (gelation) is due to chain entanglement as well as weak ionic cross-linking which corresponds with the viscoelastic behaviour of the freshly set material. Cook (1983) followed by Wasson and Nicholson (1991) suggested that there is no sequential release of Ca^{2+} and Al^{3+} ions. Instead, these ions and other species are liberated together with differential rates of reaction in matrix formation. As the cement matures over the first 24 h and beyond, progressive cross-linking occurs possibly with hydrated Al^{3+} ions since the sensitivity to moisture decreases and the percentage of bound water and glass transition temperature increase (Wilson and McLean, 1988). The final set structure has been presented as a complex composite of the original glass particles sheathed by a siliceous hydrogel and bonded together by a matrix phase consisting of hydrated fluoridated calcium and aluminium polyacrylates (Prosser et al., 1984; Wilson and McLean, 1988). Wasson and Nicholson, (1991) suggested that a silicate matrix is also slowly formed in the GIC in addition to the polyacrylate structures. Wasson and Nicholson (1993), Darling and Hill (1994) provide further evidence that a hydrated silicate network may be a secondary setting reaction in the GIC and it remains in the matrix phase with aluminium and calcium. Anderson and Dahl (1994) demonstrated a further significant release of Al^{3+} occurs at the freshly set stage with dissolution of the matrix. This is consistent with the fact that the increased stability and strength of GIC over time (Mitra and Kedrowski, 1994) arising from the increased aluminium cross-linking in both polyacrylate and silicate networks.

Water plays an essential role in the setting and structure of the GICs. It serves as a solvent and a medium for transporting the ions. Apart from water can also hydrate the siliceous hydrogel and metal poly (acrylate) salts (Wilson and Mclean, 1988). Water is present in the set cement in two different states depending on whether they can be removed or not by desiccation. These states have been classified as ‘evaporable’ and ‘non-evaporable’ or more commonly referred to as ‘loosely bound’ and ‘tightly bound’ water respectively (Nicholson, 1998). Loss of loosely bound water during initial setting stages retards the setting and produces surface crazing and cracks; whereas moisture contamination at this stage results in loss of soluble ions producing weak cements (Wilson et al., 1979; Causton, 1981). As the cement matures, the ratio of tightly bound water to loosely bound water increases decreasing its susceptibility to desiccation accompanied by an increase in strength and modulus and a decrease in plasticity (Paddon and Wilson, 1976; Nicholson and Czarnecka, 2008).

Table 1-5 Summary of the setting stages and GIC characteristics

The decomposition of the glass powder “ion leaching or extraction phase.”	• Ionisation of the carboxylic acid liberates protons (H^+) from the carboxyl (COOH) group. • Protons attack the surface of the glass liberating Al^{3+}, Ca^{2+}, Na^+, F^-, and $H_2PO_4^-$ ions into the aqueous phase (Wilson and Proser, 1982). • The formation of silicic acid which then condenses to form silica gel (Wasson and Nicholson 1991, 1993) • Increased pH (Crisp and Wilson 1974) and viscosity of the cement paste (Wilson and Mclean, 1988).
Gelation-precipitation of cations and anions	• Al^{3+} and Ca^{2+} forming metallic salt bridges with free (COO-) groups resulting in cross-linking of the polycarboxylate chains and ultimately leading to initial setting • The increased density hinders the movement of metal ions towards carboxyl sites, so the neutralisation reaction does not complete. • Calcium polyacrylate is responsible for the initial gelation and setting producing a clinically hard material within 4-10 minutes after mixing. • It followed by a slower formation of aluminium salts which may take up to 48 hours for completion (Crisp and Wilson, 1974; Barry et al. 1979). • Fluoride and phosphate ions form insoluble salts and complexes. Na^+ contributes to the formation of orthosilicic acid on the surface of the glass particles which converts to silica gel as the pH rises. The silica gel assists in binding the powder to the matrix (Mount and Hume, 1998). • Water loss at this stage compromises the completion of the reaction while extra moisture deteriorates the properties. • The final translucency and colour are not apparent until 24 h. • The final physical and mechanical properties are still not reached and the material continues to set up to a month
Maturation phase Hardening	• Continuation of aluminium salts bridges formation via cross-linking with the remaining -COOH acids or replacing the already crosslinked calcium ions (Wilson and Prosser, 1982; Pires et al., 2004) • Increase in the ratio of bound to unbound water (Matsuya et al., 1996). Water bounded to the silica gel which surrounds the residual core of each glass particle (Van Noort, 2014) • Higher compressive strength and modulus • Lower solubility



1.5 Resin modified glass-ionomer cements

RMGICs have similar composition to that of glass-ionomer cements, and comprise of a water-soluble polymeric acid, ion-leachable glass and water, together with the organic, amphiphilic photopolymerisable monomers and their initiation systems (Nicholson, 1998; Nicholson, 2016). They set via acid-base reaction and free-radical (visible light initiated) polymerisation, or chemical reaction when blended with the chemical redox initiators (McKinney and Antonucci, 1986, Mitra, 1991). The simultaneous acid-base reaction and photo-polymerisation of the ampiphilic monomers such as 2-hydroxyethyl methacrylate (HEMA) which copolymerised with the methacrylate functionalised poly (alkenoic acid) produce a cross-linked organic matrix (Mitra, 1991) which interpenetrates the ionic cross-linked network and thus enhances the properties (Nicholson and Czarnecka, 2008). This resin-modified cement combines the favourable adhesive and cariostatic properties of the GIC (De Moraes et al., 2016) with longer working time, command setting and enhanced early properties. They showed higher early compressive strength, diametral tensile strength, fracture toughness, wear resistance with less brittleness and solubility (McKinney and Antonucci, 1986; Mitra, 1991; Rusz et al., 1992; McCabe, 1998). Additionally, they enhanced the adhesion strength to teeth via the micromechanical interlocking and chemical chelation of polyacrylic acid to tooth tissues (Mitra, 1991). They also showed an ability to release fluoride and other ions including Na, Ca, Sr, Al, P and Si as conventional GIC but in different rates (Mitra, 1991; Coutinho et al., 2007; Czarnecka and Nicholson, 2006). Generally, the RMGICs showed a reasonable level of evidence regarding the clinical performance in many specific applications as; liners/bases, luting agents, core and restorative materials. Clinical reports for specific uses, such as class V restorations, showed reliable results regarding the aesthetics and durability (Gladys et al., 1998, Sidhu, 2010). Capsulated RMGIC shows more reliable seal against enamel, dentine and cementum compared to manually-mixed RMGIC and their correspondent conventional GIC (Hallett and Garcia-Godoy, 1993). Individual hand proportioning and mixing lead to inconsistency in the physical properties of the set cement as suggested by some authors (Billington et al., 1989; Smith and Martin, 1992).

The amphiphilic difunctional monomer (HEMA) in RMGIC dissolves the vinyl-containing polyacids which have limited water solubility (Xie et al., 1998). However, unreacted HEMA can diffuse through dentine to the pulp producing variety of adverse biological effects (Yoshikawa et al., 1994; Stanislawski et al., 1999) which compromises the biocompatibility of RMGICs in comparison to GICs (Nicholsona and Czarnecka, 2008). Moreover, the hydrophilic monomers increase the water uptake of RMGICs which might induce hydrolysis of the polyacrylate/polymer matrix that affect the properties over time (Anstice and Nicholson, 1993; Kanchanavasita et al., 1997).

1.5.1 Setting chemistry of resin-modified glass-ionomer cement (RMGIC)

The setting of GICs is described as series of overlapping stages (Wilson and Nicholson, 1993). Polyacrylic acid protons liberate metal ions and fluoride from the glass that form a silica hydrogel around the glass surface. The rising aqueous phase pH causes polysalt precipitation from migrating ions which cross-link the polyacrylic acid chains. The setting times approximate several minutes, however, further maturation occurs over extended times (Pearson and Atkinson, 1991). Conversely, resin reaction rate is much faster. However, the photopolymerisation produces diffusion-controlled polymer chains amid the formation of cross-linked matrix network (Lovell et al., 1999; Daronch et al., 2006), in which the final degree of conversion depends on the monomer mobility and diffusion (Daronch et al., 2006).

Since RMGIC's set by both acid-base and polymerisation reactions, Fig (1-5), and each mechanism depends upon reactant diffusion prior to gelation, it is plausible that the reaction kinetics and the extents of each setting reaction are influenced by each other. Nicholson and Anstice (1994) and Berzins et al. (2010) showed that both reactions compete with and inhibit one another during the early RMGI development, but the true extent is not yet known. Nevertheless, the acid-base reaction is delayed but not completely inhibited, which is facilitated by the presence of the oral humid conditions (Wan et al., 1999).

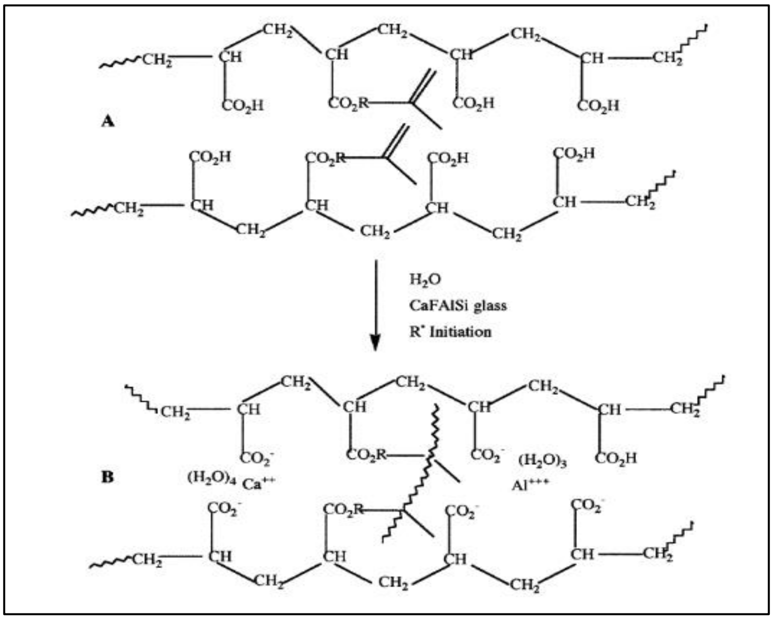


Fig 1-5 Hypothetical RMGIC structure; A: prior to VLC, B: after VLC, (taken from Culbertson, 2001)

The initial rate of acid neutralisation is slower in RMGICs in comparison to the GICs (Young, 2002). This is due to the reduced reactivity of the glass which might be coated with silane coupling agent, the partial replacement of water with monomer that reduces the ionisation of acids, and the lower polyacid levels (Nicholson, 1998; Young, 2002). Additionally, there is a reduction in the dielectric constant of the medium, due to the presence of HEMA, which affects the conformation of the polyacrylic acid that coils up more tightly than it does in water (Kakaboura et al., 1996; Wan et al., 1999), hence slowing the rates of acid-base reaction (Anstice and Nicholson, 1994).

Prior to the light curing, the acid neutralisation process is diffusion-controlled. Then, the acid-base reactant diffusion is reduced after the formation of photopolymer cross-linked network (Young, 2002) which affects the extent of the reaction. However, the exothermic enthalpy during photopolymerisation might increase the diffusion rates and accelerate the acid-base reaction. Berzins et al. (2010) showed that the delayed initiation time of the VLC enhances the rate and extent of the acid-base reaction. In contrast, the polymerisation reaction was affected by the polar nature of the GI environment (Nicholson and Anstice, 1994), added to the increased opacity due to the progression of the acid-base reaction (Wilson and Nicholson, 1993). This might increase the light attenuation and thereby hamper the polymerisation reaction.

An FTIR spectral study showed the absence of peaks at 1322 and 1300 cm^{-1} after 4 min of 20 s light exposure which assigned to the methacrylate C-O stretches that are shifted to 1273 and 1250 cm^{-1} after polymerisation due to the loss of adjacent C=C in the molecule (Hua and Dubé, 2001). This indicates the completion of polymerisation reaction in RMGICs, as shown in Fig 1-6 (Rueggeberg et al., 1990; Pianelli et al., 1999; Young, 2002). Although, the FTIR peak arising at 1633 cm^{-1} from the unsaturated -C=C-, is used to quantify the polymerisation reaction in resin composites by rationing against the carbonyl groups (Rueggeberg et al., 1990), this band cannot be applied for the RMGIC due to the presence of water (1640 cm^{-1}), which interferes with the analysis band.

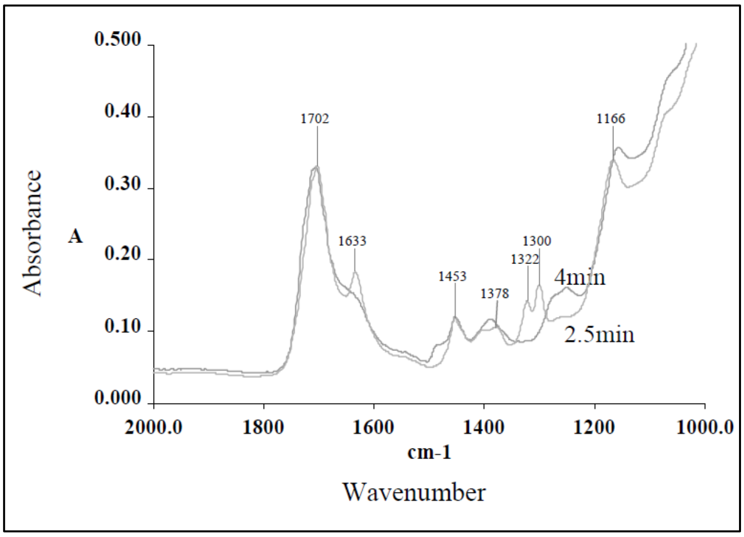


Fig 1.6 FTIR spectra of Fuji II LC at 2.5 min prior to cure and 4 min after 20 s light exposure, (Young, 2002).

The earlier VLC initiation limits the acid-base reaction and results in a material of variable structure dependent upon the time of initiation. However, the delay in light-activation decreases the HEMA conversion, as evidenced by a reduction in the polymerisation exotherm energy that is directly related to the conversion percentages (Andrzejewska et al., 2003). Berzins et al. (2010) recorded a relationship between the time of VLC initiation and the mean VLC polymerisation reaction exotherm. This relationship predicts that the delay in VLC for 3 minutes and 15 s (manufacturer's stated working time) makes the photopolymerisation reaction approximately 85% of that after immediate VLC. However, it is speculative if such polymerisation reduction will significantly affect the material properties. On the other hand, the delay in VLC is associated with the possibility of leach out of residual HEMA (Hamid et al., 1998), which compromises the biocompatibility of the RMGICs that has been questioned in many in vitro studies (Oliva et al., 1996; Geurtsen, 2000). Nevertheless, the relevant data that assess the effect of delayed VLC on the properties and characteristics of the RMGIs are limited, since most studies have evaluated RMGI materials with or without VLC.

1.5.2 Setting and dimensional changes

RMGIC exhibit volumetric shrinkage during setting similar manner to resin composites (Attin et al., 1995), which is followed by a marked expansion/swelling. It has been reported that contraction stresses as a result of polymerisation shrinkage are lower in RMGIC's compared to resin composites, nevertheless may disrupt the integrity of the tooth-restoration complexes due to gap formation at the margins (Davidson et al., 1984). There are conflicting reports on the effect of contraction stresses on the adhesive interface (Davidson et al., 1991). Irie et al. (2002) did not find a correlation between the immediate diametral shrinkage-strain, SBS to enamel/dentine and the marginal gap formation, RMGIC exhibited higher percentage of shrinkage-strain, lower SBS values to enamel and dentine than resin composite and compomer with no significant changes in the marginal gap formation. In contrast, other studies (Calheiros et al., 2004; Gerdolle et al., 2008) reported the presence of direct correlations between them. They also reported higher shrinkage strain and microleakage values in RMGICs compared to the resin composites. Of note is that these tests were performed in the absence of water, which does not take in account the the hygroscopic expansion that occurs in a wet environment and in RMGICs the initial inward cuspal contraction is reversed to an expansion (Feilzer et al., 1995; Versluis et al., 2011).

Unlike resin composites, a two-stage setting reaction occurs in RMGIC. The initial polymerisation of the resin phase sets up contraction stresses within the material whereas the slow acid-base neutralisation reaction evokes certain degree of stress relief. The presence of 10-20% of water within the liquid polymer is beneficial for the continuing acid-base reaction, which improves the mechanical strength of the cement over time. Moreover, the higher water uptake due to the presence of HEMA reduces the shrinkage stresses via hygroscopic expansion (Hinoura et al., 1993; Feilzer et al., 1995; Kanchanavasita et al., 1997). Most water gain occurs within the first week, under specific experimental conditions, this early hydration preserves the adhesion and internal coherence of the RMGIC restorations, however, it would be beneficial if both expansion and shrinkage occur simultaneously. This expansion needs to be limited to certain level, otherwise, it might damage the restoration or the tooth. Although, the clinical consequences of RMGIC swelling are unclear, but it is possible to create a considerable pressure against the cavity walls and affect marginal adaptation as reported for composites (Feilzer et al., 1988; Momoi and McCabe, 1994).

The RMGICs can absorb stresses and reduce the volumetric polymerisation contraction when applied beneath resin composites (Davidson, 1994; Ferracane et al., 2006; Mitra et al., 2009). This might be related to the viscoelastic behaviour of the RMGICs when subjected to deformation which expected to contribute to the stress relieving mechanism (Yamazaki et al., 2006). However, it is difficult to differentiate the effect of this property from the water sorption/expansion effect (El Hejazi and Watts, 1999; Cheetham et al., 2014). Feilzer et al. (1995) correlated the setting stresses of the RMGIC with the flow capacity and elastic strain of the material. The flow capacity depends mainly on the geometry of the restoration, which determines the ratio of bonded to non-bonded (free> surface of the restoration), the so-called configuration factor or C-factor (Feilzer et al., 1987). The higher the C-factor, the larger the adhesive surface, the less the possibility of the restorative material compensating for the setting shrinkage by flow.

1.5.3 Mechanical properties

The evaluation of mechanical properties of restorative materials in the mouth is essential as it can reflect on the adequacy for the intended clinical application. Although laboratory tests are helpful in predicting the clinical performance, it is possible that a material that satisfies the mechanical requirements tested in vitro, may still not exhibit a satisfactory clinical outcome. Ideally, the tests should simulate the failure of restorations intraorally, however, failures may result from a complex mixture of stresses (Tyas, 1991) that are difficult to mimic extraorally with many other confounding factors. The commonly reviewed mechanical properties for testing water based cements such as GIC/RMGICs are; compressive , flexural and diametral tensile strength, fracture toughness, microhardness (Knoop (KHN) or Vickers (VHN)), wear rate, fatigue and creep (Goldman, 1985; McKinney and Antonucci, 1987; Øilo, 1989; Hill et al., 1989; Xie etal., 2004).

Tests in compression are a reliable method that estimates the resistance of a restoration to uniaxial impact force. Compression test is widely used to identify the strength of brittle materials including GICs against deformation, crack growth, or fracture under uniaxial force. The factors that affect the CS of GIC systems are the structure, the extent of setting reaction, P/L ratio which is related to the concentration of the reinforcing glass particles in the set matrix, and the presence of voids via air inclusion or inadequate wetting of the powder particles, which is associated with the mixing mode (Fleming et al., 2006, Nomoto and McCabe, 2001). To exemplify the effect of mixing mode on compressive strength, studies on high viscosity encapsulated GIC systems reported higher compressive strength values (190-210 MPa) than the manually-mixed counterparts (130-160 MPa)(Mitra et al., 1994; Fleming et al., 2006, Nomoto and McCabe, 2001). This is atributed to the standardised powder/liquid ratio, uniform wetting of the powder particles during mixing with reduction in air inclusion during mixing, combined with the greater concentration of the glass filler (Fleming et al., 2006; Dowling and Fleming, 2009). However, Nomoto and McCabe (2001) reported a reduction in the CS of a hand-mixed product (Ketac-Molar) when mixed mechanically (182 MPa) compared to the manual mixing (152 MPa) for the same product. This indicates that other factors like the differences in composition and viscosity of the mix must be considered. However, the less viscous hand-mixed products (Ketac-Cem), in the same study, showed greater CS values (194 MPa) than those produced by mechanical mixing (Ketac-Cem Aplicap, Ketac-Cem Maxicap) (112, 119 MPa, respectively), which indicates that the lower consistency of the material might be prone to more air inclusions during the rapid mixing process in comparison to the slower manual mixing.

Generally, the failure in compression is a result of a complex stress formation in the body. However, theoretically, a material can fail only by the separation of planes of atoms *(*i.e., tensile failure), or by slipping of planes of atoms *(*i.e*.,* shear failure). Despite these limitations, the maintenance of proper compressive strength under prolonged ageing is an indication of the mechanical integrity of the restorations. Compressive strength is only indirectly related, in complex way, to these modes of failure. The tensile strength of GICs is lower than the compressive strength (McCabe, 1998). Although, the materials that fail by crack propagation favour the tensile rather than compressive loading, the direct measurements of the tensile strength are technically difficult in brittle materials. The tensile strength via diametral compression is only valid if there is no significant plastic flow. For these reasons, the measurement of flexural strength offers the best practical and reliable estimate of the tensile strength of the GICs (Prosser et al., 1986; Xie et al., 2000). The flexural strength describes the resistance to functional load without fracture and/or permanent deformation (Saskalauskaite et al., 2008) which predicts the elastic qualities of the materials when strong and durable adhesion is crucial (Darvell, 2000; Chung et al., 2004).

The biaxial flexural strength test is recommended for brittle materials when subjected to multi-axial loads, since the maximum tensile stress occur within the central loading area. Ball-on ring is widely used due to the simplicity of fixtures with minimum requirements for the alignment of specimens and the loading ball (Shetty et al., 1980). BFS is not affected by size and edge of the specimens, with less quantity of materials required that are needed to fabricate the specimens (Ban and Anusavice, 1990). This is beneficial, as it is technically difficult to prepare flaw-free long specimens without voids, which leads to uneven stress distribution within the specimen.

The flexural strength of RMGICs is affected by several factors; the composition of the cement, test methods and specimen geometry, storage medium, curing efficiency and the degree of monomer conversion. The flexural strength values of RMGICs (Vitremer, Fuji II LC) using biaxial flexural strength are higher (≃ 80 MPa) than that of four-point flexure (20-35 MPa) (Kanchanavasita et al., 1998), since the maximum tensile stress occurs under the loading ball instead of the lower surface between specimen supports in four-point flexure (Breder and Andersson, 1990). The BFS of RMGICs show less sensitivity to surface changes after storage than other tests (four-point and three-point flexure) due to the thinner specimen geometry that also leads to a better degree of monomer conversion (Kanchanavasita et al., 1998; McKenzie et al., 2003). Vitremer and Fuji II LC were reported to exhibit a BFS of ≃60 Mpa after 1 h that were found to reach the maximum (≃80 MPa) after 24 h irrespective of the storage medium (distilled water or artificial saliva), and then maintained up to a year (Kanchanavasita et al., 1998).

In comparison to the conventional GIC, RMGICs show significantly higher BFS values under all conditions and time intervals (20-40 MPa, 80-100 MPa, respectively) (Mitra et al., 1994, Kanchanavasita et al., 1998; McKenzie et al., 2003). The resinous components produce a homogeneous matrix of cross-linked poly-HEMA and polyacrylate salts, which increases the resiliency and enables RMGIC to undergo flexure without fracturing while increasing the overall strength of the matrix. Although, flexure strength tests suggest that RMGIC shows superior clinical performance to the conventional GICs. To date, there have been no clinical reports advocate the use of these materials in preference to the CGICs (Sidhu, 2010; Nicholson, 2016).

The compressive modulus is the ratio of the mechanical stress to strain in a material under compression. It indicates the stiffness and the resistance to deformation rather than the ultimate strength. The CM values is calculated from the slope of stress-strain curves. The mechanical behaviour of GIC/RMGICs in compression differs substantially as shown in Fig 1-7. The RMGICs behave like low-filler-content composite resins with appreciable permanent deformation prior to fracture as shown in Fig1-7 (curves h-j). In contrast, the CGICs are either completely brittle, Fig 1-7 (curves a-d) or exhibited a small amount of permanent deformation, Fig 1-7 (curves e-g). The compressive moduli are higher in conventional GICs (7-8 GPa) than RMGICs (3-5 GPa) (Mitra et al., 1994), while the dimetral tesile and flexural strengths are lower in the conventional systems (>20 MPa), indicating that these materials would be more prone to brittle fracture than their methacrylate-modified counterparts (40-70 MPa) (Mitra et al., 1994; Shen, 2003). On the other hand, the higher CM values indicate a greater resistance to deformation and chipping (Momoi et al., 1995). Ageing show a variable effect on the CM values among different GIC products, as some RMGICs showed enhanced CM values due to cement maturation (Lewis et al., 1992), whilst others showed a sustained or even decreased values after storage (Mitra, 1991; Mitra et al., 1994; Chun Li et al., 1999).

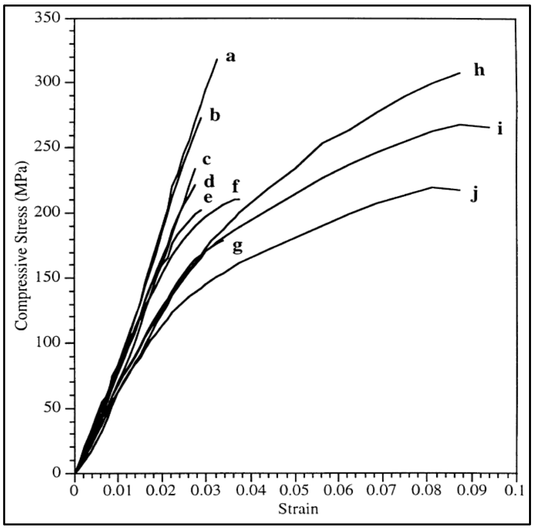


Figure 1-7 Stress-strain curves of some conventional GICs and RMGICc under compression (cures a-g GICs, h-j RMGICs), (taken from Xie et al., 2000).

Hardness is the surface resistance to plastic deformation by penetration, service scratching and finishing of a structure. It is not an intrinsic material property based on fundamental units of mass, length and time, but results from a defined measurement procedure. Knoop test is preferred for measuring the hardness of brittle materials in comparison to Vickers test, as the Knoop indenter penetrates less deeply than that of Vickers. The Knoop indenter has a pyramidal diamond shaped indenter that produces diamond shaped indentation having approximate ratio between the long and short diagonals of 7:1. The depth of indentation is about

1/30 of its length, in which only the longest diagonal of the indentation is measured (Davies et al., 1994; Darvell, 2000).

There is an agreement in the literature (Bourke et al.,1992; Matsuya et al., 1996; Yap, 1997; Kanchanavasita et al., 1998; Ellakuria et al., 2003) that the variations in MH values of the GI systems depend on the maturity status, setting reaction and the interactions with the storage medium. The early resistance is greatly influenced by the chemical composition (Prosser et al., 1986), glass structure, the concentration and molecular weight of the polycarboxylic acid (Crisp et al., 1997, Wilson et al., 1989) and the proportion of the powder/liquid ratio. Although, the beneficial effect of encapsulation of the GIC is not directly related to the surface hardness, the mixing efficiency is expected to enhance the rate of setting reaction and hence result in faster increase of surface hardness with time and possibly higher KHN for the encapsulated systems (De Moor and Verbeeck, 1998).

Xie et al. (2000) reported a correlation between KHN values and microstructure of the GICs when examined under SEM. The denser highly integrated glass particle-polymer matrix produced a greater resistance to surface indentation. Additionally, the presence of different sizes and shapes of glass particles dispersed in the polymer matrix also affected the KHN values among different commercial materials.

When the RMGICs are stored in aqueous solutions, the post-hardening process partly relies on the acid-base reaction that forms the polysalt matrix, which proceeds at a relatively slow rate (Pearson and Atkinson, 1991; Mitra, 1992). Although, the initial formation of soluble calcium polyacrylates may reduce the initial hardness values, the cross-linking via the formation of insoluble aluminium polyacrylates that take place at later stages may enhance the values over time. However, the completion of setting may not be a key factor for the property changes in RMGICs that have lower amount of carboxylic acids as the conventional counterparts (Wilson, 1990). Even though, the enhancement in strength of RMGICs is well supported in previous studies (Bourke et al.,1992; Li et al., 1995, Ellakuria et al., 2003). The post-irradiation hardening, as in composite resins (Watts, 1986), and the polymerisation which is the conversion of the methacrylate monomer to polymer may also contribute to the final hardness of the specimens (Mitra, 1992; Kanchanavasita et al., 1998). However, as with other light-curing materials, RMGICs must be adequately polymerised to obtain the optimum properties. Accordingly, the extent of setting reaction and the degree of cure of RMGICs, which relies on specimen's thickness, exposure time and the distance from the light source, affect the hardness of RMGICs (Bourke et al., 1992; Swift et al., 1995). The KHN values reported post-hardening are significantly higher than that obtained immediately after light activation or those allowed to set without irradiation (Bourke et al.,1992; Li et al., 1995, Ellakuria et al., 2003), with no pronounced effect of the delayed curing up to 180 s after mixing (Puckett et al.,1995).

RMGICs show high water uptake that tends to equilibrate within the first week (Kanchanavasita et al., 1997; McCabe, 1998). This likely explains the gradual increase in hardness that reaches the maximum during 1-7 days and is maintained up to a year. This implies that post-hardening reaction are able to overcome any plasticising effect of the absorbed water and maintain the physical properties over time (Swift et al., 1995; Uno et al., 1996; Kanchanavasita et al., 1997; Ellakuria et al., 2003). However, the cements stored in artificial saliva show a decrease in hardness values in comparison to that immersed in distilled water and fail to reach the equilibrium up to one year (Kanchanavasita et al., 1997).

Generally, the RMGICs exhibit a long-term physical integrity as compared to the GICs that are characterised by higher flexural/tensile strengths and toughness (Douglas, 1994; Mitra and Kedrowski, 1994; Li et al., 1995; McCabe, 1998). The cross-linked poly-HEMA and polyacrylate salts form a matrix that reduces the flexural modulus coupled with higher flexural strength which increases the material resiliency. Peutzfeldt and Asmussen (1992) reported a correlation between the resilience modulus and the quantitative clinical wear for RBCs and supposed that the resilience of a material influences its abrasion resistance in stress-bearing sites. This is believed to be responsible for the reduction in marginal breakdown and surface deterioration in RMGIC restorations in comparison to conventional acid-base analogues when they subjected to cyclic loading. The higher flexural and diametral tensile strengths of RMGICs coupled with the low compressive moduli indicate that these materials would be less prone to brittle fracture than their conventional counterparts (Eliades and Palaghias, 1993; Momoi et al., 1995) with higher resistance to crack propagation (Roffey, 1985).

On the other hand, the rapidly formed polymer networks of HEMA and the pendant methacrylate groups may affect the rate of acid-base reaction, apparently due to steric hindrance. The possible clinical implication of the slow reaction rate is the prolonged exposure of the dental tissues to an acidic environment. However, the pH values reported for some products after 40 s irradiation ranged from 2-5 during the initial 90 min period which are comparable to that of some conventional GICs (Tam et al., 1991). In contrast, the photopolymerisation might be enhanced via acid-base reaction through the steric orientation effect (Mitra, 1992). This is based on a preferential orientation of the polyalkenoic acid-resin chain in a way that the carboxylic groups are oriented towards the ion-etchable glass, whilst the pendant methacrylate groups are positioned to favour the crosslinking reaction (Eliades and Palaghias, 1993).

There is a contradictory effect of ageing on the mechanical properties of the GICs (Mitra, 1991; Nicholson et al., 1992; Mitra and Kedrowsk, 1994; Uno et al., 1996). The water sensitivity during initial setting of the CGIC leads to the formation of hydrated silica network which dominates the matrix and makes the cement more susceptible to brittle failure in the early stages (Wasson and Nicholson, 1993; Matsuya et al., 1996). After cement maturation, a progressive cross-linking occurs with the hydrated Al^{3+} ions since the sensitivity to moisture decreases and the percentage of bound water increases (Wilson and McLean, 1988). Then, the loss of the viscoelastic properties of the cement is associated with an increase in compressive strength and modulus (Wasson and Nicholson, 1991; Tosaki, 1994; Anderson and Dahl ,1994). In RMGIC, the formation of the silica structure is unlikely to occur to the same extent as in conventional glass-ionomers, since its formation would be hindered by the “snap-set” of the photo-curable resin. Thereafter, the enhancement in strength of RMGIC is related to the continuation of acid-base complexation and the formation of poly-HEMA and polyacrylate salts matrix surrounding the glass particles (Tosaki, 1994). Uno et al., (1996) reported that ageing of RMGIC up to 6 months has little adverse effect on the mechanical properties due to lower sensitivity to water during the early stages (Wilson, 1990; Momoi et al., 1994). In both cements, the flexural modulus increased with storage, and the rubbery characteristics observed at the early stages of setting disappeared as the cements matured.

1.5.4 Fluoride release

The prolonged and substantial release of fluoride ions from all glass ionomer materials is of major clinical significance (Guida et al., 2002; Jones et al., 2003). The fluoride is a constituent within the manufactured glass rather than being intentionally included to make the cement suitable for dental use (Williams et al., 2002). The fluoride ions arise initially from the surface of the glass powder and are held in the siliceous hydrogel matrix after setting. It is not a structural part of the matrix and has approximately the same size and mobility of the hydroxyl ion. This means that a continuing exchange of fluoride ions can occur, depending on the gradient of fluoride available in the mouth at any given time. Fluoride release usually takes place from the matrix to the adjacent environment but, in the presence of high fluoride concentration in the mouth (professional applications of fluoride as a preventive measure), fluoride ions can be taken up in to the cement again. Glass ionomer materials can, therefore, be regarded as a fluoride reservoir (Mount, and Hume 1998). However, recent studies showed that the take up of fluoride declines as the cements mature, and most of the taken fluoride remains in the cement when the concentration of surrounding fluoride falls, and it is not re-released, at least within 24 h (Nicholson and Czarnecka, 2012; Lewis et al., 2013).

Although, the mechanism of F release is not fully understood, but there is consensus on the kinetics of release that includes two phases occurring simultaneously; a short-term fluoride burst release and a long-term fluoride release (De Moor et al. 1998). Significant amount of fluoride released from GICs is during the “short-term release” phase, which occurs within a few days of placement in the oral cavity. This amount then declines sharply during the first week and stabilises after 2-3 months. The long-term release of fluoride is substantially lower, but it appears to be sufficient to prevent caries (Mount and Hume 1998). The process of diffusion seems to be associated with the long-term release of fluoride from glass ionomers (Forsten, 1995; Williams et al., 1999).

Fluoride release is affected by the composition of glass and polyalkenoic acid, the relative proportions of constituents in the cement mix, mixing process and the elution medium (De Witte et al., 2000). Many studies have reported an increase in fluoride release in acidic environments due to surface degradation (De Moor et al. 1998; Czarnecka et al*.* 2002; Shahid et al., 2008) which enhances the anticariogenic properties of GIC. Glass ionomers release more fluoride in deionised water than in artificial saliva (El Mallakh and Sarkar, 1990; Williams et al., 2002), which may be attributed to the higher ionic strength of the artificial saliva or the formation of insoluble calcium fluoride layer on the surface of the cement.

The set matrix of RMGIC is composed of the ionomer salt hydrogel and polymer in which fluoride ions might be firmly encapsulated by resin matrix that might reduce the rate of fluoride release in an aqueous environment (Wilson, 1990; Momoi and McCabe, 1993). Nevertheless, studies revealed an equivalent amount of fluoride release from both RMGICs and conventional GIC (Momoi and McCabe, 1993). The most rapid release occurs during the first week in both cements; RMGICs reach an equilibrium value after a month, while in GIC, fluoride release continued up to 3 months. In GIC, the rate of fluoride release depends on the formation of complex fluorides (Crisp and Wilson, 1974) whilst for RMGICs, in addition to this factor, the type and amount of resin used for the photochemical polymerisation reaction may affect the rate of fluoride release. However, some studies reported higher fluoride release in RMGIC than the conventional GICs (Mitra, 1991, Kato, 1993; Forsten, 1995; Robertello et al., 1999). They suggested that poly-HEMA can absorb sufficient water to enable diffusion of the fluoride ions otherwise it will be firmly encapsulated within the polyacrylate matrix. RMGICs can also maintain fluoride release if topped up via topical fluoride applications (Forsten 1995; Takahashi et al., 1993).

Although there is no convincing evidence of the required levels of fluoride to produce a therapeutic effect, RMGIC showed a caries inhibition effect equivalent to a conventional GIC when tested in vitro (Glasspoole and Erickson, 1993; Momoi and McCabe, 1993). However, the documented values for fluoride release vary considerably from one study to another. This might attributed to the lack of uniformity in specimen shape, experimental regime, nature of the aqueous environment used and even the units used to express fluoride release. Therefore, despite a large number of reports, it is difficult to compare values directly (El Mallakh and Sarkar, 1990).

Fluoride is not the only ionic species to be released by GIC/RMGICs. Studies have shown that Na, Al, P and Si are also released in neutral conditions (Forss, 1993; Czarnecka et al., 2002; 2007b; Nicholson and Czarnecka, 2008), either as free positively charged ions or as negatively charged oxy-ions (e.g. PO_4^{3-}). Ca^{2+} ions are also released under acidic conditions (Czarnecka et al. 2002). The release of all ions is higher under acidic conditions than neutral ones and is associated with a buffering effect. This results in a strong, durable interfacial zone that enhances the adhesion of the cement to the tooth, and contributes to the long-term retention of glass-ionomer restorations. This property is of a clinical advantage in the presence of active caries, which typically has a pH of about 4.9, greater amounts of fluoride ion are released from glass-ionomers, and this may have a preventive effect on the decay process (Czarnecka et al., 2007b). However, low pH is also associated with the release of other ions like aluminium or hydrogen which might form stable complexes with fluoride. Experimental studies (Czarnecka et al., 2002) demonstrated that, under acidic conditions, most if not all of the fluoride released from glass-ionomer cements is bound in complexes. Lewis et al. (2013) revealed that the complexation of fluoride with aluminium or protons does not interfere with the fluoride uptake or even might enhance it. Accordingly, they showed that complexed fluoride in acidic solutions will interact with hydroxyapatite much as free fluoride does under neutral conditions, and thus will be effective in protecting the tooth against further demineralisation (Lewis et al., 2013).

1.5.5 Adhesion

Glass-ionomers remain the self-adhesive material to tooth tissue even without surface conditioning. However, pre-treatment with weak polyalkenoic acid conditioner have been reported to improve their bonding efficiency (Van Meerbeek et al., 2006). This is attributed to the cleansing and partial demineralisation effects which increases the surface area and produces micro-porosities for micro-mechanical interlocking. Additionally, there is the potential chemical interaction of the polyalkenoic acid with residual hydroxyapatite, Figure 1-8.

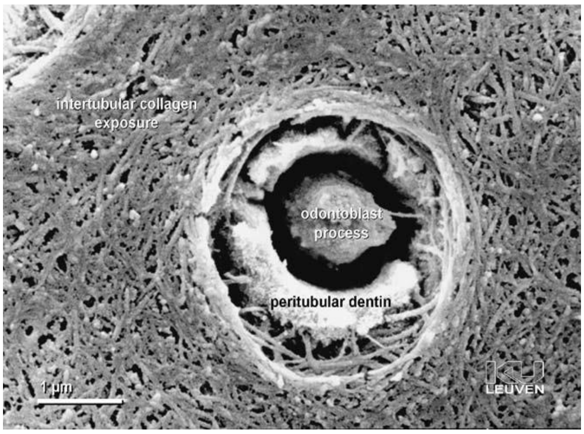


Figure 1-8 Fe-SEM photomicrograph illustrating the effect of a polyalkenoic acid conditioner (GC) applied for 10 s on dentine. The intertubular dentine was exposed, HAp was remained as receptors for additional chemical interaction. Micro-pores were created to enable micro-mechanical interlocking through hybridisation (taken from Van Meerbeek et al., 2006).

Initially, the GIC adheres to tooth surface via hydrogen bonds facilitated by the hydrophilic nature of both interfaces. Bonds are proposed between the free carboxylate groups of the cement and the layer of tightly bound water at the surface of the mineral phase (Hinoura et al., 1992). These hydrogen bonds are gradually replaced by genuine ionic bonds formed by calcium ions in the tooth and carboxylate groups in the polymer (Yoshida et al., 2000). The auto-adhesion of GIC to tooth tissue based on two inter-related phenomena; micromechanical interlocking achieved by shallow hybridisation with the micro-porous and hydroxyapatite-coated collagen fibril network (Lin et al., 1992; Pereira et al., 1997; Mitra et al., 2009). Secondly, the self-adhesion mechanism in which a chemical bonding involves the formation of ionic bonds between the carboxylate functional groups and the calcium ions of hydroxyapatite that remained around the exposed surface collagen (Van Meerbeek et al., 2006). The ions from both interfaces travel in opposite directions via a diffusion process towards the interfacial zone creating an ion-exchange layer (Ngo et al., 1997; Tyas and Burrow, 2004; Ngo et al., 2006). This layer is consisted of calcium and phosphate ions from HAp, and aluminium, silicic, fluoride and calcium and/or strontium ions (depending on glass composition) from the GIC (Sennou et al., 1999), which suggests the evidence of a genuine chemical union between both interfaces, as seen under SEM (Figure 1-9). This was observed experimentally on the hydroxyapatite using X-ray photon spectrometry (Yoshida et al., 2000; Van Meerbeek et al., 2003) and also on enamel and dentine (Lin et al., 1992; Nezu and Winnik, 2000; Fukuda et al., 2003). However, the interpretation using XPS with high vacuum technique is critical since the true chemical bonds to the tooth surfaces formed through a strongly adherent layer of water. The extent to which such bonds can form in vivo is unclear (Nicholson, 2016). Some GICs show a ‘gel-phase’ at the interface, as shown by TEM (Fig. 1-10, A) and AFM (Fig. 1-10, B). The correlation of TEM and XPS data elucidated that the gel phase confirmed the formation of a calcium polycarboxylate salt from either the polyalkenoic acid conditioner or the GI material itself (Inoue et al., 2001). This phase intermediates the shallow hybrid layer (0.5-1 µm) and the GIC matrix binding the two interfaces firmly with high resistant to acid attack (Tyas and Burrow, 2004). In µTBS testing, the interface typically fractures above the gel phase within the matrix of the GIC (Van Meerbeek et al., 2006). The adhesion of glass-ionomers to enamel is stronger than that to dentine which suggests that bonding to the organic phase of the tooth are less important. However, the collagen contains both amino and carboxylic acid groups that might interact with the carboxylate groups producing chemical bonding to collagen (Beech, 1973). However, such bonds are not particularly important in the mechanism of adhesion of GIC to the tooth (Powis et al., 1982; Nicholson, 2016).

Although there are no comparable studies on the RMGICs, it is expected that they are equally capable of forming such layers (Nicholson, 2016) as they release similar ions under similar conditions (Czarnecka and Nicholson, 2006). RMGICs provided additional bonding mechanism that is comparable to the bonding analogous of resin composites via resin tags into enamel and the establishment of a hybrid layer into dentine. However, the experimental evidence seems equivocal. Some studies supported the presence of resin tags intruded into the dentinal tubules (Lin et al., 1992; Friedl et al., 1995; Pereira et al., 1997), while others not (Sidhu and Watson, 1998; Sidhu et al., 2002). Alternatively, bonding via an ion exchange layer (Lin et al., 1992) and ionic bonds are well supported (Ramos and Perdigao, 1997).

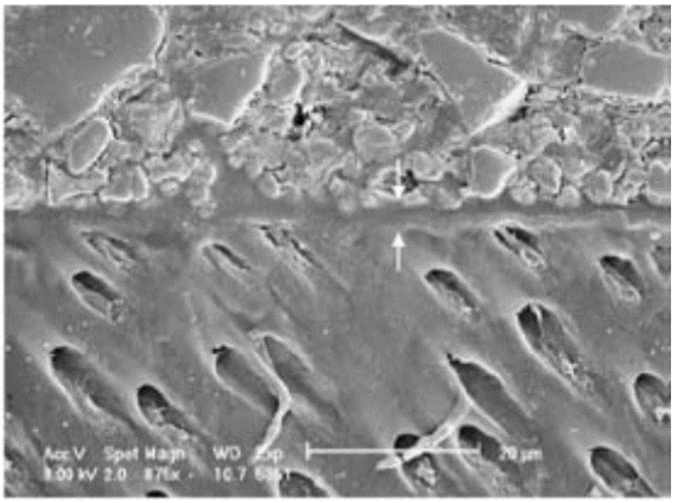


Figure 1-9 A bonded specimen of GIC to demineralised dentine. The ‘acid-base resistant layer’ can be observed at the interface (arrows). This has also been referred to as the ‘ion-exchange layer (taken from Tays and Burrow, 2004).

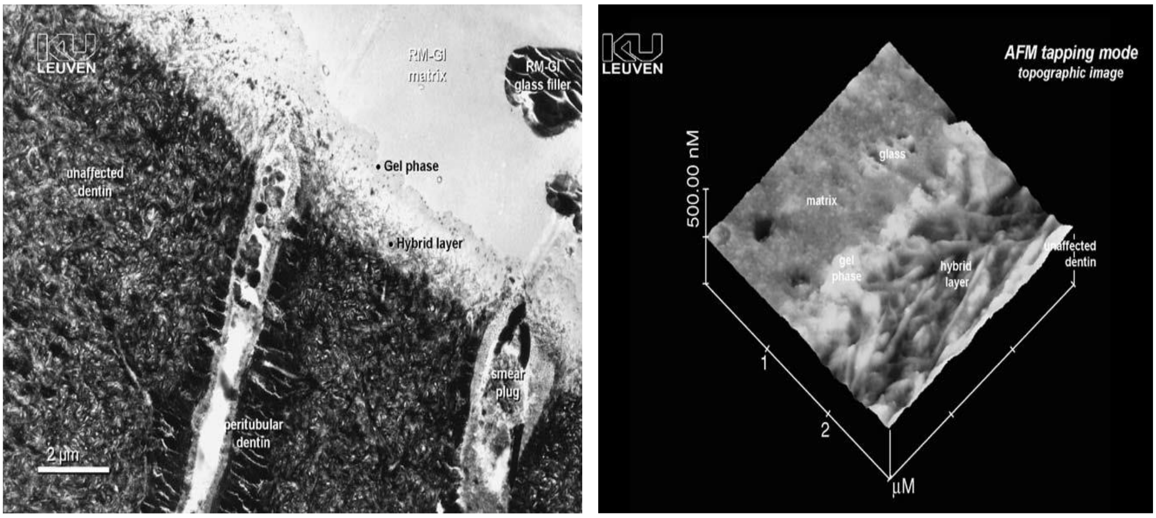


Figure 1-10, A: TEM photomicrograph through the GIC-dentine interface illustrating the two-fold structural appearance of a GIC-dentine interface resulting from the application of RMGIC adhesive (Fuji Bond LC). On top of the hybrid layer there is an amorphous grey gel phase represents the morphologic manifestation of the reaction product formed through interaction of the polyalkenoic acid with calcium that was extracted from the dentine surface. Fig B: Atomic force microscopy photomicrograph illustrating the 3D-topography of the interface between the (Fuji Bond LC) and dentine (taken from Van Meerbeek et al., 2006).

Debonding tests are divided into qualitative screening and quantitative tests. They measure the bond strength, mode of failures, load capacity or the durability of bonding (Øilo, 1987). The quality tests can be tensile, shear, torsion, cleavage, pull/extrusion or 4-point bending (Øilo, 1987; Pashley et al., 1995, 1999). Shear strength is the easiest among these tests, in which the load can be distributed in lap-shear, blunt-end shear bar, or interfacial wire loop in shear. The tensile tests, theoretically, is expected to develop more uniform stress distributions if there is a correct alignment between the adherent interfaces (Pashley et al., 1995). However, stress distributions in such tests are reported to be not uniform (Van Noort et al., 1989; 1991). This is because of the complex elastic and plastic deformations of adhesive materials, demineralised surfaces and the mineralised dentine subsurface which occur simultaneously. True interfacial testing, whether in shear or tension, becomes a cleavage test as soon as the first crack begins to propagate from defect, void or other source of stress concentrations. The assessment of bond strength of the GICs to enamel and dentine is complicated by the brittle nature of the cement. Both shear or tension modes can be used. The tensile bond strength is considered to be more closely replicating the pattern of load that cement experienced under clinical conditions. However, SBS still considered as useful tool for testing the bond strength to GICs with all inherent limitations (Sudsangiam and Van Noort, 1999; Armstrong et al., 2010). The quantitative bond testing includes the determination of the fracture toughness or the energy of fracture (O’Brien and Rasmussen, 1984; Tam and Pilliar, 1993). This is done by measuring the toughness of the bulk materials or the interfacial bonds between adhesive resins and tooth structure (Harashima et al, 1988; Tam and Pilliar, 1993).

The finite element analysis (FEA) is reported as a useful tool to predict stress distributions within teeth and at the tooth-restoration interfaces (Van Noort et al., 1989; 1991). This modelling requires the knowledge of the strength and elastic moduli of the materials vs. dental tissues. Then, the three dimensional stress distribution within these structures can be calculated during various types of loading. However, there is a difficulty to measure the large stress gradients that develops at the bonded interfaces because of lack of knowledge of the elastic modulus of the demineralised dentine collagen, resin-infiltrated demineralised dentine matrix, resin tags and length of resin tags. Otherwise, the FEA may provide an evident insight into the dynamics of the bonded interfaces during the application of cyclic loading, thermal gradients or other types of stressing.

Apparently there are large differences in the stress distributions at the tooth-restoration interfaces based on the differences in the chemical and physical properties of both surfaces which lead to different modes of fracture. Generally, these failures are classified visually using naked eye or dissecting microscopes. However, adhesive failures from the smear layer-covered dentine revealed that the failure is cohesive through the smear layer rather than adhesive when the specimens are carefully examined under SEM (Tao and Pashley, 1988). Thus, there is a danger in classifying the mode of failure visually, even though, such visual classification is helpful in providing the overall descriptions of the obvious modes of bond failure, especially the cohesive fractures (Pashley et al., 1995). The failure in GIC systems is mainly cohesive within the cement rather than adhesive (Glantz, 1977; Tay et al., 2001; De Munck et al., 2005) which might represent the tensile strength of the cement rather than the true adhesive bond strength. These considerations showed that the determination of the real strength of the adhesive bond is difficult, and the reported values in the literature may not be a real representative for the true strength of the adhesive bond (Tyas and Burrow, 2004).

Fracture strength is given per unit area which necessities the standardisation of surface areas of the adherent surfaces, and testing methods (Söderholm, 1991; Pashley et al., 1995). Surface area is correlated to the stress distribution, the smaller surface areas produce higher bond strengths than the larger bonded surface areas, which show cohesive failures in dentine at relatively low bond strengths (Smith and Cooper, 1971; Pashley et al., 1995; Armstrong et al., 2010). This is also true for the tensile bond strengths in which very small surface areas (5.0-1.2 mm^2) produce higher bond strength values and failures that are mostly adhesive (Sano et al., 1994). In brittle materials, the larger cross-sectional areas contain more defects than smaller specimens. Similarly, larger dentine surface areas showed air bubbles, phase separation and surface roughness at the interfaces which lead to non-uniform stress distributions. The finite element stress analyses indicate that both tensile and shear bond strength measurements are highly dependent upon the geometry of the interface, the nature of load application, and the presence or absence of adhesive flash, etc. Accordingly, the probability that the defects may dominate the propagation of the fracture is very high because of nonuniformity of the interfacial stress distribution, especially at the edge of the interface. This may initiate fractures at the defects, resulting in lower bond strengths than might be measured in smaller samples (Van Noort et al., 1989; 1991).

The use of microtensile bond strength test with small surface areas is expected to improve the stress distributions and interfacial bond strengths with a possibility of more adhesive failures. It permits the measurements of the small and irregular shaped surfaces, and the regional bond strengths. It gives the advantage of calculating means and variances from a single tooth. However, this test is technically demanding with a difficulty to measure the low bond strength values (< 5 MPa) with a possibility of rapid dehydration of small samples.

Nevertheless, all mentioned adhesion tests, while not perfect, have contributed to the development of the improved adhesive systems and techniques (De Munck et al., 2005).

There is a real need for modifying the standards regarding the substrate and various steps in bonding and testing methods, since the newest products showed high intrinsic strength that lead to cohesive failures in the substrates rather than the bonded interfaces. Conventional bond testing methods can no longer be useful to detect the further improvements in the products development or bonding procedures. These tests should be simple to use, bond equally well to enamel, superficial or deep dentine, and be relatively insensitive to moisture. This should permit more uniform, consistent, dentine bonding which will be of significant benefit to the adhesive dentistry (Armstrong et al., 2010).

A new rational approach was proposed by Van Meerbeek et al. (2010) is called mold-enclosed shear bond strength (ME-SBS), Figure 1-11, b. This technique is expected to provide evenly distributed load as close as possible to the adhesion zone of the adherent. It exhibited significantly higher mean µSBS values and adhesive failures when used in bonding adhesives to metals (Cheetham et al., 2014). However, the bonding performance to the dental tissues and ceramics is not reported yet. Later, Jin et al. (2016) suggested a novel concept called a lever-induced mold-enclosed shear bond strength (*de novo* LIME-SBS) which modifies the conventional SBS settings. This technique applied a cylindrical adherent bonded to block shaped adherend with enclosed mould through adhesive. Then, a fulcrum is added into the SBS configuration at the mid-point between the load and the adhesive interface in which the load is applied at a relatively far distance from the adhesion zone, as shown in Fig 1-11, c. This method is suggested to be an effective tool in evaluating the bond strength under true “shear” mode eliminating the tensile stress at the interface as compared to the conventional SBS. The pattern of stress distribution is reported to be similar in titanium-adhesive, dentine-adhesive and porcelain-adhesive models.

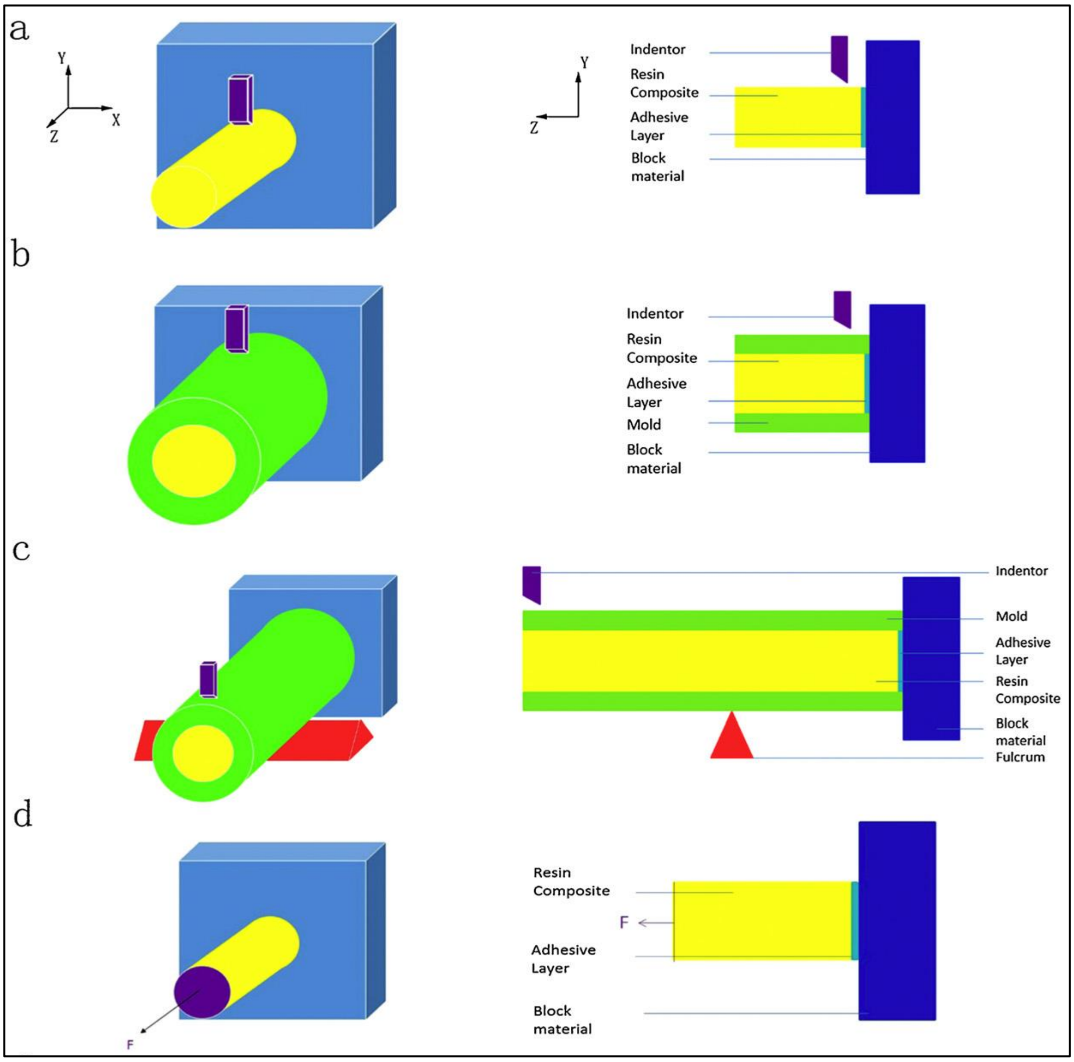
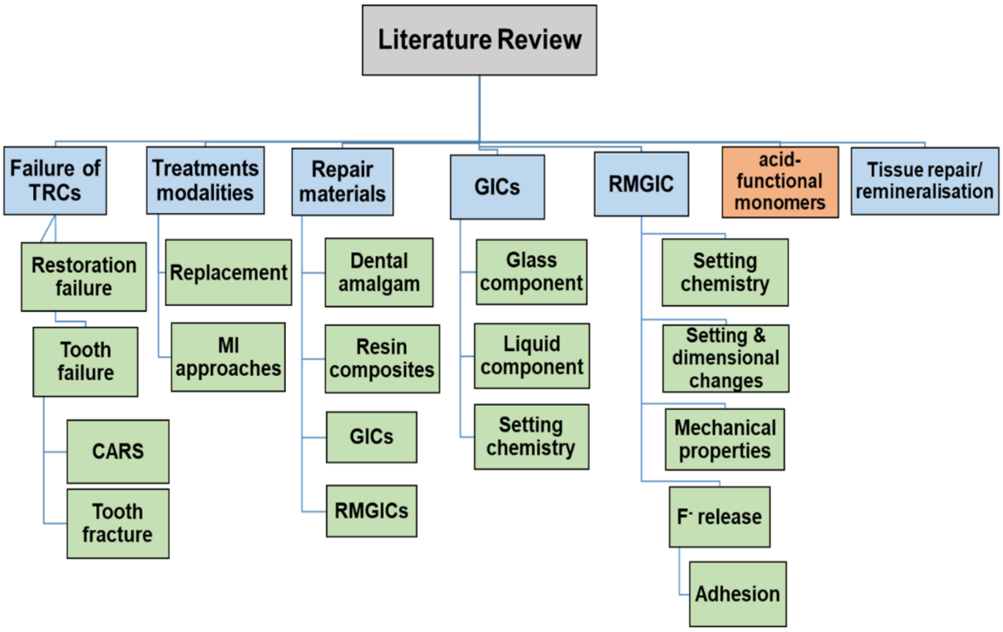


Figure 1-11 The schematic picture of (a) SBS test, (b) MESBS test, (c) LIME-SBS test, (d) TBS test, (Taken from Jin et al., 2016)



1.6 Phosphate methacrylated acid-functional monomers

Acidic-functional monomers are predominantly (meth) acrylate monomers with either carboxylic acid groups such as, 4-methacryloxyethyl trimellitic anhydride (4-META), pyromellitic glycerol dimethacrylate (PMGDM), or phosphoric acid groups like, 2-methacryloxyethyl phenyl hydrogen phosphate (Phenyl-P), 10-methacryloxydecyl dihydrogen phosphate (MDP), bis (2-methacryloxyethyl) acid phosphate (BMP) and dipentaerythritol pentaacrylate monophosphate (Penta-P). The acidic groups in these monomers (carboxyl, phosphate or phosphonate) can release one or more protons which demineralise and simultaneously chemically integrate to the tooth surface (Van Landuyt et al., 2007).

There are many monomers containing phosphates or phosphonates that have been developed specifically to demineralise enamel/dentine and form stable CaP salts. The chemical interaction with the tooth tissues (HAp and collagen), added to the micromechanical interlocking promote bonding strength, prevent nanoleakage and thus prolonging the intraoral lifetime of adhesive restorations (Yoshida et al., 2000, 2004; Fu et al., 2005; Inoue et al., 2005). According to the ‘Adhesion-Decalcification’ concept, functional groups are either decalcified or bonded to the tooth substrate (Yoshida et al., 2001; Yoshioka et al., 2002). Firstly, the acidic group ionically interacts with calcium in HAp. Depending on the hydrolytic stability of the resulting calcium-monomer complex, these ionic bonds may either decompose and demineralise the tooth surface, or remain stable and chemically bonded to calcium, as shown in Fig 1-12.

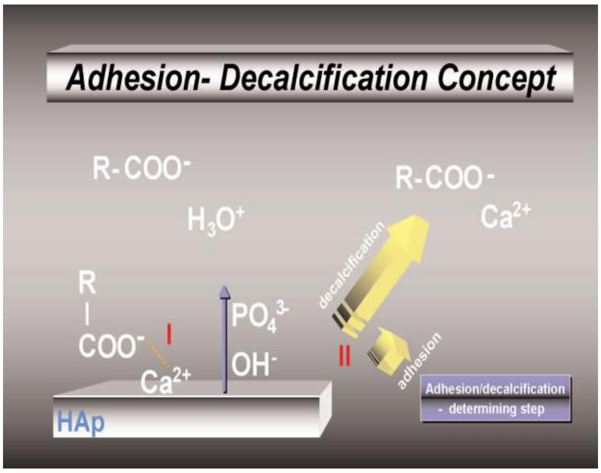


Figure 1-12 Schematic drawing presenting the “Adhesion-Demineralization model” that explains why molecules that contain functional carboxyl groups either adhere to or decalcify hydroxyapatite tissues (Yoshida et al, 2001).

Functional groups are ranked upon their chemical bonding potential (Yoshida et al., 2004) which referred to the differences in the chemical structure and polarity that contributes to the wetting behaviour of the self-etch adhesives (Nakabayashi et al., 1982). The differences in the functionality and polarity of these monomers affect the stability of the Ca/P complexes. For example, the relatively hydrophobic MDP monomer can produce hydrolytically stable complexes, whereas 4-Met and Phenyl-P produce complexes that have limited stability to dissolution (Yoshida et al., 2004; Inoue et al., 2005; Van Landuyt et al., 2008). The concentration of the acidic monomer should be balanced to achieve an acceptable degree of self-etching character and chelation to enamel and dentine whilst avoiding the excessive hydrophilicity in the final polymer matrix that causes swelling and compromises the mechanical strength and the dimensional stability.

10- methacryloxydecyl dihydrogen phosphate (10-MDP) monomer is the main component of most currently used self-etch adhesives which showed effective bonding capacities in most laboratory and clinical studies. It enhances the interfacial strength via the formation of stable bonds to cations. This is mainly due to its amphiphilic structure, as the phosphate group represents the polar moiety while the spacer group that is composed of saturated carbon chain is the apolar moiety. This configuration produces a hydrolytically stable monomer with a potential to interact with the hydroxyapatite and collagen chains producing stable CaP salts that improves the chemical bonding to dentine and enamel (Van Meerbeek et al., 2006; Yoshida et al., 2012). Such adhesives are considered mild acidic as they partially demineralise the hybrid layer retaining the hydroxyapatite crystals for chemical bonding (Van Landuyt et al., 2008). Additionally, MDP improves the bonding to dental alloys through the chemical union with the oxide layer at the alloy surface, and to other substrates including; zirconia, noble and non-precious metals, and silica-based ceramics without the need for dedicated and separately placed primers such as silane, metal and zirconia primers (Dos Santos et al., 2006; Blum et al., 2012; Balkaya et al., 2018).

The self-adhesive resin cements contain acidic and hydrophilic monomers in their composition, which simultaneously demineralise and infiltrate enamel and dentine, resulting in strong bonding. Therefore, they require no conditioning or priming pretreatments of the tooth substrate. These hybrid cements are used for the cementation of inlays, onlays, crowns, posts and veneers. They utilised acid-functionalised methacrylate monomers within a polyalkenoate matrix allowing the chemical bonding to the tooth/or restorations (Ferracane et al., 2011). The type and concentration of the acidic groups in the functional monomers, combined with the moisture content reduce the initial pH to 1.5-3 of the freshly mixed cement, which is certainly acidic enough to demineralise the hard tooth surfaces. Then the pH subsequently rises on the onset of the acid-base reaction (Ferracane et al., 2011). The self-adhesive resin cements are close to compomers in nature, while they differ in the concentration of the acidic monomer and lower filler content. In addition, the limited luting cements’ thickness provokes an efficient hydration to the substrate thus facilitating the ionisation of the acidic monomers followed by acid-base neutralisation reactions with the teeth and the basic fillers. Unlike compomers, the self-adhesive resin cements produce a reasonable degree of unassisted adhesion to dentine, whilst the direct bonding to enamel presents a greater challenge (De Munck et al., 2004). These cements contain sodium fluoride or related salts in their composition as a source for the fluoride release, however, the clinical evidence for fluoride release is not reported yet (Ferracane et al., 2011).

RelyX^{TM} Unicem is one of the self-adhesive resin cement that contains phosphate acidic-functional groups that are claimed to react with the hydroxyapatite of the tooth, resulting in additional retention through chemical bonding (Gerth et al., 2006). Moreover, these acidic monomers are claimed to interact chemically with the basic inorganic fillers of the cement, leading to an additional acid-base setting reaction, apart from the free radical polymerisation of the material (Vrochari et al., 2009). However, many studies report that the bonding effectiveness of RelyX^{TM} Unicem is characterised by low demineralisation capacity, an interaction with the superficial dentine only, and failure of resin tags formation (De Munck et al., 2004; Yang et al., 2006; Al-Assaf et al., 2006). Even though, the chemical interactions have been confirmed via XPS, and the bond strength to dentine is comparable to other widely used resin systems (De Munck et al., 2004; Behr et al., 2006). In contrast, the bonding to enamel (De Munck et al., 2004; Hikita et al., 2007) and root dentine (Walter et al., 2005; Goracci C et al., 2006) are less effective.

The setting pattern of the self-adhesive cements is closer to other hybrid cements including RMGIC and polyacid-modified resin (compomers). This is initiated by free radical methacrylate polymerisation as the primary reaction mode activated by the chemical or photochemical routes. Initially, the cements are hydrophilic to facilitate the wetting and adaptation to the tooth surface. Then, the system becomes more hydrophobic upon acid-base reaction between acidic monomers and the calcium from the HAp of the tooth tissue or the metal oxides of the ion-leachable glass. This seems to be beneficial in hindering the post-cure swelling and material deterioration. The glass fillers are composed of combinations of fluoroaluminoborosilicate glass, strontium calcium aluminosilicate glass, quartz, colloidal silica and ytterbium fluoride. The partial surface dissolution of the acid-soluble glass, neutralise the acidity and enhance the release of sodium, calcium, silicate and fluoride ions that can either take part in the setting reaction or be released locally. Whilst the phosphate groups bind to the calcium in the HAp and stabilise the attachment between the methacrylate network and tooth structure. The dynamic dimensional changes after curing are compensated by the expansion effects of the acid-base reaction (Spinell et al., 2009).

There is agreement in the literature regarding the ability of the phosphorylated monomers to promote adhesion to different restorative and prosthetic materials that are used for indirect restorations. The phosphate ester monomer of the MDP can bind directly to the metal oxides of chromium, nickel, aluminium, tin, titanium, and zirconium via their hydroxyl groups (Wada, 1986; Yoshida et al., 2005). However, these chemical reactions did not maintain their strength after thermal cycling (Yoshida et al., 2005). Similarly, the self-adhesive resin cements revealed higher bond strength to air-abraded or silica coated zirconia when compared to the GIC, RMGIC and zinc phosphate cement (Piwowarczyk et al., 2005; Blatz et al., 2007). In addition, the bonding of noble alloy crowns to titanium abutments was higher when a self-adhesive resin cement was used in comparison to polycarboxylate, GIC, zinc phosphate and zinc oxide eugenol cements (Wolfart et al., 2006). They enhanced the seal and marginal integrity with less microleakage at the tooth/restorative interfaces. Rosentritt et al. (2004) and Piwowarczyk et al. (2005) demonstrated that the microleakage of the RelyX^{TM} Unicem is equivalent to the etch-and-rinse and self-etch resin cements (Variolink II, Panavia F, respectively) after thermal and mechanical cycling. In contrast, RMGI (Fuji Plus) and self-adhesive compomer cements showed poor marginal adaptation with greater leakage. The carboxylic acid groups of RMGIC with conditioning allows a sufficient seal at the tooth interface due to chelation with the calcium ions but it is insufficient at the restorative interface (all-ceramic inlays). This limited integrity might be attributed to the polymerisation shrinkage, water sorption, plasticity or hygroscopic expansion of the cement (Diaz-Arnold et al., 1999) which leads to de-bonding of the restorations. Capsulated delivery may improve the marginal integrity due to greater homogeneity and a smaller number of inclusions or voids (Mitchell and Douglas, 1997).

The self-etching properties of the self-adhesive cements without using surface pretreatment retain the smear layer on the tooth surface and hinders the creation of resin tags in the dentine tubules. The preserved or modified smear layer may act as a natural barrier which seals the tubules but is associated with incomplete penetration of resinous cement which might weaken the resulting bond strength (Koibuchi et al., 2001) associated with adhesive failures either above or below the hybrid layer (Kiyomura, 1987). However, the use of adhesive cement containing methacrylated phosphoric groups showed an enhanced bond strength based on the complexation reaction to the HAp. The phosphoric acid may cause a slight etching to dentine with the presence of water that generated during neutralisation which moisten the tooth surface and improve the inter-diffusion into the smear layer which enhance the bond strength (Irie et al., 2002). In contrast, the poor marginal seal was reported at enamel interfaces without using phosphoric acid enamel etch (Frankenberger et al., 2008). This is consistent with the reported minimal ability of the self-adhesive and self-etching resin cements to etch and subsequently interact with enamel substrates compared to etch-and-rinse adhesives. Leakage and marginal gaps were also reported when self-adhesive cement bonded to inlays, crowns (Piwowarczyk et al., 2005) and porcelain veneers (Ibarra et al., 2007) unless a separate etch was previously applied (Ibarra et al., 2007).

The presence of functional monomers in both self-etching and self-adhesive cements may interfere with some of the photo-initiator systems that lead to a significant decrease in the degree of conversion %DC (Vrochariet al., 2009). This decrease produces inferior clinical performance in terms of ultimate hardness (Sobrinho et al.., 2000), fracture toughness (Ferracane and Berge, 1995), wear resistance (Ferracane et al., 1997), elastic modulus (Harris et al., 1999), solubility and hydrolytic degradation (Söderholm et al., 1984), as well as biocompatibility. Camphorquinone CQ is the most widely used photo-initiator system in the visible light-curing restorations. It photoinitiates the polymerisation process at a relatively low rate, therefore co-initiators like tertiary amines are added to accelerate the polymerisation (Jakubiak et al., 2003). It has been previously reported that acidic moieties in the functional monomers can affect the polymerisation reaction in all-in-one adhesives (Tay et al., 2001a). This is because of the neutralisation of the tertiary amines that lose their activity as a reducing agent in the photochemical redox curing mechanism, thus hampering the initiation reaction. The reduced %DC in many self-etching and self-adhesive cements raises the apprehension whether it is possible to use these materials successfully in clinical applications. Increasing the irradiation times further than the recommended by manufacturers could potentially lead to higher %DC, if the light is not completely blocked by overlying restorations (Vrochariet al., 2009). To overcome the incompatibility between the acidic monomers and amine initiator, proprietary activator/initiator systems should be used like those contain sodium aryl sulfate or aryl-borate salts (Suh et al., 2003). However, the use of different type of initiation systems may produce a different polymerisation behaviour, which may involve low initial %DC values (Tezvergil-Mutluay et al., 2007).

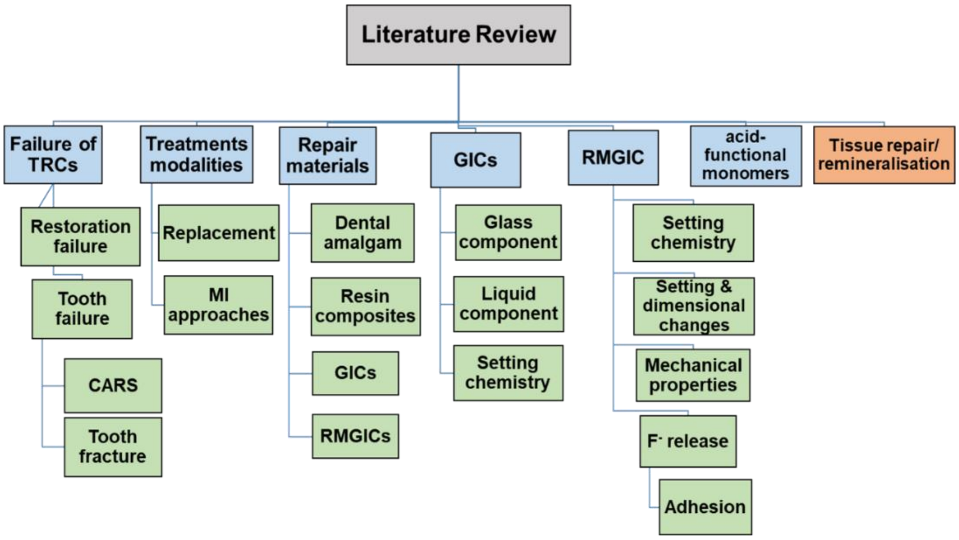
Ethylene glycol methacrylate phosphate (EGMP) is a photoreactive proton-conducting monomer with pendant phosphate groups. The complexation behaviour of the carbonyl and phosphoryl ligating groups in EGMP has been reported to enable remineralisation in hydrogels. Nuttelman et al. (2006) found that the incorporation of different concentrations of EGMP into the polyethylene glycol hydrogel (PEG) promote their attachment to human mesenchymal stem cells (hMSCs) via the formation of mineral phase that it is similar to the biological apatites in their structure and atomic composition, as shown in the compositional analysis and X-ray diffraction. EGMP creates a network of covalently linked phosphate groups that acts as nucleators and source of inorganic phosphate ions for mineralisation inside the hydrogel which sequestered the charged proteins (osteopontin) in the presence of $CaCl_2$, and thus promoted the hMSCs adhesion and spreading. It is believed that the positively charged, divalent Ca^{2+} ions in the fluid medium act as “bridges” between the negatively charged tethered EGMP molecules and the osteopontin molecules which counteract the repulsive electrostatic interactions between them. Both mechanisms may explain the ability of EGMP-containing PEGDA hydrogels to isolate osteopontin which in turn increase the cell viability of the gel-encapsulated hMSCs from 15% in the absence of EGMP to 97% in the presence of 50 mM of EGMP. Although the EGMP is cytotoxic at concentrations greater than 10 mM to hMSCs cultured in monolayer in vitro conditions. The viability of the encapsulated hMSCs in EGMP containing gels (10 and 50 mM) was shown to be much higher than that without EGMP. This study indicates that EGMP improves the survivability of the hMSC more than the un-modified hydrogels and the higher concentration of EGMP could successfully be utilised during the photoencapsulation of the hMSCs (Nuttelman et al., 2006).

Kemal et al., 2011, copolymerised different feed ratios of EGMP with HEMA to enhance the swelling dynamics of the hydrogel and facilitate cell adhesion and mineralisation. The study reported a direct correlation between the amount of EGMP and the degree of hydration of the HEMA-*co*-EGMP gel due to the ability of the pendant phosphate groups to ionise in low or high pH solution. The evaluation of the thermal behaviour showed that Tg increased with increasing the

EGMP. However, the presence of water influenced the transitions within the EGMP polymer networks, it did not have a deteriorating effect on the stiffness within the target temperature range even when fully hydrated. This improves the ability of pHEMA based hydrogel sponges to facilitate drug release, cell and tissue adhesion with an ability to bond to bone, thus expanding their biomedical application.

Münchow et al., 2015 developed a new self-primer via incorporating different concentrations of acidic-functional monomer (poly-propylene glycol phosphate methacrylate- Poly-P) to enhance the bonding performance of the self-etch adhesive to enamel. The synthesised monomer consists of long carbon chain molecule with low content of the polar groups to reduce the hydrophilicity of the primer with better chemical compatibility with enamel. The higher content of Poly-P (30-50% by weight) increased the acidity of the primer significantly with higher potential to dissociate into its ionic form and generate protons. This acidic self-primer modified the etching pattern with higher prisms rod exposure under SEM, which enhances the micromechanical interlocking of the self-etch adhesive to enamel with higher microtensile bond strength values.

The literature provides comprehensive understanding of the adhesion promoting effect of the phosphate acidic-functional monomer in self-etch adhesives and self-adhesive resin cement to different teeth/restoration interfaces. Ethylene glycol methacrylate phosphate (EGMP) is a proton-conducting electrolyte and the complexation behaviour of the carbonyl and phosphoryl ligating groups has been reported to enable remineralisation in hydrogels (Kemal et al., 2011) due to the charge in the gel and also improves the bonding efficacy and durability of self-etching adhesives (Inoue et al., 2005). Combining the properties of EGMP monomer as an adhesion promoter within RMGICs is a unique and interesting concept especially as the pendant phosphate groups are expected to interact with metallic cations during setting of the cement to form complexes that might alter the physical/biological properties of the cement itself and influence its adhesion to other restorative and dental substrates. Understanding the nature and dynamics of such modification would be valuable for the clinical applications and as a reparative material for failed tooth/restoration complexes.



1.7 Tissue repair and remineralisation

GICs' interaction with the tooth surfaces via the ion exchange mechanism can also occur in the presence of demineralised carious tissues (Ngo et al. 2006). The ions released from both interfaces combine to buffer the low pH which could replenish the demineralised tissues' ions and tipping the balance in favor of apatite re-formation. These processes involve the diffusion of calcium/strontium ions into the hypomineralised matrix, accompanied by the polyalkenoic acids that induce further demineralisation (Sennou et al. 1999), which eventually create an ion-rich layer followed by mineral deposition on pre-existent nuclei. However, in the absence of nucleation sites, no mineral deposition will occur (Kim et al., 2010b). Several laboratory and clinical studies supported the ability of GICs to remineralise the caries affected dentine (Ten Cate et al., 1995; Ngo et al., 2002; Ngo et al., 2006, Lee et al., 2008). Ngo et al. (2006) confirmed the diffusion of a substantial amount of strontium and fluorine into the partially demineralised dentine adjacent to GIC and their contribution in remineralising carious tissues. However, no significant changes were observed in the mineral levels of the demineralised dentine before and after GIC placement (Ngo et al., 2011). This confirms the assumption that the remineralisation of demineralised dentine may occur physiologically over time when a proper seal is evident (Fusayama, 1997). Kim et al. (2010b) and Atmeh et al. (2012) reported the failure of GIC to induce remineralisation in totally demineralised dentine with no mineralisation features observed in the intra- and inter-tubular dentine or detection of phosphate minerals. Such variations among studies could be due the differences in the GIC's effect on the partially demineralised CAD in comparison with totally demineralised dentine. This difference could be attributed to the fact that GIC might not be able to induce homogenous remineralisation without the presence of pre-existent nucleation sites which could exist in partially demineralised CAD only. However, the remineralisation is more complex in enamel, since there are no seed mineral crystals remaining like in partially demineralised dentine (Niu et al., 2014).

GICs are believed to prevent caries progression activity by their ability to remineralise the underlying dentine and or interfere with the remaining cariogenic bacterial growth and metabolism by the release of various ions and providing an initial low pH. The initial high influx of fluoride decreases the viability of bacteria within the cavity. In addition, fluoride is incorporated in the formation of fluorapatite crystals which are more resistant to acid dissolution and, along with calcium and strontium ions, provide the GIC with the capability to remineralise carious tissues (Ngo et al., 2006; Ramasetty et al., 2014). In addition, the presence of silica ions favours the mineralisation effect and enhances the apatite formation (Saito et al., 2003). Many in vitro studies suggested that acid-neutralising property and fluoride release of GIC/ RMGIC restorations are efficient to reduce the carious lesions progression in adjacent enamel surfaces in primary teeth when compared to amalgam restorations over 8 years (Qvist et al., 2004), and increase the resistance of enamel/dentine to demineralization (Hatibovic-Kofman et al., 1997; Jang et al 2001; Itota et al., 2010). The caries-preventive effect of GIC/RMGICs is higher than compomers and resin composites that did show any antibacterial effects (Vermeersch et al., 2005; Yengopal and Mickenautsch, 2011). Furthermore, the direct relationship between the acidity and growth inhibition of *S. mutans* in conjunction with fluoride release affect the *S. mutans* metabolism via inhibition of numerous enzymes and the fermentative activities and subsequently their viability (Seppä et al.; 1995; Vermeersch et al., 2005). However, Kuhn et al. (2016) observed that sealing the cavity using inert materials which isolate bacteria from oral environment and active biofilm is sufficient to arrest carious progression and allow tissue repair via the defence mechanism from the pulp-dentine complex.

Studies conducted in this thesis have used a glass ionomer cement (Fuji IX GP) and RMGIC (Fuji II LC) as a control material to which the new formulated cement (pRMGIC) was compared. The fact that all cements selected for this study, are water-based restorative materials with similar clinical application and potential for remineralising carious dentine, justify this selection. However, the effect of the added functional phosphate group to the commercial RMGIC and its interaction with adjacent enamel/dentine (sound versus diseased) surfaces has not been explored before and calls for more in-depth investigations to enhance understanding the potential of this phosphorylated cement.

Chapter two

2 Characterisation of eight commercial encapsulated and hand-mixed GIC/ RMGIC systems.

2.1 Introduction

Glass ionomer cements (GICs) are used widely in restorative dentistry due to their reliable, long-term chemical adhesion to tooth tissue, their low coefficient of thermal expansion, good tissue biocompatibility and fluoride release potential reducing the incidence of caries associated with restorations and sealants (CARS -formerly known as secondary caries) (Moshaverinia, et al., 2011; Lohbauer, 2010). Since its introduction in the early 1970s (Wilson, 1978) GICs have been supplied as separate powder/liquid formulations with the relative proportions being determined by the technical experience of the operator. The problems identified with hand-mixing GICs in clinical practice were identified to stem mainly from powder/liquid variations. These are related to differences in the powder packing densities achieved on filling the scoop and the manner in which the bottle is held, and the drop of liquid is dispensed. The more recent encapsulated form of GICs enables the powder/liquid ratio and mixing regime to be standardised by the manufacturer so that the functional properties of the mixed GIC cements are not influenced by operator-induced variability (Fleming et al., 1999, Fleming et al., 2002; Dowling and Fleming, 2009). Resin-modified glass ionomer cements (RMGICs) exhibit similar mechanical properties to conventional GICs, but the flexural strength of certain formulations was reported to have improved (Pameijer et al., 2015). However, their compressive strengths were found to be generally inferior to conventional GICs (Xu and Burgess, 2003) which may be attributed due the presence of the polymer chains interrupting the cement structure. Glass carbomer cement (GC) is a form of high-viscosity GIC containing nano-sized glass particles, hydroxyapatite and fluorapatite as fillers within its structure that are expected to transform into an apatite-like material over time (Van Duinen et al., 2004). Its clinical applications are similar to conventional GICs except that heat application is recommended to assist its setting reaction. This can be achieved by using a light curing device in the range of 1400 mW/cm^2. This cement is claimed to promote remineralisation of tooth tissue due to the presence of apatite, which provides nucleation sites for the remineralisation process after ageing (Zainuddin et al., 2012). The compressive testing regime allows sensitive changes in the mechanical properties of brittle materials to be distinguished through changes in their composition, P/L ratios and mixing methods (Nomoto and McCabe, 2001). Compressive modulus can be measured alongside the compressive fracture strength to predict the effect of the internal structural properties on the behaviour of the GICs when subjected to load (Saskalauskaite et al., 2008). Biaxial flexural strength (BFS) provides information about the material's mechanical integrity until fracture. It relates to the even distribution of the load within the specimen, in a way that it bends to its maximum capacity without crack formation (Darvell, 2012). It has been previously reported that modifications in both powder and/or liquid components of various commercial GICs lead to major changes in the cements' physical properties. However, it is not clear at the present time if different mixing regimes and/or ageing for the current improved GICs formulations have a direct effect on their physical properties. This chapter was conducted to assess the influence of two mixing regimes of six commercially available GIC/RMGICs on their physico-mechanical properties. Furthermore, two newer GICs containing ultrafine glass or apatite in their formulations dispensed only in encapsulated forms were tested under identical conditions to compare their properties. This experiment helped in understanding the physical properties of different commercial GIC systems and comparing the properties through changes in their composition and mixing mode. Secondly a number of GIC formulations with additives such as ultrafine glass or apatite are being advocated for clinical use hence a change in the GICs' composition is expected to influence the properties, thus two GICs dispensed in encapsulated form, were included as a part of the study to ascertain the effects on the physical properties. The null hypotheses proposed was that mixing regimes (mechanical vs. hand-mixing), the inclusion of reactive glass additives in GICs' composition, and short-term ageing do not affect their physical properties.

2.2 Materials and Methods

The cements used in this study are listed in Table 2-1. The components of each material were mixed under controlled room temperature (23±2°C) and humidity (35±5%), according to the manufacturers' instructions.

Table 2-1 Capsulated (C) and hand-mixed (H) glass ionomer cements (GICs) tested including the manufacturers' details, composition and powder/liquid ratios.

Materials	**Manufacturers**	**Code**	**Composition**	**P/L ratio**
Fuji IX GP Extra (GIC)	GC Corp., Japan	F9E (C),	CAFS-glass, PAA	0.4/0.12
Fuji IX GP Extra (GIC)	GC Corp., Japan	F9E (H)	CAFS-glass, PAA	0.34/0.1
Ketac^{TM} Fill Plus Aplicap^{TM} (GIC)	3M, ESPE, Germany	KFPA (C)	CAFS-glass, PAMA 35-55%, TA 5-10%	0.36/0.1
Ketac^{TM} Fill Plus (GIC)	3M, ESPE, Germany	KFP (H)	CAFS-glass, PAMA 35-55%, TA 5-10%	0.32/0.1
Fuji II LC (RMGIC)	GC Corp., Japan	F2LC (C)	CAFS-glass, PCA 5-10%, HEMA 25-50%, UDMA 1-5%, initiators, pigments	0.33/0.1
Fuji II LC (RMGIC)	GC Corp., Japan	F2LC (H)	CAFS-glass, PCA 5-10%, HEMA 25-50%, UDMA 1-5%, initiators, pigments	0.32/0.1
Glass Carbomer Cement (GIC)	GCP, The Netherlands	GC (C)	CAFS- glass 90% Apatite<6%, Polyacids<4%	------
Equia® Forte Fil (GIC)	GC Corp., Japan	EF (C)	CAFS-glass, ultra-fine reactive glass, PAA	0.4/0.13

CAFS-glass: Calcium aluminofluorosilicate glass; PAA: Poly acrylic acid; PAMA: Copolymer of acrylic and maleic acid, TA: Tartaric acid; PCA: Polybasic carboxylic acid; HEMA: 2- Hydroxyethyl methacrylate; UDMA: Urethane dimethacrylate**.**

2.2.1 Specimen preparation

Hand-mixed GIC and RMGICs (Fuji IX Extra (F9E), Ketac Fil Plus (KFP), and Fuji II LC (F2LC) were mixed according to the manufacturers' recommended P/L mixing ratio (Table 2-1) at ambient temperature (23±2°C) and humidity (35±5%).

After the bottle was tapped and shaken to unsettle the powder, it was dispensed using a levelled scoop, placed on a glass slab and separated into two equal parts. The liquid's bottle was tipped onto its side, inverted and squeezed gently allowing the dispensing of a clear drop without air bubbles. Half of the powder was mixed with the liquid for 10 s. The remaining powder was further mixed for 25 s in accordance with manufacturers' instructions. The encapsulated equivalents of the hand-mixed GICs (Fuji IX Extra (F9E), Ketac Fil Plus Aplicap (KFPA), and Fuji II LC (F2LC) capsules), Equia Forte (EF), and Glass Carbomer Cement (GC) were also included (Table 2-1). All capsules were tapped, activated for 10 s and placed in the appropriate mixing machine (Ultramat 2, mixing frequency 4600 oscillations/minute, SDI, Germany) or a Rotomix™ machine (mixing speed: 2850 rpm, centrifugation speed: 2950 rpm, 3M ESPE, USA) and mixed for 10 and 8 s respectively in accordance with the manufacturers' instructions. Cylindrical specimens (6.0 ± 0.1 mm height and 4.0 ± 0.1 mm diameter) were prepared for the CS and modulus, and the MH tests, while a stainless-steel mould (8.3 ± 0.1 mm diameter and 1.3 ± 0.1 mm thickness) was used to prepare the disc specimens for the BFS test. The hand-mixed material was allowed to flow into the mould using a stainless-steel spatula to minimise air entrapment. For the encapsulated GIC, the nozzle of the capsule was positioned to one side of the unfilled mould and the plastic mass extruded slowly to encourage laminar flow (Dowling and Fleming, 2008). The mould was slightly over-filled with each material and sandwiched between two glass plates under constant pressure with standard load 500 mg over the mould to extrude any excess and provide parallel flat specimen ends (Lloyd and Mitchell, 1984).

F2LC (capsule and hand-mixed) specimens were photo-polymerised after 3 min and 45 s (following the manufactures' recommendations), using a light curing device (Model 503, Dentsply, Germany) with light intensity of 450 mW/cm^2 for 30 s at each end of the cylindrical mould (McKenzie et al., 2003). The curing light intensity was monitored with a light meter (Curing Radiometer Model 100, Demetron /Kerr, and Danbury, CT, USA). A CarboLED CL-01 (GCP Dental, Vianen, The Netherlands) with a light intensity of 1400 mW/cm^2 was used for the Glass Carbomer specimens. The output temperature was measured every ten second during a one-minute period, using Heat/Glare Radiometer (Model 200, Demetron /Kerr, and Danbury, CT, USA), placed at the tip of the curing light.

Twenty seconds light curing was applied to the top surface of disc specimens. All samples were left to set at 37°C for 15 min (ISO 9917-1 standard 2007 for water-based cements, and ISO, 9917-2 standard 2010 for the resin-modified cements, Geneva: ISO). Then, specimens were removed from the moulds and stored in artificial saliva that was prepared according to the formula provided by Eisenburger et al., (2001) at 37°C until the time of mechanical testing (1, and 30 days) and changed once a week up to four weeks (Okada et al., 2001).

2.2.2 Mechanical properties

A sample size of n=8 was used to determine compressive strength (CS) and compressive modulus (CM), microhardness (MH), and biaxial flexural strengths (BFS), based on the specifications outlined in ISO 9917-1 standard 2007 for water-based cements, and ISO, 9917-2 standard 2010 for the resin-modified cements, Geneva: ISO.

2.2.2.1 *Compressive strength and modulus*

Sixteen cylindrical specimens of each commercial material were prepared for the compressive strength test (CS). After storage (1 and 30 days), the diameter of the cylindrical specimens were measured at three points using a digital micrometre screw gauge with an accuracy of 10 µm (Moore and Wright, Sheffield, England), and the mean diameter was calculated prior to testing. To mimic the oral environment, the specimens were tested 'wet' by placing a wet filter paper (Whatman No. 1, Whatman International Ltd., Maidstone, England) between the two ends of the specimen and test machine platens. A universal testing machine (Instron model 5569, USA) with a 500 N load cell was used for testing the compressive strength and modulus at a crosshead speed of 0.5 mm/min. The compressive load was applied to the long axis of each specimen and the maximum load to failure was recorded (Fleming et al., 2003). The compressive strength, *P* (MPa), of each individual cylindrical specimen was calculated by dividing the fracture force (*F*) by the area of the specimen where *D* was the specimen diameter using Eq. (1), (Dowling and Fleming, 2009)

$$P = \frac{4F}{\pi D^2} \qquad \text{Eq (1)}$$

Where F was the load at fracture (N) and D was the mean specimen diameter (mm).

In order to assess the mechanical deformation behaviour, the compressive stress-strain curves were obtained for the investigated materials. The compressional stress (σ) was determined by dividing the applied force F (N), by the cross-sectional area A (mm^2), Eq (2):

$$\sigma = \frac{F}{A} \qquad \text{Eq (2)}$$

The strain was measured by dividing the change in length by the original length, Eq (3):

$$\boldsymbol{\varepsilon} = \frac{\Delta l}{l} \qquad \text{Eq (3)}$$

Δl is the extension (change in length), while l is the original length.

Individual stress/strain plots were measured for each cylindrical specimen tested in compression. Then the compressive modulus (the ratio of stress to strain below the fracture limit) was determined by calculating the slope of the initial linear segment of the stress-strain curve (Xie et al., 2000; Dowling and Fleming, 2009; Samuel et al., 2009).

2.2.2.2 *Surface Microhardness*

At the end of the specified storage times (1 and 30 days), the surface hardness of sixteen cylinders of each glass-ionomer cement was determined using Knoop hardness test (Duramin10, Struers, Japan) at ambient temperature (23±2°C) and humidity (35±5%). Each specimen was placed on the instrument platform with the tested surface facing the indenter that was brought into contact with this surface. A minor load of 1 gf (0.01 N) was applied to the surface for 5 s to ensure the contact between them (Kanchanavasita et al, 1998), while the depth indicator at the dial gauge was adjusted to zero. The indentations were performed using 50 gf load force for 10 s, since the application of higher loads or a longer contact time can initiate cracks at the surface of these cements (De Moor and Verbeeck, 1998). Under these conditions, the sharply defined indentation marks were obtained with a size allowing the determination of the surface hardness with a sufficient accuracy. The Knoop Hardness Numbers (KHN) were recorded as an average of 6 readings at randomly selected areas which are at least 1 mm far away from the adjacent indentations or the margin of the specimens.

2.2.2.3 *Biaxial Flexural strength*

Twenty disc-shaped specimens (8.3 mm diameter, 1.3 mm thickness) of each group were prepared and tested for BFS test after 1 and 30 days. The dimensions were determined to the nearest 0.001 mm using a digital micrometre (Micro 2000, Moore and Wright Ltd., Sheffield, England). The measurements were made at three different sites on the specimens. The tests were conducted at the ambient temperature (23±2°C) and humidity (35±5%). The specimen was placed centrally on a 6.5 mm diameter circular support in such a manner that the edge extended beyond the support by the same amount around the whole specimens. Then, this specimen was centrally loaded with a 1.5 mm diameter round ended indenter in a way that the area of maximum tensile stress was located at the centre of the lower face of the disc, as shown in Figure 2-1. The load was applied using a universal testing machine (Instron Model 5569, USA) at a crosshead speed of 0.5 mm/min until the specimens yielded or fractured.

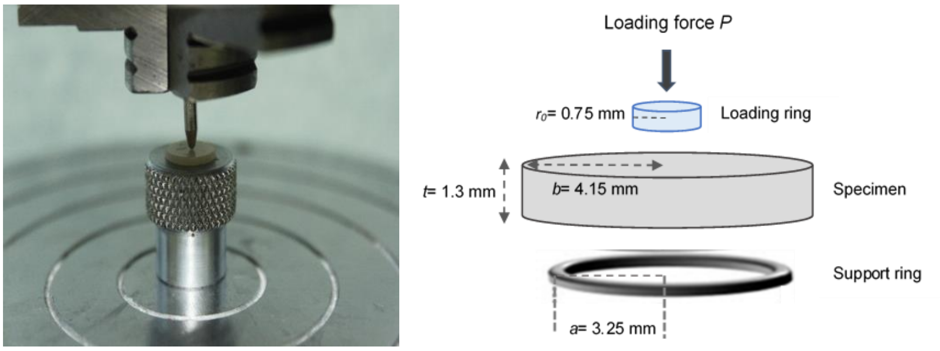


Figure 2-1 Biaxial flexural strength testing apparatus

The load at failure was obtained directly from the loading curves. Each fractured specimen was inspected for significant voids and irregularities. When the fractures occurred at some obvious voids or flaws, the specimen was excluded from further analysis (Kanchanavasita et al., 1986).

The BFS value was calculated using the following equations, Eq (4-6), (Shetty et al., 1983):

$$\sigma = \frac{AP}{t^2} \tag*{Eq (4)}$$

$$A = 3/(4\pi)[2(1 + v)\ \ln(a/r_0^*) + (1 - v)\{\frac{2a^2 - r_0^{*2}}{2b^2}\} + 1 + v] \tag*{Eq (5)}$$

Where *P* is the applied load at failure, v is Poisson's ratio (0.35 for GIC restoratives), *a* is the radius of support circle, *b* is the radius of disc specimen, t is the thickness of the disc specimen, and r_0 is the radius of the ball used on the loading surface, as shown in figure 2-1:

$$r_0^*\sqrt{(1.6r_0^2 + t^2) - 0.675t} \tag*{Eq (6)}$$

Where r_0 is an equivalent radius of contact between the loading ball and the disc specimen, where loading can be considered to be uniform.

The specific value of Poisson's ratio for RMGICs is not reported in the literature. Since the physico-mechanical characteristics of the RMGICs lie between the composite resins and conventional GICs (Gladys et al., 1997), it is reasonable to assume that the Poisson's ratio of these materials must lie somewhere between 0.24 for composite resins (Nakayama et al., 1974) and 0.35 for conventional GICs (Akinmade and Nicholson, 1995). The average value is equal to 0.295 which is speculated for the RMGICs. However, a preliminary investigation showed that the variation in the Poisson's ratio of 0.05 (e.g. from 0.30 to 0.35) resulted in differences in the calculated BFS and modulus of 5%. Accordingly, and following McKenzie et al. (2003), 0.35 was used in the calculation of BFS of both GIC and RMGIC in this study.

2.2.3 Fluid uptake

Fluid uptake was measured as percentage hydration using ten discs (8.3 mm diameter, 1.3 mm thickness) of each group immersed in artificial saliva at 37°C. Initial weight measurement for each sample W_0 was carried out using an electronic balance analyser (Mettler Toledo XS105DU, Switzerland) to an accuracy ±0.2 mg. After storage in an incubator at 37±1°C, the specimens' surfaces were gently dabbed on a filter paper and weighed daily until 30 days. The percentage fluid uptake was determined using the following equation, Eq (7) (Rojo et al., 2008):

$$\% \, weight \, change \; = \frac{Wt - W0}{W0} \times 100 \qquad \text{Eq (7)}$$

Wt is the weight at time t, and W_0 is the initial weight of the specimen.

2.2.4 Fluoride ion release

Fluoride ion release measurements were recorded up to 30 days (n=10 per group) using disc-shaped specimens (8.3 mm diameter, 1.3 mm thickness) for each group. Each specimen was immersed in an individually capped polystyrene tube containing 2 ml of artificial saliva (pH 7.0) and stored at 37°C. To avoid fluoride saturation of the solution, the storage medium was refreshed every 48 h up to 4 week (Geurtsen et al., 1999). An equal volume (2 ml) of total ionic strength adjustment buffer (TISAB I BDH Ltd., Poole, England) was added prior to fluoride ion measurements, which increases the ionic strength of the solution to a relatively high level and hence increases the accuracy of the reading. Fluoride concentrations were recorded in ppm using a selective fluoride electrode (Cole Parmer 27502) connected to an ion analyser (OAKTON 510 ion series, Singapore). The amount of fluoride eluted from the GICs were converted into milligrams of F^- released per unit surface of area (mg F/cm^2) (Fukazawa et al., 1987).

2.2.5 Scanning electron microscopy (SEM) and energy dispersion X-ray spectrometry (EDX) analysis of glass carbomer cement (GC)

Representative surfaces from the GC cement specimens were dried, carbon-coated, and examined using a scanning electron microscope (SEM, FEI Co. Ltd., Cambridge, UK) with an accelerating voltage: 10 kV, working distance: 20 µm, 10 µm, and 2 µm, and magnification: x2500, x10000 and x25000, respectively, coupled to an energy dispersive X-ray spectroscope (EDX) (EDAX Inc., 91 McKee Drive, Mahwah, NJ 07430 USA). Elemental analysis of the GC cement at 24 h and 30 days were carried out to observe the structural changes within this material after ageing.

2.2.6 Statistical analysis

One-way analysis of variance (ANOVA) with Games-Howell post-hoc tests (alpha level=0.05) were used to assess differences in the physical properties among groups. The mean values were further compared by using Games-Howell post-hoc tests for multiple comparisons (IBM®, SPSS® statistics20, Chicago). Independent t-tests ($p<0.05$) were used to compare the effect of time (1 and 30 days) on the mechanical properties of each tested material.

2.3 Results

2.3.1 Mechanical properties

Auto-mixed GICs/RMGIC (F9E, KFPA and F2LC) exhibited statistically significantly ($p<0.05$) higher immediate and delayed CS values compared to their hand-mixed equivalents (Table 2-2). EF(C), F9E(C) and KFPA(C) showed the highest values after both intervals. F2LC (C) showed comparable values to F9E(C) and EF(C) post-ageing. In contrast, the hand-mixed version of F2LC recorded the lowest CS ($p<0.05$) that was comparable to GC at the early term, and both GC and F9E (H) post storage. Short-term ageing reduced the CS of F9E(C&H) and EF(C), but enhanced it in KFPA(C&H), RMGICs (C&H) and GC (t-test, $p<0.05$).

The same trend was seen in the early CM values, Table 2-2. The encapsulated GICs (F9E, KFPA and F2LC) recorded higher values in comparison to the correspondent hand-mixed versions. However, over time, such difference was only significant in RMGICs ($p<0.001$). Auto-mixed F9E and KFPA showed the highest initial compressive modulus among all groups. After short-term ageing, these values are comparable to KFP (H) and EF (C). It is not surprising that RMGICs (C&H) showed lower CM values than the conventional GICs, however, the values of the auto-mixed version are comparable to F9E (H) and GC at the early term, and to GC at the delayed term. The CM of all materials was enhanced post-ageing, however, it was only statistically significant in F9E (H), KFP (H), F2LC (C), GC and EF (t-test, $p<0.05$).

Table 2-2 Compressive strength (CS) and compressive modulus (CM) for the GIC-RMGICs after 1 and 30 days, shown as mean (SD), n=8.

Groups	**CS [MPa] 1 day**	**CS[MPa] 30 days**	**CM [GPa] 1 day**	**CM[GPa] 30 days**
F9E (C)	205.2 (14.6)^{*a}	181.9 (13.3)$^{*\wedge d}$	8.6 (0.3)^{*gi}	8.8 (0.9)^{ln}
F9E (H)	153.8 (11.2)^{b}	141.8 (7.4)$^{e\wedge}$	5.0 (0.9)^{h}	7.9 (0.6)$^{\wedge mn}$
KFPA (C)	193.1 (10.8)^{*a}	210.1 (14.9)$^{*\wedge f}$	9.7 (0.5)^{*g}	10.1 (0.5)^{l}
KFP (H)	165.1 (13.7)^{b}	171.1 (12.3)^{d}	7.7 (0.7)^{ij}	9.4 (0.5)$^{\wedge ln}$
F2LC (C)	169.8 (8.9)^{*b}	181.9 (8.7)$^{*\wedge d}$	4.4 (0.2)^{*hk}	5.8 (1.4)$^{\wedge * o}$
F2LC (H)	108.1 (12.6)^{c}	125.8 (17.6)$^{\wedge e}$	2.4 (0.3)	2.7 (0.6)
GC	110.0 (6.4)^{c}	134.6 (6.4)$^{\wedge e}$	3.6 (0.7)^{k}	7.0 (0.6)$^{\wedge mo}$
EF	216.4 (18.1)^{a}	186.6 (11.7)$^{\wedge df}$	7.1 (0.8)^{j}	9.0 (0.7)$^{\wedge l}$

(*) significant difference between capsulated and hand-mixed GICs. (^) significant difference within each group after short-term ageing in artificial saliva (t-test, *p*<0.05). Similar letters in columns indicate no significant differences among GICs (Games-Howell test post-hoc tests, an alpha level of 0.05).

The surface hardness was higher in the encapsulated GICs/RMGIC than the corresponding hand-mixed at both time periods, Table (2-3). F9E (C) and EF (C) recorded the highest early KHN among all groups which was comparable to KFPA (C) post-ageing. RMGIC (C&H) showed lower hardness properties in comparison to the CGCs, however, the encapsulated group showed comparable early values to KFPA (C&H), and F9E (H) and GC post-ageing. All conventional GICs displayed an enhancement in MH values after storage, however, it was only significant in F9E (H), KFPA (C&H) and GC. In contrast, RMGICs (C&H) exhibited a reduction in KHN over time, however, this decrease was not significant for the hand-mixed version.

The BFS values of the mechanically-mixed GICs/RMGIC are also higher than the manually-mixed version in both time intervals. However, the differences are not statistically significant in F9E after 24 h and in KFP after 30 days. The flexural strength of the RMGIC (C&H) are significantly higher than the conventional GICs at both time intervals (*p*<0.001). The encapsulated KFPA reported the highest early BFS value among CGICs, but after storage, both versions (C&H) of KFPA showed this trend. Short-term ageing showed a variable effect on the BFS of the tested GICs. Some materials revealed a significant enhancement (KFPA (H), F2LC (C) and GC), some remained unchanged (F9E (C) and EF), while others were compromised post-ageing (F9E (H), KFPA (C), F2LC (H), (Table 2-3).

Table 2-3 Microhardness (MH) and biaxial flexural strength (BFS) for the GIC-RMGICs after 1 and 30 days (mean and (SD), n= 8).

Groups	**MH [KHN] 1 day**	**MH[KHN] 30 days**	**BFS [MPa] 1 day**	**BFS[MPa] 30 days**
F9E (C)	62.3 (4.4)^{*a}	63.9 (4.5)^{*f}	48.1 (6.2)^{hi}	44.4 (5.9)^{*jk}
F9E (H)	35.3 (2.5)^{b}	39.3 (4.8)$^{\wedge g}$	40.7 (4.2)^{h}	34.3 (4.9)$^{\wedge l}$
KFPA (C)	52.1 (2.9)^{*c}	63.3 (4.1)$^{*\wedge f}$	70.0 (4.5)*	61.9 (2.1)$^{\wedge m}$
KFP (H)	44.4 (4.9)^{d}	54.9 (2.9)$^{\wedge}$	42.9 (6.4)^{hi}	60.7 (4.0)$^{\wedge m}$
F2LC (C)	49.4 (4.2)^{*cd}	37.0 (2.6)$^{*\wedge g}$	135.8 (8.2)*	174.4 (7.0)$^{*\wedge}$
F2LC (H)	32.5 (2.9)^{be}	28.0 (4.2)	122.8 (7.8)	91.5 (9.5)$^{\wedge}$
GC (C)	28.0 (2.6)^{e}	40.6 (1.7$^{)\wedge g}$	27.2 (3.5)	38.0 (3.9)$^{\wedge jl}$
EF (C)	60.0 (3.0)^{a}	61.1 (3.0)^{f}	50.9 (3.7)^{i}	50.7 (5.4)^{k}

(*) significant difference between capsulated and hand-mixed GICs. (^) significant difference within each group after short-term ageing in artificial saliva (t-test, $p<0.05$). Similar letters in columns indicate no significant differences among GICs (Games-Howell test post-hoc tests, an alpha level of 0.05).

2.3.2 Fluid uptake

All investigated materials did not reach the equilibrium after four weeks' immersion in artificial saliva (Figure 2-2). Accordingly, the solubility and coefficient of diffusion for the fluid uptake process are not calculated. Hand mixed GICs-RMGIC displayed higher fluid uptake than the encapsulated equivalents up to 30 days. Fluid uptake was higher in GC followed by the hand-mixed F2LC and F9E, whereas F9E (C) recorded the least uptake over time. The encapsulated RMGIC showed higher level of weight change percentage than all capsulated CGICs except GC.

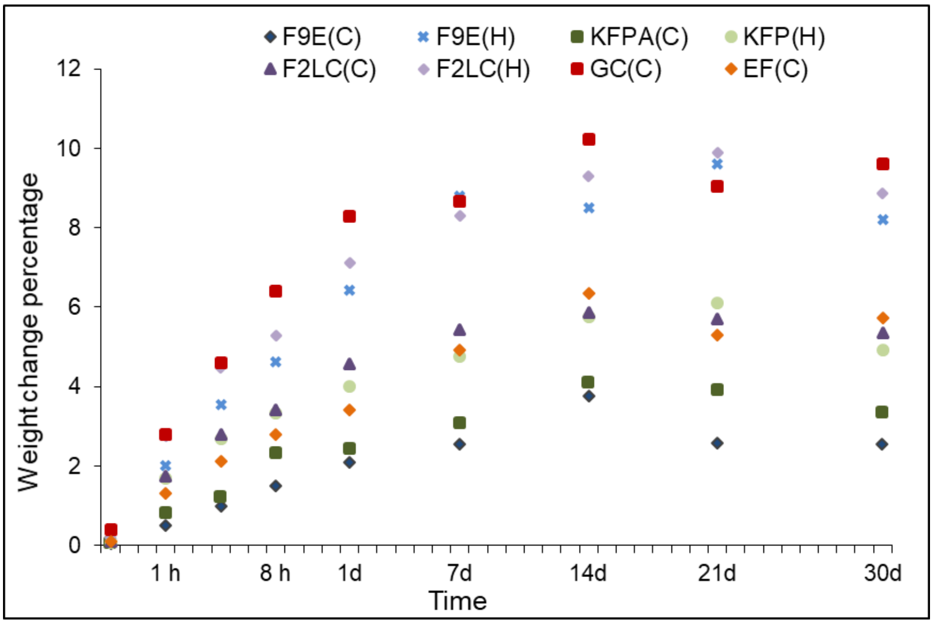


Figure 2-2 Fluid uptake of the GIC-RMGICs over 30 days

2.3.3 Fluoride release

All GICs/RMGIC exhibited a similar pattern of fluoride release (Figure 2-3), which is characterised by a strong initial release in the first 48 h, after that decreasing with time until reaching an asymptotic tendency to equilibrium after the second week. In CGIC, the early fluoride release was higher in the hand-mixed version (F9E and KFP) than the encapsulated equivalents, whilst a contrary trend was observed for F2LC. The early fluoride release was lower in RMGICs (C&H) in comparison to the conventional GICs. However, after 30 days, the amount of fluoride release was comparable in all investigated materials.

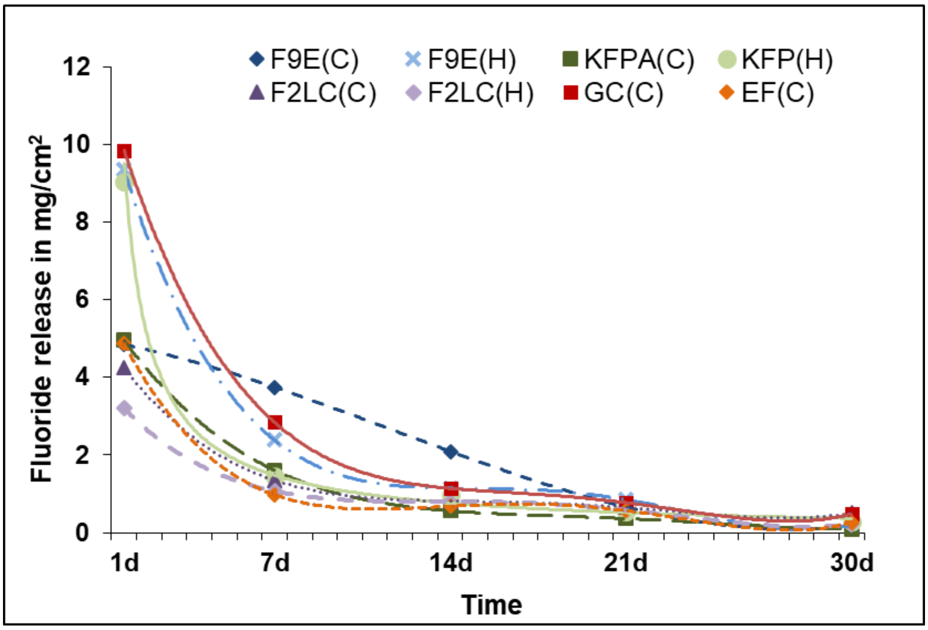


Figure 2-3 Fluoride release profile in mg/cm^2 from the tested GIC-RMGICs

2.3.4 SEM-EDX analysis of glass carbomer cement (GC)

The glass carbomer cement showed the dispersion of particles with varying size and shape whilst the specimens aged in artificial saliva for 30 days showed evidence of mineral deposits on the surface that were distinctly different from the particles with the cement. Mineral depositions were observed clearly on the surface of the GC-30 samples with x10000 and x25000 magnification, as shown in Figure 2-4 (B-1 and 2). EDX analysis of GC-24 and GC-30 samples provided the distribution of F, Si, Al, in addition to P and Ca, within their matrices, Fig 2-5. Abundant quantities of Ca and P was observed within GC matrix post-ageing, Figure 2-5 (b).

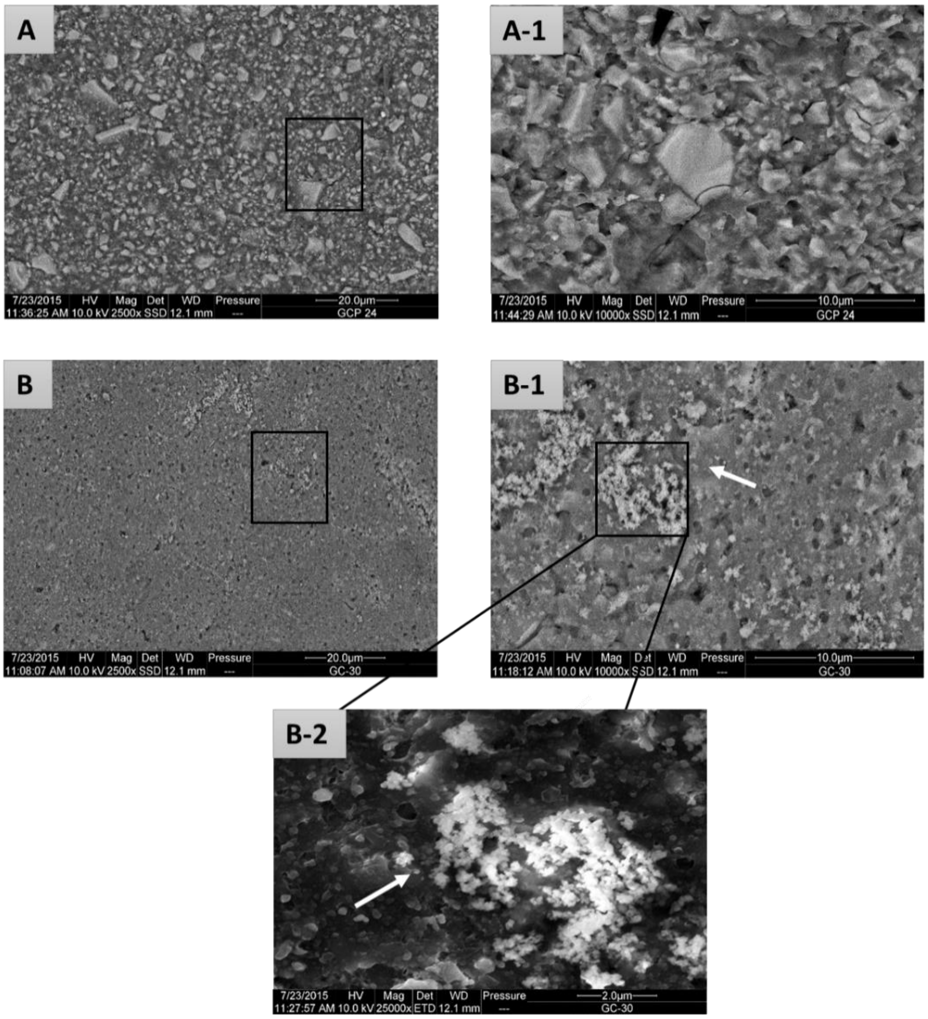


Figure 2-4 SEM micrographs of GC at different magnifications (x2500, and x10000). (A) GC-24 and (B) GC-30. White arrow in (B-1) showed the presence of mineral deposition on the surface of GC-30 which is more clearly observed at x25000 (B-2).

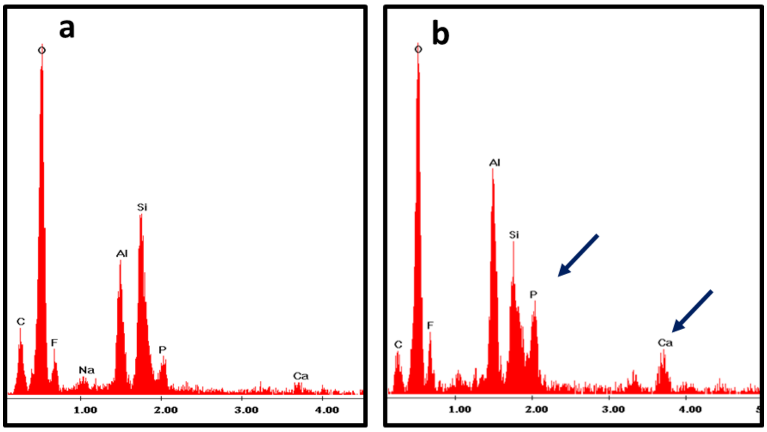


Figure 2-5 EDX analysis of GC-24 and GC-30 (a, b, respectively), blue arrows at (b) showed an increase in P and Ca ions peak in the structure of GC-30.

2.4 Discussion

2.4.1 Mechanical properties

2.4.1.1 *Effect of mixing (mechanical vs. hand-mixing)*

In line with previous findings (Dowling and Fleming, 2009; Molina et al., 2013), encapsulated GICs/RMGIC (F9E, KFPA and F2LC) revealed higher compressive strength (CS) and compressive modulus (CM), microhardness (MH) and biaxial flexural strength (BFS) than the hand-mixed equivalents after two-time intervals, Table (2-2,3). The encapsulated versions eliminated the inaccurate dispensation prior to mixing and the mixing regime was standardised by mechanical mixing in accordance with the manufacturers' instructions. Such mixing reduces porosity with more thorough wetting of the powder particles which enhance the setting reaction and thus the mechanical strength of the cement (Nomoto et al., 2004).

Furthermore, the selected encapsulated materials have higher P/L ratios compared to the corresponding hand-mixed forms. This increases the viscosity of the material and homogeneity of the mix, thus improving the mechanical properties as suggested by Nomoto and McCabe (2001). Earlier studies (Pearson and Atkinson, 1991; Williams and Billington, 1991) reported higher compressive strength of the hand-mixed GIC's, however, they utilised encapsulated GIC with lower powder content for a constant volume of liquid compared with the hand-mixed equivalent. Regarding clinical handling, the encapsulated systems are more convenient to use and reduce variations in the P/L ratio. Therefore, the null hypothesis stated in the current study that encapsulated GIC restoratives would not perform more favourably regarding compressive strength, modulus, biaxial flexural strength and microhardness to their hand-mixed equivalents was rejected.

2.4.1.2 *Effect of composition of the tested GICs*

The differences in the composition and P/L ratios of the tested GICs have a direct influence on their physical properties. The higher powder/liquid ratio in EF, F9E (C) and KFPA (C) lead to enhanced mechanical strength immediately and after storage. This fact is well supported by previous studies (Yap et al., 2001; Behr et al., 2006; Moshaverinia et al., 2008) since the unreacted powder particles may act as reinforcing fillers within the matrix which impede crack propagation within the cement.

The inclusion of fine smaller-sized reactive glass particles coupled with higher P/L ratio in EF encouraged higher cross-linking with a possibility to act as strengthening fillers that increase the resistance of the cement to compressive loading (Pereira et al., 2002; Sidhu, 2011; Zanata et al., 2011). This leads to improved mechanical properties including compressive strength and modulus, flexural strength, and surface hardness compared to the other tested GICs. However, the inclusion of fine small-sized reactive hydroxyapatite and fluoroapatite particles (<6%) within the glass powder in GC did not show a beneficial effect in term of mechanical strength. They might disrupt the cement forming process producing a cement with inferior mechanical properties (Yamamoto, 1984; Nicholson et al., 1993; Arita et al., 2011; Arslanoglu et al., 2015).

The addition of resin to the conventional GIC produces integrated network composite analogue composed of unreacted glass particles surrounded by a silica hydrogel, which are embedded in a cross-linked poly (alkenoic acid) -ion- resin copolymer. The cross-linked poly-HEMA and polyacrylate salts form a homogeneous matrix that reduces the flexural modulus coupled with higher flexural strength which increases the material resiliency. The elasticity produced by the polymerisable components which have low elastic modulus enables the RMGICs to undergo greater flexure without fracturing, hence increasing the immediate and prolonged BFS strength of the F2LC (C&H) and accounts for the lower CM and MH values. This finding is well supported by previous studies (Mitsuhashi et al., 2003; Cefaly et al., 2009; Pameijer et al., 2015).

In contrast, the extent of the acid-base glass ionomer reaction was significantly delayed when specimens are light-cured after mixing. The rapidly formed polymer network between 2-HEMA and the pendant methacrylate groups of both the ionised and unionised fractions of polyacrylic acid reduced the rate of the acid-base reaction, apparently due to steric hindrance phenomena. This phenomenon may hinder the complete formation of poly-salt bridges (Eliades and Palaghias, 1993; Peutzfeldt et al., 1997) with possibility of phase separation during setting reaction (Nicholson, 1994), which might compromise the compressive strength of RMGICs as compared to their conventional counterparts. Hand-mixed F2LC showed inferior CS, CM and MH values than all conventional GICs except GC, but the encapsulated form displayed comparable values to some CGICs. This observation is also supported by Gladys et al. (1997); Saskalauskaite et al. (2008); Arslanoglu et al. (2015).

2.4.1.3 *Effect of ageing*

The setting reaction of glass ionomer cements involve the reaction of the Ca^{2+} and Al^{3+} ions released from the aluminofluorosilicate glass with the water-soluble polymeric acid. During the maturation of the cement the Al^{3+} ions that initially exist in the four-coordination state to accommodate the tetrahedral silicate network of the glass, progress to the six-coordination state that enhances the mechanical properties to an extent. The evolution of strength in GICs with time shows distinct patterns of change since strengthening is attributed to the additional crosslinking and build-up of a silica gel phase, whereas weakening may result from erosion and plasticising effect of water (Cattani-Lorente et al., 1993). In agreement to previous studies (Uno et al., 1996; Gladys et al., 1997), the mechanical properties (CS, MH and BFS) of KFP (C&H), GC (C) and F2LC (C) showed a tendency to increase post-ageing. The same trend could be seen in the CM of all tested materials, whereas the rest materials were displayed a variable range of values among tested properties after storage.

The most noticeable enhancement in strength properties post-ageing was seen in the GC group as compared to its inferior early values. The mechanisms underlying this enhancement is thought to be partially due to cement maturation (Mesquita et al., 1999), as well as the presence of abundant Ca^{2+} ions arising from the dissolution of HAp within the GC matrix, which participate in hardening the cement. SEM observation confirmed these findings, as it showed dispersion of mineral deposits on the surface of aged cement which were recognised at the higher magnifications, Figures 2-4 (B-1, and 2). This was also associated with abundant quantities of Ca and P was observed by EDX within the cement matrix after ageing (Figure 2-5, b). These results are consistent with studies of Moshaverinia et al. (2008) and Zainuddin et al. (2012) which revealed a dramatic rise in the mechanical properties of the cement containing HAp and FAp post-ageing as it produces stable hard, brittle material with a highly cross-linked polyacid salt matrix.

2.4.2 Fluid uptake

All investigated materials immersed in artificial saliva did not reach the equilibrium after four weeks since the artificial saliva contains water and other components that could diffuse in and out of the sample. Droplets also might be formed within the specimen due to the presence of soluble components inside the matrix (Parker and Braden, 1989). After reaching a maximum weight change some of the components that diffused into the cement, and those that were not bound to any chemical groups, diffused out again thus accounting for the weight loss due to a potential chemical gradient (Riggs et al., 2001). Mechanical mixing reduces air spaces between adjacent particles which minifies the porosity and enhances the wetting of the powder particles and thus improves the bulk properties of the resultant cement which might interfere with fluid diffusion into the matrix. Following this concept, all encapsulated GICs/RMGIC in the present study presented less fluid uptake than their hand-mixed equivalents over time, Fig (2-2). In contrast, air voids that are generated by hand-mixing can accelerate the water uptake and solubility of these cements leading to less than optimal performance (Kanchanavasita et al., 1997; Cefaly et al., 2003).

GIC and RMGICs absorb water that is necessary for the acid-base setting reaction and ionic crosslinking. Water usually diffuses through the bulk of the cements via micro-voids or binding to the resinous groups which contain hydrophilic moieties such as HEMA (2-hydroxyethyl methacrylate) (Yap, 1996; Burrow et al., 1999). This might explain the higher fluid uptake observed in RMGICs in comparison to the conventional counterparts.

2.4.3 Fluoride release

Despite the diversity in the reported amount of fluoride release from GICs from previous studies (Verbeeck et al., 1993; Hattab and Amin, 2001; Thanjal et al., 2010), the pattern of release remains consistent. It is characterised by an initial short-term burst release, followed by a prolonged and more slowly occurring elution which would be responsible for the long-term fluoride release, Fig (2-3). In corroboration with previous studies (Verbeeck et al., 1993; Thanjal et al., 2010), mechanical mixing produces more tightly bonded polyalkenoate matrix resulting in slow diffusion of fluoride from the cement matrix as the initial elution depends on the ability of F^- ions to diffuse through cement voids, cracks and microporosities. Accordingly, auto-mixed GICs (F9E and KFPA) exhibited lower fluoride in comparison to their hand-mixed equivalents during the first 48 h. Theoretically, the set matrix of RMGIC is composed of the ionomer salt hydrogel and polymer in which fluoride ions might be firmly encapsulated by resin matrix that might reduce the rate of fluoride release (Wilson, 1990; Momoi and McCabe, 1993). Following this concept, early fluoride release in F2LC (C&H) was significantly less than conventional GICs. However, some studies suggested that poly-HEMA can absorb sufficient water to enable diffusion of the fluoride ions. Otherwise it will be firmly encapsulated within the polyacrylate matrix (Mitra, 1991; Forsten, 1995).

2.5 Limitations and future work

The mechanical testing regimes used in the current study are based upon the method outlined in ISO 9917-1 and 9917-2 for the water-based cements. These methods appear to be sensitive to changes in the physical properties of the GIC/RMGICs through changes in their composition and mixing mode. However, the actual values of loads at failure in the current study are not intended to be clinically relevant rather the relative probabilities of survival for the restorative cements that were compared. Furthermore, the results highlighted the effect of mixing method and properties among materials from the same generic groups and manufacturer. However, other factors such as the different composition of the powder and liquid and P/L ratios used for the encapsulated and manual mixed cements are not provided by the manufacturer for the same commercial brand (C vs. H), which is expected to affect the properties rather than the mixing mode alone. Since the type, size and the composition of the glasses affect their reactivity with the polyacid liquid, which may also have different polymeric constituents in the different cement versions. These factors can affect the rate of setting reaction and thus the final properties, added to the differences in P/L ratios which also may affect the initial viscosity of the mix and influence the early and long-term properties. However, this study was performed to mimic the clinical usage of GIC systems, and the two different cements provided by the manufacturer were compared, which identified that the hand-mixed GICs can lead to inferior properties. This primarily arises due to the variation in powder/liquid ratio which are related to the differences in the powder packing densities on filling the scoop and the manner in which the bottle is held, and the drop of liquid is dispensed, added to the possibility of presence of air bubbles inside the bottle which affect the quantity of the drops. Thus, ideally the powder in the scoop and liquid should be weighed to exclude these variations.

Despite artificial saliva is an appropriate medium to mimic the oral environment the results for calculating water uptake, solubility and coefficient of diffusion is not conclusive since equilibrium is not reached within a month. This might also indirectly affect the surface properties of the aged cements, in addition to the possible interference with the amount of fluoride release. Thus, these results are only an indication of the interaction of the cements in artificial saliva and since they were carried under same conditions can only be considered as a comparison.

2.6 Conclusions

Within the limitation of the present study, the mixing mode and short-term ageing have a significant effect on the physical properties of the current commercial GICs. Encapsulated GICs and RMGICs exhibited superior physical properties compared to their hand-mixed equivalents after 1 and 30 days. Encapsulated RMGIC showed satisfactory mechanical properties in comparison to the conventional GICs, while the hand-mixed RMGIC exhibited inferior CS, CM, and MH. The addition of nano-sized reactive glass particles in Equia® Forte Fil exhibited an enhanced immediate and short-term mechanical properties. The modification of RMGIC's with <6% hydroxyapatite/fluorapatite (HAp/FAp) nanoparticles and liquid silica in the glass carbomer cement showed inferior mechanical strength in comparison to the other commercial GIC & RMGIC tested in this study. However, ageing of the GC produced a dramatic rise in the CS, CM, MH and BFS values partly due to cement maturation and the precipitation of HAp within the cement matrix, as shown in the SEM. GC showed enhanced fluoride ion release with a potential remineralising capability.

Chapter Three

3 New RMGIC containing a phosphate-based monomer (EGMP) as a reparative material for repairing failed TRCs.

3.1 Introduction

Maintaining the integrity of the tooth-restoration complex (TRC) remains a challenge in clinical dentistry. Replacement of existing restorations promotes the acceleration of the "restoration death spiral", while minimally invasive management increases TRC longevity (Green et al., 2015). The common reasons for TRC failures include caries associated with restorations and sealants (CARS) (Dobloug and Grytten, 2015), and the loss of retention/marginal adaptation (Mjör and Gordan, 2002). Resealing of marginal discrepancies is a favourable procedure to limit the complex restorative therapy that allows conservation of the tooth structure and consequently increase the longevity of the tooth-restoration complex (Blum et al., 2014).

Glass ionomer cements (GICs) provide a reliable, long-term chemical adhesion to tooth tissues, low coefficient of thermal expansion, good tissue biocompatibility and fluoride release with a potential reduction in the incidence of CARS (Mayanagi et al., 2014). Glass ionomer cements set via an acid-base reaction when the polymeric acid reacts with the basic glass releasing Ca^{2+} and Al^{3+} ions, which crosslink with the polyalkenoic acid chains. The cement is sensitive to moisture in the early stages of setting and the maturation of the cement continues over time. These cements are capable of exhibiting adhesion through the formation of chemical bonds between the cations in the tooth and the anionic functional groups (Wilson and McLean, 1988; Guggenberger et al., 1998). The brittleness and low wear resistance of the conventional GICs with low tensile and flexural strengths limited their use as a restorative or even repair material in high stress-bearing areas (Zhao and Xie, 2011).

The incorporation of photopolymerisable monomers such as 2-hydroxyethyl methacrylate (HEMA) and their associated initiator systems (McLean, 1994, Nicholson, 1998) enabled a light activated setting of GIC's and the ampiphilicity allowed the organic/aqueous phase compatibility leading to the RMGICs (Antonucci, 1989, Mitra, 1991). They retain the advantages of GICs regarding the chemical adhesion with the tooth surface (Mitra, 1991), and fluoride release (Forss and Seppä, 1995) whilst allowing an extended working time, low moisture sensitivity and significantly improved diametral tensile and flexural strengths as well as fracture toughness as compared to CGIC (Guggenberger et al.,1998; Davidson and Mjör, 1999). Nevertheless, like all water-based cements, brittleness and low mechanical strength are remaind the weakness that may occasionally lead to fracture and wear as compared to the contemporary dental adhesive restoratives (Scholtanus and Huysmans, 2007). Additionally, the degradation effect caused by water uptake that competes with cement maturation might induce hydrolysis and plasticisation of the polyacrylate/polymer matrix which deteriorate the mechanical properties of RMGICs (Anstice and Nicholson, 1993; Kanchanavasita et al., 1997). This might be associated with the leachout of the water-soluble species such as HEMA (Yoshikawa et al., 1994), which affect the biocompatibility of the RMGICs regarding the cytotoxicity towards pulp cells (Stanislawski et al., 1999).

Phosphate functional monomers have been widely used in dental adhesives. They are mainly incorporated into the self-adhesive bonding system as an adhesion promoter. The acidity and reactivity of these functional monomers provoke strong chemical interactions with the hydroxyapatite minerals of the tooth structure (Münchow et al., 2015). They are expected to decalcify the hydroxy and carbonate apatites (HA/CHA) and chelate with the calcium ions simultaneously (Fu et al., 2005; Suzuki et al., 2006). Thus through, etching and chemisorbtioning a strong and stable chemical bonding to the tooth substrate occurs, which is believed to play a significant role in enhancing the bond durability and considered to be superior than, even better as compared to the carboxylic acid group interactions (Yoshida et al., 2000; Yoshida et al., 2004).

Ethylene glycol methacrylate phosphate (EGMP) is a photoreactive proton-conducting monomer with pendant phosphate groups. The complexation behaviour of the carbonyl and phosphoryl ligating groups bearing ethylene glycol methacrylate phosphate (EGMP) monomer has been reported to enable remineralisation in hydrogels (Kemal et al.,2011) due to the charge in the gel and also improve the bonding efficacy and durability of self-etching adhesives (Münchow et al., 2015).The EGMP-HEMA allows for the polymerisation, which is hypothesised to not only create a network of covalently linked phosphate groups but additionally improve the adhesion to resin-based composites, RMGIC/GIC’s and amalgams by virtue of the polar phosphate groups. Combining the properties of EGMP monomer as an adhesion promoter within RMGICs is a unique and interesting concept especially as the pendant phosphate groups are expected to interact with metallic cations during setting of the cement to form complexes that might alter the physical/biological properties of the cement itself and influence its adhesion to other dental and restorative substrates.

Thus, the aim of the chapter was to develop a new class of material based on RMGIC via the inclusion of the different proprtions of EGMP (10-40% wt.),and investigate the efficacy of the modified cement to be used for repairing failed TRCs. The influence of this monomer on setting time, mechanical properties (CS, CM, MH, and BFS), fluid uptake behaviour, fluoride ion release and bonding to sound dentine were reported. The hypothesis tested was the incorporation of different proportions of EGMP (10-40% wt.) in to a commercial RMGIC have no significant effect on their physical and bonding properties, and subsequent ageing has no bearing on the properties of the modified formulations.

3.2 Materials and Methods

3.2.1 Formulation and characterisation of the modified cement

Ethylene glycol methacrylate phosphate was purchased from Polysciences Europe GmbH, Germany (Batch No.: 52628-03-2, molecular weight 210.12 g/mol, density1.31 g/mL). The chemical structure is shown in Fig (3-1). Commercial RMGIC Fuji II LC (Improved), shade A2 (batch numbers 141118, and 1412081, GC Corp., Europe) was used as a control. It consists of a calcium fluoroaluminosilicate glass, and an aqueous solution containing 25-50% 2-hydroxyethyl methacrylate (HEMA), 5-10% polyacrylic acid, and 1-5% urethane dimethacrylate (UDMA), initiators and pigments.

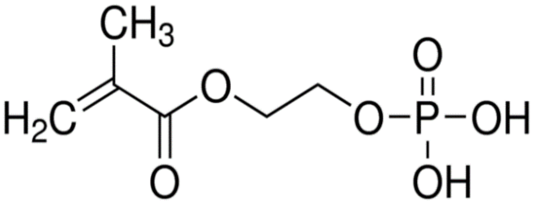


Figure 3-1 Chemical structure of ethylene glycol methacrylate phosphate

3.2.2 Specimen preparation

The new cements were prepared using four different proportions of ethylene glycol methacrylate phosphate monomer (10, 20, 30 and 40% by weight) blended with the liquid phase of the Fuji II LC. The powder component of the commercial RMGIC Fuji II LC was used without any modification in all cement formulations. The commercial Fuji II LC cement was used as a control (EGMP0), whereas the four experimental groups were prepared by hand-mixing of the commercial Fuji II LC glass powder with the modified liquid (EGMP10, EGMP20, EGMP30, and EGMP40) using a powder/liquid ratio 3.2/1.0, at ambient temperature (23±2°C) and humidity (35±5%). The cement formulations with their respective codes are shown in table 3-1. The substitution of the liquid phase with EGMP monomer up to 40% did not compromise the mixing and handling nor the resultant properties. This might be attributed to the comparable densities of both EGMP (1.31 g/mL) and HEMA (1.073 g/mL) which produced an initial viscosity that was comparable to the control cement. In addition this would also allow the assessment of the properties of the cement with the unmodified cement by keeping this parameter constant as P/L ratio can affect the properties of GIC and RMGIC. In order to reduce the P/L ratio variations among groups, specimens were prepared using fixed amount of powder and liquid to reduce variations and standardise the cement preparation for all formulations. The bottle was tapped and shaken to unsettle the powder, then dispensed using a levelled scoop. The powder was placed on a glass slab and separated into two equal parts. The liquid's bottle was tipped onto its side, inverted and squeezed gently allowing the dispensing of a clear drop without air bubbles. Half of the powder was mixed with the liquid for 10 s. The remaining powder was further mixed for 25 s in accordance with manufacturers' instructions. The freshly mixed cement pastes were placed in cylindrical polyethylene split moulds (4±0.5 mm diameter, 6±0.5 mm height) to prepare test specimens for determining the compressive strength (CS) and microhardness (MH) (ISO, 2.2010.9917- water-based dental cements). A stainless-steel mould of 8.3 mm diameter and 1.3 mm thickness was used to prepare the disc specimens for the biaxial flexural strength (BFS) test. Specimens were photo-polymerised after 3 min and 45 s from the mixing time using a light curing device (Elipar™ DeepCure-S LED, 3M USA) with a light intensity of 1470 mW/cm^2 for 30 s at each end of the cylindrical mould, and 20 s on the top surface of disc specimens. The curing light intensity was monitored with a light meter (Curing Radiometer Model 100, Demetron /Kerr, and Danbury, CT, USA). The CS, MH and BFS tests were carried out after 1, 14, 28 and 180 days storage in simulated body fluid (SBF) at 37°C. The solution was prepared following Kokubo and Takadamas' formula to examine the reactivity of the added phosphate based monomer for apatite induction at the cement surface in a simulated body fluid with ion concentrations nearly equal to those of human blood plasma (Kokubo and Takadama, 2006). Solutions were replaced on a weekly basis. The water uptake and fluoride release were also assessed to optimise the modified formulations in accordance to ISO guidelines (ISO 4049. Dentistry-resin based dental fillings. ISO; 2009).

Table 3-1 Experimental resin modified glass ionomer cements

Codes/ Liquid phase	**Solid phase**	**EGMP (wt %)**	**P/ L ratios**
EGMP0 *(Fuji II LC liquid)*	GC Fuji II LC powder	0	3.2 / 1.0
EGMP10	GC Fuji II LC powder	10	3.2 / 1.0
EGMP20	GC Fuji II LC powder	20	3.2 / 1.0
EGMP30	GC Fuji II LC powder	30	3.2 / 1.0
EGMP40	GC Fuji II LC powder	40	3.2 / 1.0

3.2.3 The determination of working and setting times

The measurements of the working and setting times are based on the ISO standard ISO 9917-2, 2010. RMGIC is classified as Class 3 materials in which the setting reaction is initiated following the mixing of components and also may be light-activated. These measurements were determined in the absence of activation radiation, since the materials hardened within specific time without light exposure, using an oscillating rheometer (Sabri Dental Enterprises, 1404 Brook drive, USA) at ambient temperature (23±2°C) and a humidity (50±20%). The cement powder and liquid were mixed with a powder/ liquid ratio of 3.2/1.0 using a spatula on a mixing paper for 25 s following the manufacturer instructions. The mixture was placed on a plate of the rheometer and pressed with a top plate. The instrument records the materials trace patterns of working time as a straight line and the setting time as wide oscillating pattern on the attached strip chart recorder for an easier determination of results with a high degree of accuracy shown in time (milliseconds). Measurements were made in triplicate.

3.2.4 Spectral analysis by Fourier transform-infrared spectroscopy (FTIR)

ATR/FTIR vibrational analysis (IR) was performed using a Perkin Elmer Spectrum One FTIR Spectrometer (Perkin-Elmer, Beaconsfield, UK) with a resolution of 4 cm^{-1}. The infrared spectra were recorded in the spectral range of 4000-600cm^{-1}, with eight scans each.

3.2.5 Mechanical properties

The mechanical tests (compressive strength (CS) and compressive modulus (CM), microhardness (MH), and biaxial flexural strengths (BFS)), and sample size (n=8 per each group) are based upon the specifications outlined in ISO, 9917-2 standard 2010 for the resin-modified cements, Geneva: ISO. 320 cylindrical specimens were prepared for CS and MH tests. 160 disc-shaped specimens (8.3±0.1 mm diameter, and 1.3±0.1 mm thickness) were prepared and tested for the biaxial flexural strength test. These properties were tested after 1, 14, 28, and 180 days storage in SBF at 37°C.

3.2.5.1 *Compressive strength (CS) and compressive modulus*

160 cylindrical specimens of each group were prepared for the compressive strength test (CS). After specific storage time (1, 14, 24 and 180 days), the diameters of each cylindrical specimen was measured at three points using a digital micrometre screw gauge with an accuracy of 10 µm (Moore and Wright, Sheffield, England), and the mean diameter was calculated prior to testing. To mimic the oral environment, the specimens were tested 'wet' by placing a wet filter paper (Whatman No. 1, Whatman International Ltd., Maidstone, England) between the two ends of the specimen and test machine platens. A universal testing machine (Instron model 5569, USA) with a 500 N load cell was used for testing the compressive strength and modulus at a crosshead speed of 0.5 mm/min. The compressive load was applied to the long axis of each specimen and the maximum load to failure was recorded (Fleming et al., 2003). The compressive strength, *P* (MPa), of each individual cylindrical specimen was calculated by dividing the fracture force (*F*) by the area of the specimen where *D* was the specimen diameter using Eq. (1), (Dowling and Fleming, 2009)

$$P = \frac{4F}{\pi D^2} \qquad \text{Eq (1)}$$

Where *F* was the load at fracture (N) and *D* was the mean specimen diameter (mm).

In order to assess the mechanical deformation behaviour, the compressive stress-strain curves were obtained for each specimen. The compressional stress (σ) was determined by dividing the applied force *F* (N), by the cross-sectional area *A* (mm^2), Eq (2):

$$\sigma = \frac{F}{A} \qquad \text{Eq (2)}$$

The strain was measured by dividing the change in length by the original length, Eq (3):

$$\boldsymbol{\varepsilon} = \frac{\Delta l}{l} \qquad \text{Eq (3)}$$

Δl is the extension (change in length), while l is the original length.

Individual stress/strain plots were measured for each cylindrical specimen tested in compression. Then the compressive modulus (the ratio of stress to strain below the fracture limit) was determined by calculating the slope of the initial linear segment of the stress-strain curve (Xie et al., 2000; Dowling and Fleming, 2009; Samuel et al., 2009).

3.2.5.2 *Surface microhardness*

At the end of the specified storage time (1, 14, 24 and 180 days), the surface hardness of each cylinder from each group (n=8 per group) was determined using Knoop hardness test (Duramin10, Struers, Japan) at ambient temperature (23±2°C) and humidity (35±5%). Each specimen was placed on the instrument platform with the tested surface facing the indenter that was brought into contact with this surface. A minor load of 1 gf (0.01 N) was applied to the surface for 5 s to ensure the contact between them (Kanchanavasita et al, 1998), while the depth indicator at the dial gauge was adjusted to zero. The indentations were performed using 50 gf load force for 10 s, since the application of higher loads or a longer contact time can initiate cracks at the surface of these cements (De Moor and Verbeeck, 1998). Under these conditions, the sharply defined indentation marks were obtained with a size allowing the determination of the surface hardness with a sufficient accuracy. The Knoop Hardness Numbers (KHN) were recorded as an average of 6 readings at randomly selected areas which are at least 1 mm far away from the adjacent indentations or the margin of the specimens.

3.2.5.3 *Biaxial flexural strength*

Thirty two disc-shaped specimens (8.3±0.1 mm diameter, 1.3±0.1 mm thickness) of each group were prepared and tested for the BFS test after 1, 14, 28 and 180 days. The dimensions were determined to the nearest 0.001 mm using a digital micrometre (Micro 2000, Moore and Wright Ltd., Sheffield, England). The measurements were made at three different sites on the specimens. The tests were conducted at the ambient temperature (23±2°C) and humidity (35±5%). The specimen was placed centrally on a 6.5 mm diameter circular support in such a manner that the edge extended beyond the support by the same amount around the whole specimens. Then, this specimen was centrally loaded with a 1.5 mm diameter round ended indenter in a way that the area of maximum tensile stress was located at the centre of the lower face of the disc, as shown in Figure 2-1. The load was applied using a universal testing machine (Instron Model 5569, USA) at a crosshead speed of 0.5 mm/min until the specimens yielded or fractured. The load at failure was obtained directly from the loading curves. Each fractured specimen was inspected for significant voids and irregularities. When the fractures occurred at some obvious voids or flaws, the specimen was excluded from further analysis (Kanchanavasita et al., 1986). BFS values were calculated using the equations (4-6) that are mentioned in chapter two.

3.2.6 Mass change during water uptake

The control and experimental specimens were prepared using disc shaped moulds (10 mm diameter,1 mm thickness) at (23 ± 1) °C, following ISO standard 4949:2009. The thickness and diameter of each specimen (n=5 per each group) were measured at 4 and 2 points respectively, using a digital electronic calliper (Mitutoyo Corpo-ration, Japan). Mean values were used to calculate the volume of each specimen in mm^3. Initial weights were measured for each sample (w_0) using an electronic balance (Mettler Toledo XS105DU, Switzerland) to an accuracy of ±0.0001 g. Specimens were immersed in 10 ml distilled water at 37°C in individual plastic containers for total immersion time of 60 days. At defined time intervals, the specimens' surfaces were gently dabbed on a filter paper and weighed. Care was taken so the whole process did not take more than 20-30 s to avoid water loss. Several readings (w_t) were taken on the first day, daily for a week, then weekly thereafter until equilibrium was achieved, indicated by four successive weights being the same. The mass recorded at equilibrium was denoted as (w_e)

$$Weight\ change\ (\%) = \frac{w_t - w_0}{w_0} \times 100 \qquad \text{Eq (7)}$$

Where w_t, is the weight at t time, w_0 is the initial weight of the specimen before immersion in the solution. The mean weight change (%) and standard deviation (SD) during water uptake were plotted against time$^{1/2}$(seconds) to create the weight change profile for each tested group. After equilibrium was reached the specimens were allowed to dry at 37°C to obtain water loss, until reaching a constant weight (w_d). The water sorption (WSP) in µg/mm^3 at the equilibrium stage was also calculated by:

$$WSP = w_e - w_d / V \qquad \text{Eq (8)}$$

w_e weight at the equilibrium after uptake stage, w_d the constant weight after dryness, *V* volume (Zankuli et al., 2014). The solubility percentage for the tested materials was calculated by subtracting the weight after desorption (w_d) from the initial specimen weight (w_0), Eq (9). This is equivalent to the total mass of components leached from the material.

$$Solubility\ (\%) = \left(\frac{W_0 - W_d}{W_0}\right) \times 100 \qquad \text{Eq (9)}$$

W_0 initial weight, W_d weight at the equilibrium after desorption.

The solubility (*W*SL) in µg/mm^3 calculated using the following equation:

$$WSL = W_0 - W_d / V \qquad \text{Eq (10)}$$

W_0 initial weight, W_d weight at the equilibrium after desorption, *V* volume.

The water uptake data were plotted as Mt/M∞ against time$^{1/2}$ (seconds$^{1/2}$) to obtain the slope, which was then used to calculate the diffusion coefficient for the water uptake process, using Eq (11), (Crank, 1979; Agha et al., 2016).

$$D = \frac{s^2\, \pi\, l^2}{4} \qquad \text{Eq (11)}$$

Where *s* = slope of graph; Mt= the mass uptake/loss at time *t* (s); M∞= equilibrium uptake/loss, *l* = the thickness of the specimens, and *D* is the diffusion coefficient.

3.2.7 Fluoride release

Fluoride ion release measurements were recorded over 60 days (n=5 per each group) using disc-shaped specimens (8.3 mm diameter and 1.3 mm thickness). Each specimen was immersed in an individually capped polystyrene tube containing 2 ml of distilled water (pH 7.0) and stored at 37°C for a total immersion time of 60 days. The storage medium was refreshed every 48 h to avoid fluoride saturation in the solution, and fluoride concentration was measured daily for the first week and weekly up to 60 days (Geurtsen et al., 1999). An equal volume (2 ml) of total ionic strength adjustment buffer (TISAB I BDH Ltd., Poole, England) was added prior to fluoride ion measurements, which increases the ionic strength of the solution to a relatively high level and hence increases the accuracy of the reading. Fluoride concentrations were recorded in ppm using a selective fluoride electrode (Cole Parmer 27502) connected to an ion analyser (OAKTON 510 ion series, Singapore). The amount of fluoride eluted from the RMGICs was converted into milligrams of F^- released per unit surface of area (mg F/cm^2) (Fukazawa et al., 1987).

3.2.8 Bonding to sound dentine

The bonding efficacy of the experimental cements (EGMP10, EGMP20, EGMP30 and EGMP40) to sound dentine was evaluated and compared to the commercial cement (EGMP0) by determining the microtensile bond strength test (µTBS) and using scanning electron microscopy (SEM) to analyse the mode of failure. Ten permanent sound molars were collected using an ethics protocol reviewed and approved by NHS health research authority (16/SW/0220). All teeth were stored at 4°C in distilled water and used within one month. Teeth were randomly and equally assigned to five groups based on the RMGICs used (four experimental and one commercial, n=2). The occlusal enamel was removed using a low-speed water-cooled diamond saw microtome (Isomet 1000, Buehler, Lake Bluff, IL, USA). Teeth were examined using stereomicroscope to check the dentine surfaces and the absence of any remnants of enamel. Dentine surface of each tooth was polished for 60 s using 600 grit polishing paper to standardise the smear layer, followed by the application of dentine conditioner (10% polyacrylic acid, GC Corp) for 20 s to remove the created smear layer. The conditioner was washed with air/water spray for 15 s and dried with a gentle stream of dry compressed air for 15 s. A matrix band was sured around each specimen, and the RMGICs (control and experimental groups) were placed over the dentine surfaces and photo-polymerised for 40 s using a light curing device (Elipar™ DeepCure-S LED, 3M USA) with a light intensity of 1470 mW/cm^2. The restored specimens were stored at 37°C and 100% humidity for 24 h before sectioning. Then, each tooth was sectioned with the Isomet saw through the RMGIC build-ups and dentine at 0.9 mm increments, to produce a series of 0.9 mm-thick slabs. Individual slabs were further sectioned occluso-gingivally to produce 0.9 mm x 0.9 mm beams, with the RMGICs comprising the upper half of the beam and dentine comprising the lower half. Each tooth yielded ≃24 beams for bond testing. Twenty beams per group were selected and stored for two weeks in SBF at 37°C. Beams with premature bond failure during storage time were assigned a null bond strength value and were included in the compilation of the mean tensile bond strength as well as the failure mode assessment. Specimens were stressed to failure under tension using a universal testing machine (SMAC Europe Ltd, Crawley, UK) at a crosshead speed of 0.5 mm per min, according to the technique reported by Shono et al. (1999). Bond strength data obtained were analysed via one-way ANOVA and compared with Bonferroni post-hoc tests at significance, *p*= 0.05. The failure modes of the bonds were initially evaluated at x40 with a stereoscopic microscope. Failures were classified as interfacial failure between dentine and the RMGIC, cohesive failure within the RMGIC/ or dentine, and mixed (combinations of cohesive failure in the RMGIC/dentine and interfacial failure along the dentine surface), Fig. 3-2.

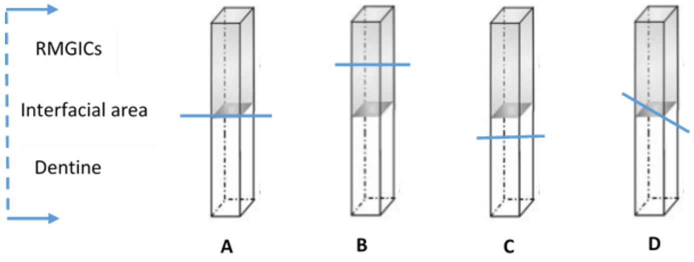


Figure 3-2 Mode of failure, A: adhesive failure, B: cohesive failure within the RMGICs, C: cohesive failure within the dentine, D: mixed failures

3.2.9 Scanning electron microscopy and energy dispersion X-ray spectroscopy

Representative surfaces from mechanical testing (CS and BFS) were dried, carbon-coated, and viewed under scanning electron microscope (JCM-6000 PLUS, NeoScope - Benchtop SEM, USA) with an accelerating voltage of 10kV. Scanning electron micrographs of the fractured specimens from CS test showed the microstructural changes for selected cement formulations (EGMP0, EGMP20, and EGMP30) at different magnifications (x50, x100 and x400) and a working distance of 500µm, 200µm, and 50µm, respectively. Scanning electron micrographs at x50, x600, and x1000 magnification and working distance 500µm, 50µm, and 20µm, respectively, were also performed to assess the surface morphology of the fractured specimens from BFS test for all formulations after four weeks' storage in SBF at 37°C. These are coupled to an energy dispersive X-ray spectroscope (EDX) (JCM-6000 PLUS, JED-2300 Analysis Station Plus, USA) to perform elemental analysis for all tested cements. This was carried out to detect the structural changes within the cement with the incorporation of different proportions (10-40%) of EGMP monomer. For the µTBS test, SEM of representative debonded specimens of the EGMP0 and EGMP30 only (n=2 per group) which showed mixed or adhesive failures were obtained. Specimens were dried and gold coated at 45 mA currents for 2 minutes and viewed under a SEM (JCM-6000 PLUS, NeoScope - Benchtop SEM, USA) at two magnification power (x100, and x1000) and working distance (200µm, and 20µm, respectively).

3.2.10 Statistical analysis

Data were tested for normality using Q-Q plots and Shapiro-Wilk tests and were analysed parametrically as the data followed a normal distribution. One-way analysis of variance (ANOVA) and Bonferroni HSD post hoc tests were employed to calculate significance (alpha level=0.05) in mean values amongst the tested groups at each time interval. Independent t-tests ($p<0.05$) was also applied to determine the effects of different storage time on the mechanical properties per each group. All analyses were conducted using SPSS statistical package (version 24; SPSS® Inc., IBM®, Chicago, IL, USA).

3.3 Results

3.3.1 Working and setting times

The effect of incorporating different proportions of EGMP, a phosphate based monomer to the RMGIC (Fuji II LC) on working and setting time is summarised in Table 3-2. The addition of 10% and 20 wt.% of EGMP monomer did not significantly affect the working time of the new cement (p>0.05). In contrast, higher concentrations of EGMP (30% and 40wt %) significantly prolonged the working time (4.2 and 4.0 min, respectively) as compared to the commercial product (3.45 min). Nevertheless, the working time of the EGMP-containing cements were within acceptable limits as stipulated by the ISO standards and meet the requirements for water-based cements (ISO, 9917-2, 2010). The setting time of all formulations were comparable to the control cement and remained unaffected by the inclusion of EGMP at the concentrations studied.

Table 3-2 the working and setting time of the RMGICs

Groups/ n=3	**Working time (min)**	**Setting time (min)**
EGMP0	3.45± 0.2	5.33± 0.1
EGMP10	3.44± 0.1	5.40± 0.2
EGMP20	3.46± 0.2	5.43± 0.2
EGMP30	4.24±0.1*	5.63± 0.2
EGMP40	4.00± 0.1*	5.60± 0.2

(*) significant difference of the experimental RMGICs from the control group (Bonferroni test post-hoc tests, alpha level of 0.05).

3.3.2 Spectral analysis by FTIR

The ATR-FTIR spectrum of EGMP-PAA liquid mixtures are shown in Figure 3-3 and compared with the liquid phase of the RMGIC. The intense peak ~1700 cm^{-1} arises due to the carbonyl stretching frequencies from the ester and carboxylic acid groups of HEMA, EGMP and the polyacid (Young et al., 2000; Young, 2002). The peaks arising at ~1630, and 1300 cm^{-1} indicate the methacrylate C=C bond stretch with α-methyl group vibrations at 1375 cm^{-1}. The bands at 1404, 1451 cm^{-1} attributed to the =CH2 deformation, and bending vibration of C-H bonds of the -CH3 group, respectively (Young et al., 2000). The absorption band at ~1169 cm^{-1} is attributed to the C-O-C stretching vibration (MacDonald et al., 2000). The modified liquid exhibited absorption peaks at 988-1006 cm^{-1} due to the presence of the phosphate groups of EGMP that are absent in the control group, moreover the peak height at 988 cm^{-1} increases with increasing EGMP indicating the inclusion of the monomer. All groups showed peaks at 947 cm^{-1} with different intensity which might be referred to the C-O acid band. The bands at 2934 and 2959 cm^{-1} are assigned to the C-H bond stretching vibrations of the -CH3 and -CH2- groups, respectively and the peak at 3391 cm^{-1} can be attributed to the -OH group stretching vibrations.

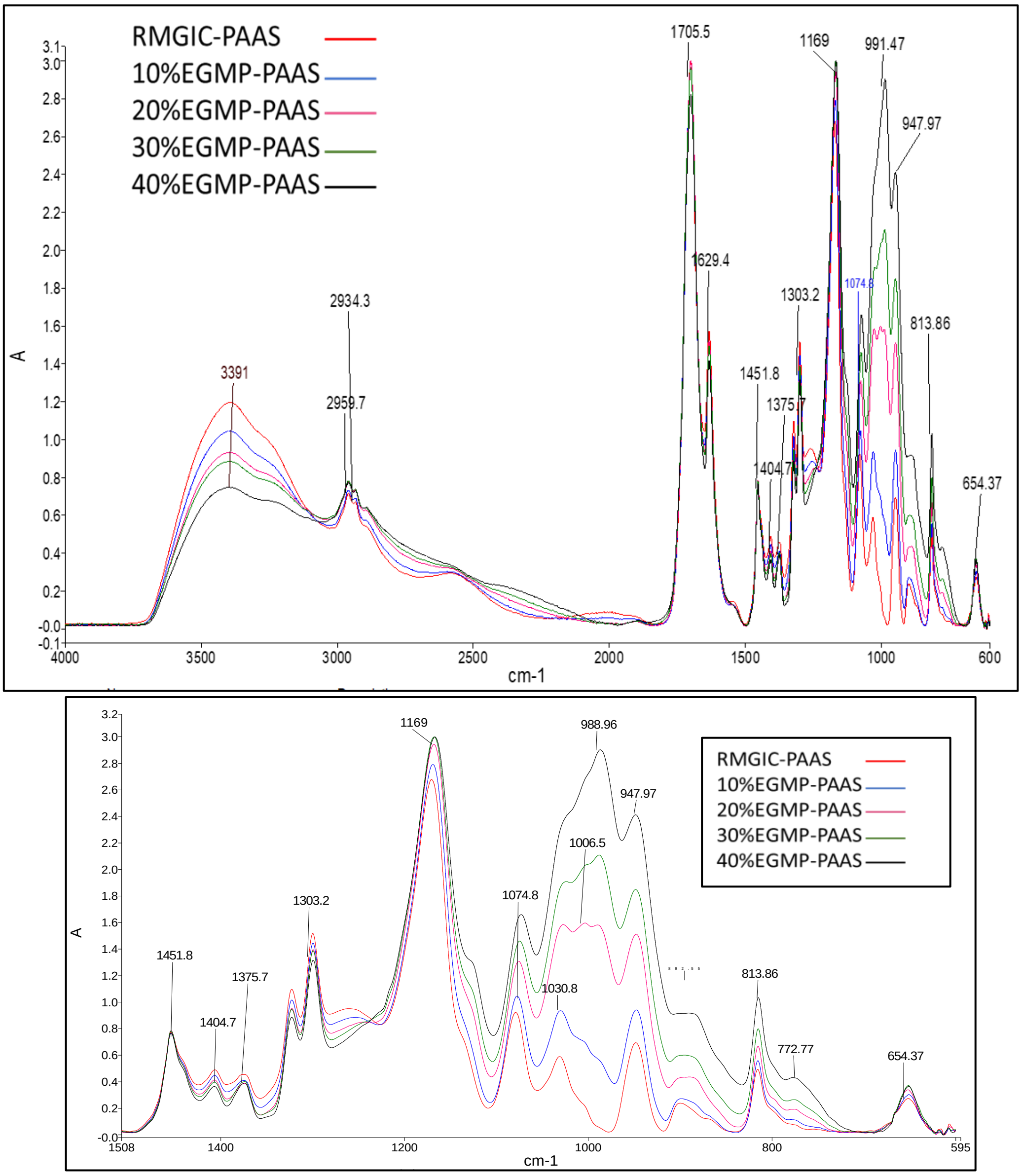


Figure 3-3 The ATR-FTIR spectrum of EGMP-PAA liquid mixtures compared with the liquid phase of the RMGIC.

The ATR-FTIR spectrum of RMGIC and the pRMGIC cements after 24 h immersion in SBF is shown in Figure 3-3a. A comparison of the RMGIC with the pRMGIC show differences due to inclusion of the phosphate moiety present in the pRMGIC that appears at 966 cm^{-1} as a shoulder of the peak at 1024 cm^{-1} arising due to the stretching vibration of SiO_4 tetrahedral with different number of bridging oxygen atoms. This is a very intense and broad peak and has considerable overlap with P-O stretching modes hence the peaks at the lower EGMP concentration is masked to an extent (MacDonald et al., 2000), however with increasing concentration of EGMP in the pRMGIC a shoulder centred at around 966 cm^{-1} becomes apparent due to the v1 stretching vibrations in the PO_4 tetrahedral structure of the phosphate moiety. The characteristic peaks of carboxylate salt formation with the symmetric and asymmetric -COO stretching bands are evident at 1429 cm^{-1} and 1598 cm^{-1}, respectively (Nicholson, 1998). The absorption band at 1700 cm^{-1} assigned to C=O stretching vibration of the ester group of HEMA, EGMP, and COOH group in polyacid (Young, 2002). The broad band around 3356 cm^{-1} is associated with the O-H stretching vibrations of water (Fu et al., 2005).

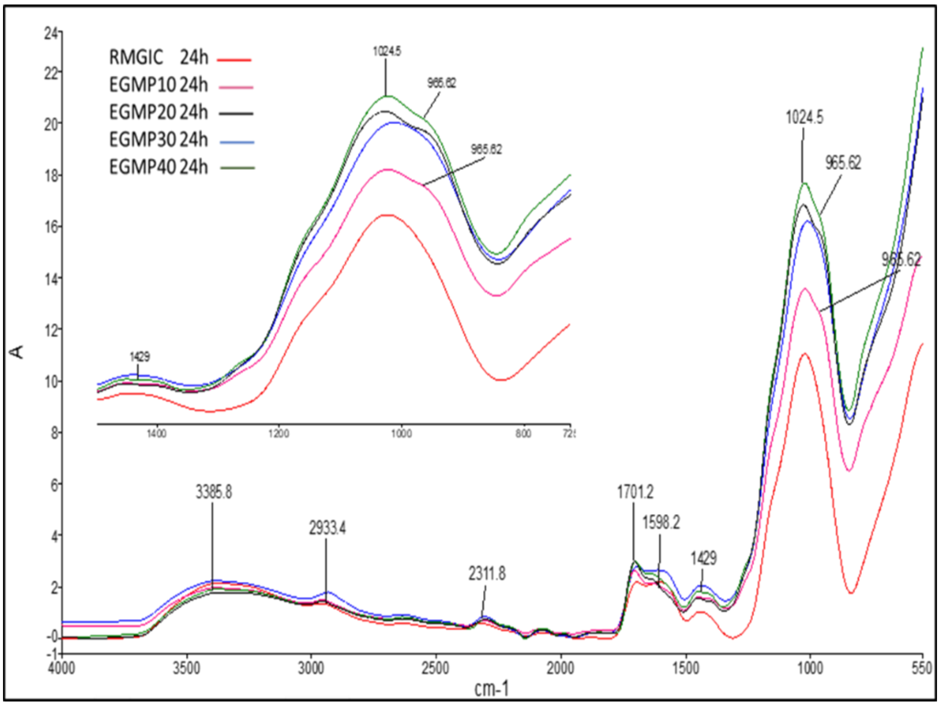


Figure 3-3a The ATR-FTIR spectrum of RMGIC and the pRMGIC cements after 24 h immersion in SBF.

The FTIR spectrum of the EGMP-containing cements and control RMGIC after 4 weeks immersion in SBF is shown in Figure 3-3b. The spectra of the modified cements showed the characteristic peaks of the polyacrylate salt formation with symmetric and asymmetric -COO stretching bands at approximately 1370, 1450 and 1580 cm^{-1}. The aluminium polyacrylate peaks were 1335, 1460 and 1570 cm^{-1} (Matsuya et al., 1996). The peak at 1024 cm^{-1}, and a new band at 966 cm^{-1}, being more prominent with increasing the EGMP content within the matrix. These peaks (1024, and 966 cm^{-1}) are assigned to *v*3 and *v*1 stretching vibrations in the PO_4 tetrahedral structure, which might indicate precipitated minerals within the cement matrix with time (Rey et al., 1990; Gadaleta et al., 1996). In spectra of all cements (experimental and commercial), there is an absorption band around 1728 cm^{-1} assigned to C=O stretching vibration of the ester group of HEMA, EGMP, and COOH group in polyacid (Young, 2002). Furthermore, the absence of bands at 1322 and 1300 cm^{-1} referred to a quantified polymerisation reaction within the RMGICs (Rueggeberg et al., 1990; Pianelli et al., 1999). Moreover, a broad band around 3356 cm^{-1} is associated with the O-H stretching vibrations of water molecules (Fu et al., 2005).

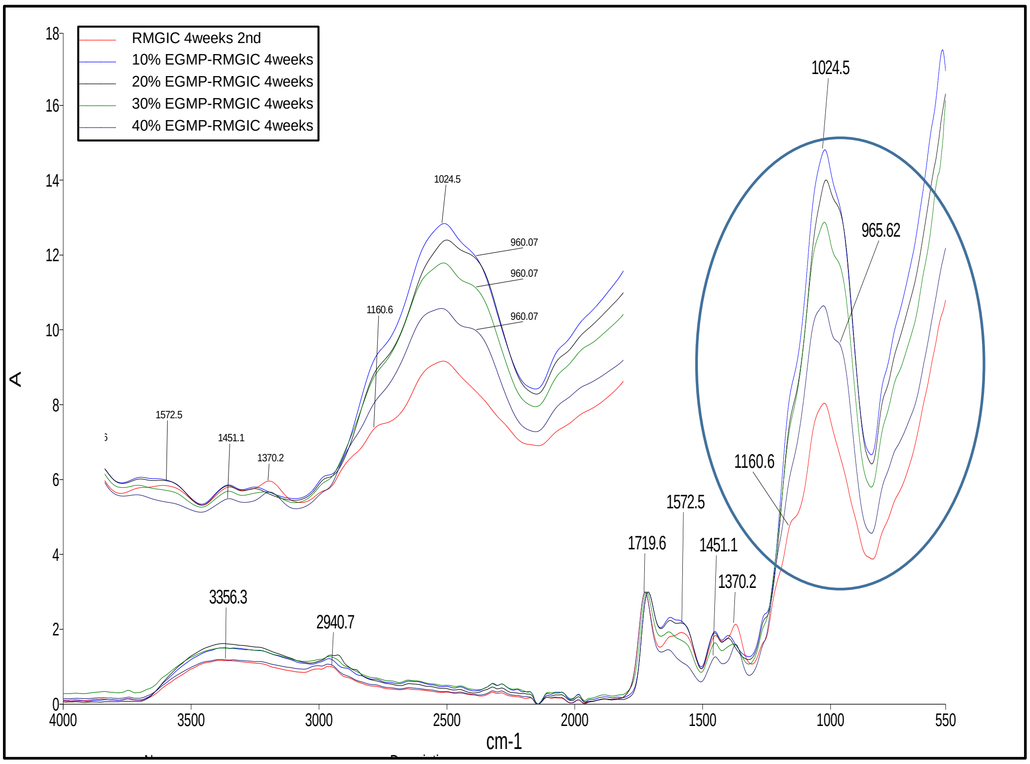


Figure 3-3b The FTIR spectrum of the EGMP-containing cements and control RMGIC after 4 weeks immersion in SBF.

3.3.3 Mechanical properties

The immediate compressive strength of the experimental cements (EGMP10, 30 and 40) was similar to the control, Table 3-3. After 14 days of storage in SBF at 37°C, all groups exhibited enhanced CS in comparison to their early values ($p<0.05$) with no statistically significant differences among them. The higher EGMP-contained cement (EGMP30 and 40) continued gaining the strength after 4 weeks ageing (141.0±9 and 140.4±8 MPa, respectively) that are significantly higher than EGMP0 (128.8±7 MPa) ($p<0.05$). On long-term ageing (180 days), the CS was deteriorated in the control group but maintained in all modified formulations that showed statistically significant differences from the commercial reference and their initial values ($p<0.05$).

Table 3-3 Compressive strength CS (MPa) of the EGMP-RMGICs (0-40%) at 1, 14, 28, and 180 days, shown as mean (SD), n = 8.

Days	**EGMP0**	**EGMP10**	**EGMP20**	**EGMP30**	**EGMP40**
1	108.3 (6.5)^{a}	102.9 (8.4)^{a}	119.6 (7.4)^{*b}	117.3 (7.3)^{ab}	116.7 (7.4)^{ab}
14	131.0 (7.6)$^{c\wedge}$	132.7 (7.9)$^{c\wedge}$	131.8 (7.8)$^{c\wedge}$	138.9 (9.5)$^{c\wedge}$	131.6 (6.0)$^{c\wedge}$
28	128.8 (7.5)$^{d\wedge}$	131.4 (7.1)$^{de\wedge}$	132.9 (6.6)$^{de\wedge}$	141.0 (8.8)$^{*e\wedge}$	140.4 (8.0)$^{*e\wedge}$
180	107.8 (7.8)	121.9 (8.4)$^{*f\wedge}$	131.5 (8.0)$^{*fg\wedge}$	138.9 (8.4)$^{*g\wedge}$	121.3 (7.1)^{*f}

(*) significant difference of the experimental RMGICs from the control group, (^) significant effect of ageing for the same group from the day1 values with in each column. Similar letters in rows indicate no significant differences among groups (Bonferroni test post-hoc tests, alpha level of 0.05).

EGMP-RMGIC demonstrated a significant enhancement in the compressive modulus in comparison to the control cement at most time points ($p<0.05$). The recorded values increased proportionally with higher EGMP content. Prolonged ageing raised the CM of high EGMP-containing formulations (EGMP20, 30, and 40) as compared to the control group, and their corresponding immediate values ($p<0.05$), Table (3-4).

Table 3-4 Compressive modulus CM (GPa) of the EGMP-RMGICs (0-40%) at 1, 14, 28, and 180 days, shown as mean (SD), n = 8.

Days	EGMP0	EGMP10	EGMP20	EGMP30	EGMP40
1d	2.4 (0.4)	3.2 (0.3)^{*a}	3.4 (0.3)^{*ab}	3.8 (0.3)^{*bc}	3.9 (0.3)*c
14	3.1 (0.3)$^{\wedge}$	4.0 (0.3)$^{*d\wedge}$	4.3 (0.4)$^{*de\wedge}$	4.5 (0.2)^{*ef}	4.9 (0.3)$^{*f\wedge}$
28	2.8 (0.3)	3.7 (0.4)^{*g}	3.8 (0.4)^{*gh}	4.2 (0.5)^{*gh}	4.4 (0.3)^{*h}
180	3.0 (0.3)$^{i\wedge}$	3.6 (0.6)^{ij}	4.2 (0.4)$^{*jk\wedge}$	4.5 (0.3)$^{*k\wedge}$	4.5 (0.4)$^{*k\wedge}$

(*) significant difference of the experimental RMGICs from the control group, (^) significant effect of ageing for the same group from the day1 values with in each column. Similar letters in rows indicate no significant differences among groups (Bonferroni test post-hoc tests, alpha level of 0.05.

The presence of phosphate groups enhanced the microhardness of the RMGIC. However, the immediate effect was noticed in EGMP30 and 40 only, but after storage, all experimental groups showed statistically significant higher values from the control up to 6 months and from their initial values (p<0.05), Table (3-5).

Table 3-5 Microhardness MH (KHN) of the EGMP-RMGICs (0-40%) at 1, 14, 28, and 180 days, shown as mean (SD), n = 8

Days	EGMP0	EGMP10	EGMP20	EGMP30	EGMP40
1d	31.7(1.8)^{a}	33.0 (1.3)^{a}	35.8 (3.0)^{ab}	38.3 (3.1)^{*bc}	39.9 (3.9)^{*b}
14	32.0(1.9)	39.6 (2.9)$^{*\wedge}$	43.3 (1.7)$^{*\wedge}$	48.8 (2.0)$^{*d\wedge}$	46.8 (2.3)$^{*d\wedge}$
28	29.7 (3.3)	42.1(2.9)$^{*e\wedge}$	45.1(3.4)$^{*ef\wedge}$	49.0(2.7)$^{*f\wedge}$	45.9(3.6)$^{*ef\wedge}$
180	31.5 (2.8)	36.4 (3.3)$^{*f\wedge}$	39.2 (3.5)^{*fg}	46.6 (2.7)$^{*\wedge}$	45.0(2.8)$^{*g\wedge}$

(*) significant difference of the experimental RMGICs from the control group, (^) significant effect of ageing for the same group from the day1 values with in each column. Similar letters in rows indicate no significant differences among groups (Bonferroni test post-hoc tests, alpha level of 0.05

The profound effect of the added EGMP was seen in the biaxial flexural strength which showed a two-fold increase in values (p<0.001) in comparison to the control at all time intervals. Ageing shows a variable effect on the values among the experimental groups, but EGMP20 and 30 maintained high flexural strength up to 6 months storage, Table (3-6)

Table 3-6 Biaxial flexural strength BFS (MPa) for the EGMP-RMGICs (0-40%) at 1, 14, 28, and 180 days, shown as mean (SD), n = 8.

Days	**EGMP0**	**EGMP10**	**EGMP20**	**EGMP30**	**EGMP40**
1	121.8 (7.3)	249.2 (15.7)^{*a}	282.7 (15.2)*	228.2 (14.6)^{*a}	200.9 (14.9)*
14	143.6 (16.6)$^{\wedge}$	291.7 (15.7)$^{*b\wedge}$	290.6 (17.8)^{*b}	265.6 (20.2)$^{*b\wedge}$	219.1 (15.7)*
28	94.6 (8.6)$^{\wedge}$	241.1 (11.7)^{*c}	269.0 (16.4)^{*d}	254.2 (11.3)$^{*cd\wedge}$	246.0 (13.9)$^{*cd\wedge}$
180	133.6 (12.1)	231.6 (11.5)^{*e}	251.4 (13.1)$^{*f\wedge}$	253.8 (9.5)$^{*f\wedge}$	237.2 (11.6)$^{*ef\wedge}$

(*) significant difference of the experimental RMGICs from the control group, (^) significant effect of ageing for the same group from the day1 values with in each column. Similar letters in rows indicate no significant differences among groups (Bonferroni test post-hoc tests, alpha level of 0.05.

3.3.4 Mass change during water uptake

EGMP-modified cements exhibited higher water uptake percentages as compared to the control (EGMP0) which increased proportionally with increasing the EGMP content within the matrix, Fig (3-4). All groups reached equilibrium within a week. The water uptake was calculated in μg/mm^3 after 60 days incubation in distilled water 37°C. All experimental cements (EGMP10-40) recorded higher water uptake values (56.7- 60.7 μg/mm^3, respectively) than the control cement (54.7 μg/mm^3).

The solubility (percentages and in μg/mm^3) was significantly lower in EGMP-RMGIC in comparison to the control cement (EGMP0) (*p*<0.001), Table (3-7). The recorded values (3.3-2.2 μg/mm^3) are below the maximum recommended by the ISO 4049 (7.5 μg/mm^3). All groups showed a straight line when Mt/M∞ plotted against time (t^{½}) which indicates that the kinetics of water uptake follows Fick's law of water diffusion, Fig 3-6. Diffusion coefficient was decreased with increasing concentration of EGMP.

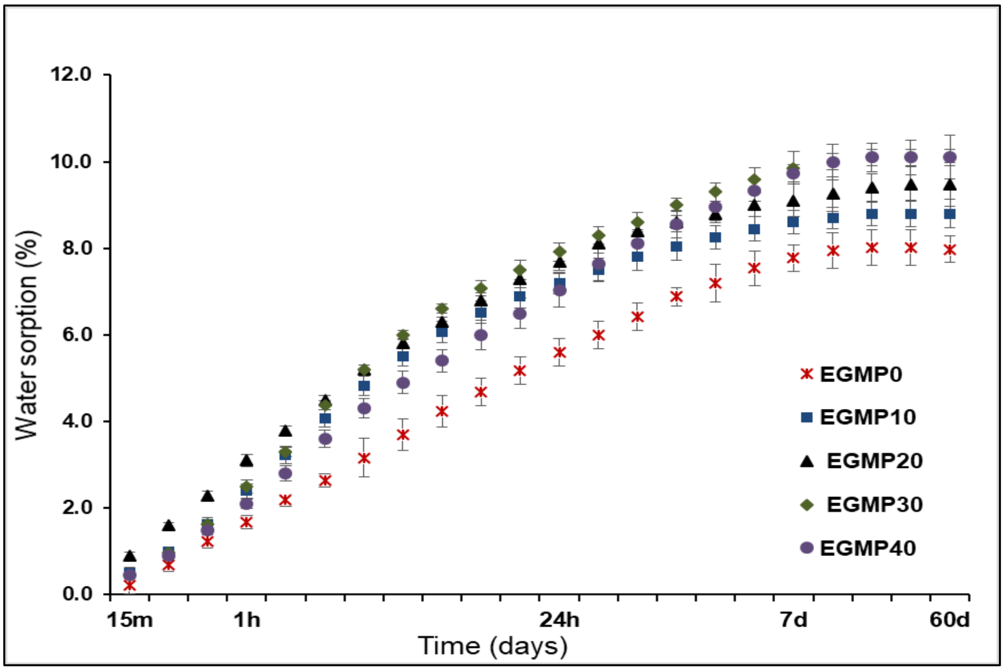


Figure 3-4 Weight change percentages of the control and EGMP-RMGICs over time

Table 3-7 water uptake parameters of the control and experimental RMGICs up to 60 days in distilled water 37°C (n=5).

Days	**Equilibrium water uptake**		**Solubility**		**Diffusion coefficient**
	%	**µgmm³**	**%**	**µgmm³**	**($10^{-11}m^2s^{-1}$)**
EGMP0	7.9 (0.3)	54.7 (2.0)	1.5 (0.10)	9.6 (0.3)	2.29
EGMP10	8.8 (0.3)	56.7 (2.0)	0.6 (0.02)*	3.3 (0.1)*	2.04
EGMP20	9.4 (0.3)*	57.5 (1.6)	0.4 (0.05)*	2.5 (0.1)*	1.89
EGMP30	10.1(0.4)*	58.5 (1.9)	0.4 (0.06)*	2.2 (0.1)*	1.52
EGMP40	10.1(0.4)*	60.7 (1.3)*	0.3 (0.04)*	2.2 (0.1)*	1.02

(*) significant difference of the experimental RMGICs from the control group. One-way ANOVA with Bonferroni post-hoc test, an alpha level of 0.05.

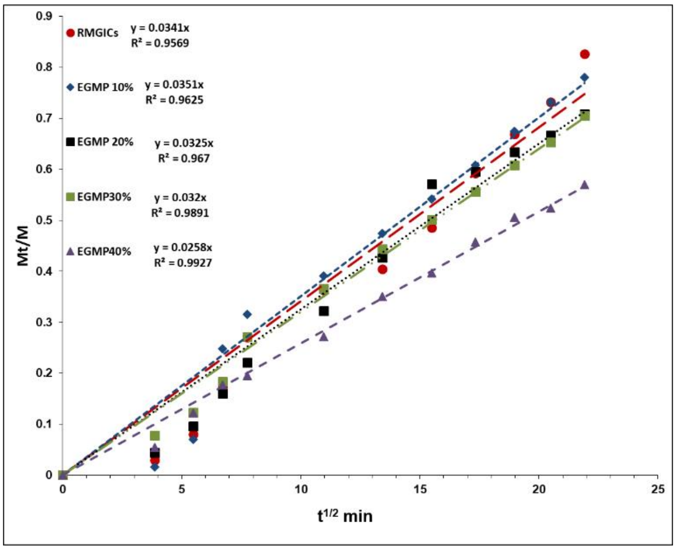


Figure 3-5 early water uptake behaviour and linear fit of the RMGICs

3.3.5 Fluoride release

The fluoride release up to 60 days are presented in Figure 3-6. Short-term fluoride release was significantly lower in experimental cements (EGMP10-40) in comparison to the control group (p<0.05). The reduced elution was proportional to the amount of EGMP in the matrix at the early period. However, these correlations did not exhibit statistically significant difference (p>0.05) after seven days immersion in distilled water, showing a similar pattern of release up to 60 days.

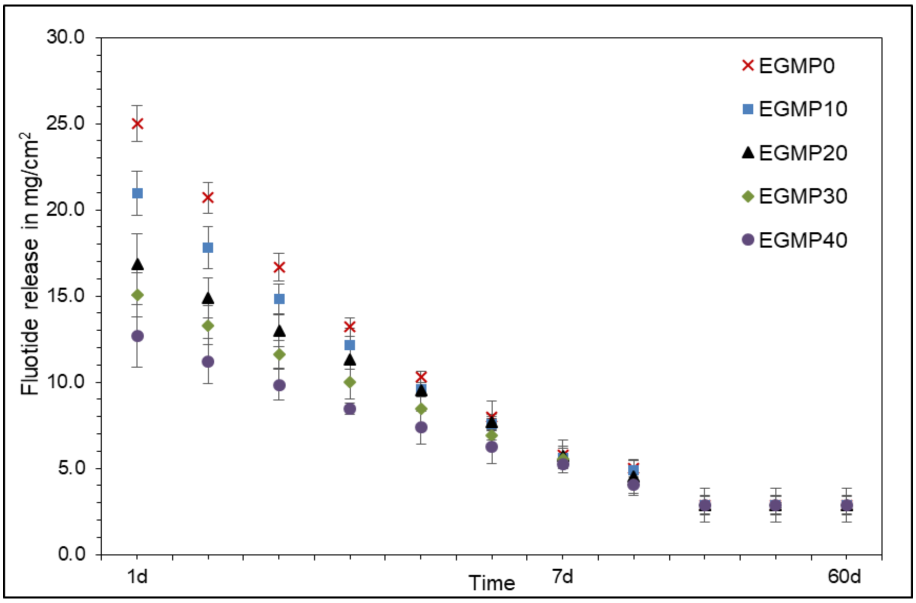


Figure 3-6 Fluoride release in mg/cm^2 over 60 days

3.3.6 Bonding to sound dentine

The results of the microtensile bond strength and mode of failure are presented in Figure 3-7, and 3-8, respectively. One-way ANOVA revealed statistically significant differences among tested groups (*p*<0.001). Further analysis using Bonferroni multiple comparison tests (*p*<0.05) showed that the addition of 20-40% by weight a phosphate-based monomer to the commercial RMGIC had significantly (*p*<0.05) enhanced its adhesion strength to sound dentine after two weeks storage in SBF at 37°C. EGMP10 presented similar adhesion strength to the control (*p*=1.000). The bond strength values are comparable between EGMP20, EGMP30, and EGMP40 (*p*=1.000), Figure 3-7. Interfacial failure analysis revealed higher adhesion failure mode in the control and EGMP10 and 20 (~40%). All groups showed cohesive failure within the cement, but it was higher in EGMP30 and 40. Cohesive failure within dentine was also seen in all groups, but it was higher in EGMP10, 20, and 40. Additionally, mixed failure could be recognised in EGMP30, 40, and the control, Figure 3-8.

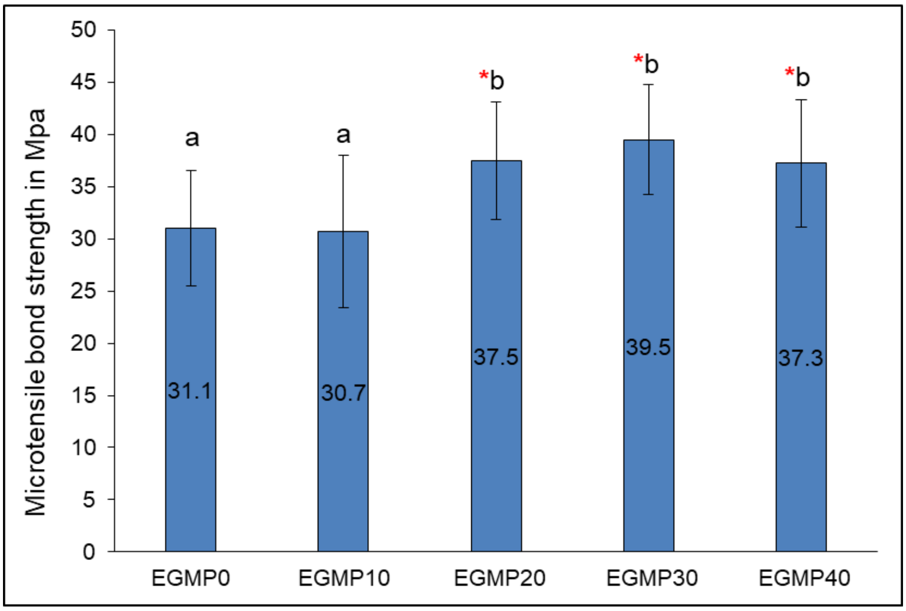


Figure 3-7 Mean microtensile bond strength (µTBS) of the experimental RMGICs and the control. (*) indicate a statistically significant difference of an experimental group from the control (*p*<0.05), similar letters indicate no statistically significant differences between groups.

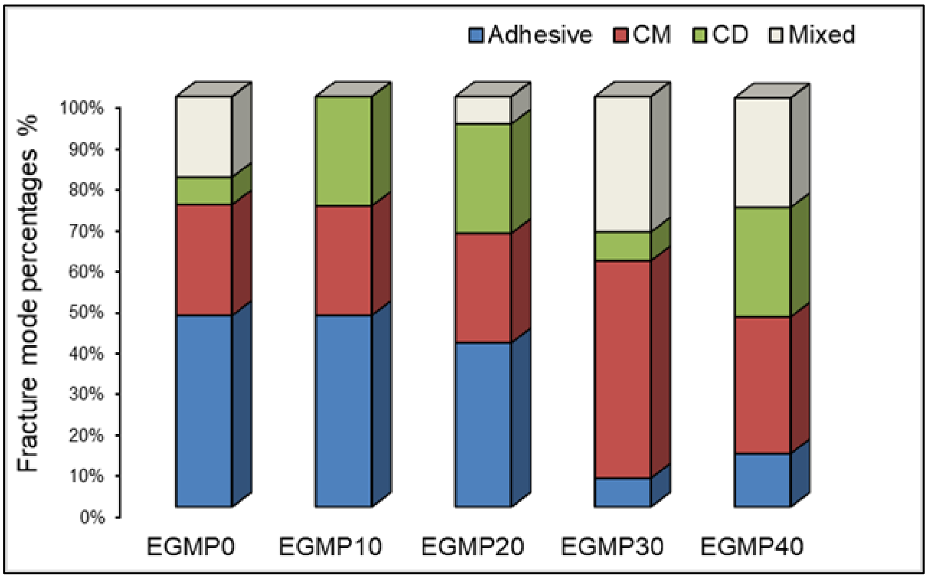


Figure 3-8 Mode of failures of the experimental groups with sound dentine after two weeks storage in SBF at 37°C.

3.3.7 SEM-EDX

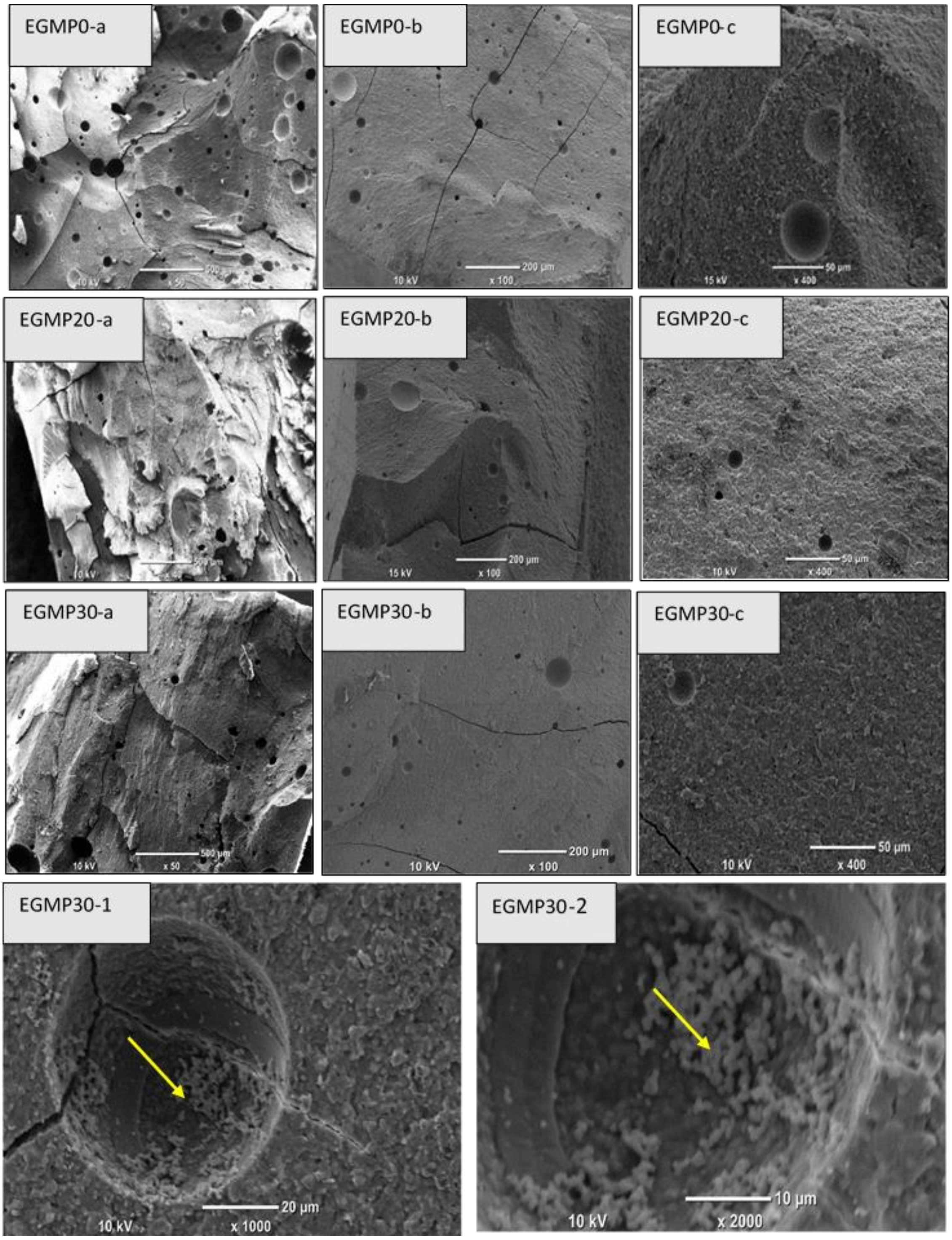


Figure 3-9 SEM of the fractured surfaces from the compression test after 4 weeks' immersion in SBF, A more integrated smooth and homogeneous surface are shown in EGMP20, and 30, as compared to that of Fuji II LC, EGMP30-1, 2 show the presence of a mineral kind of deposit inside the pores after four weeks ageing in SBF (yellow arrows).

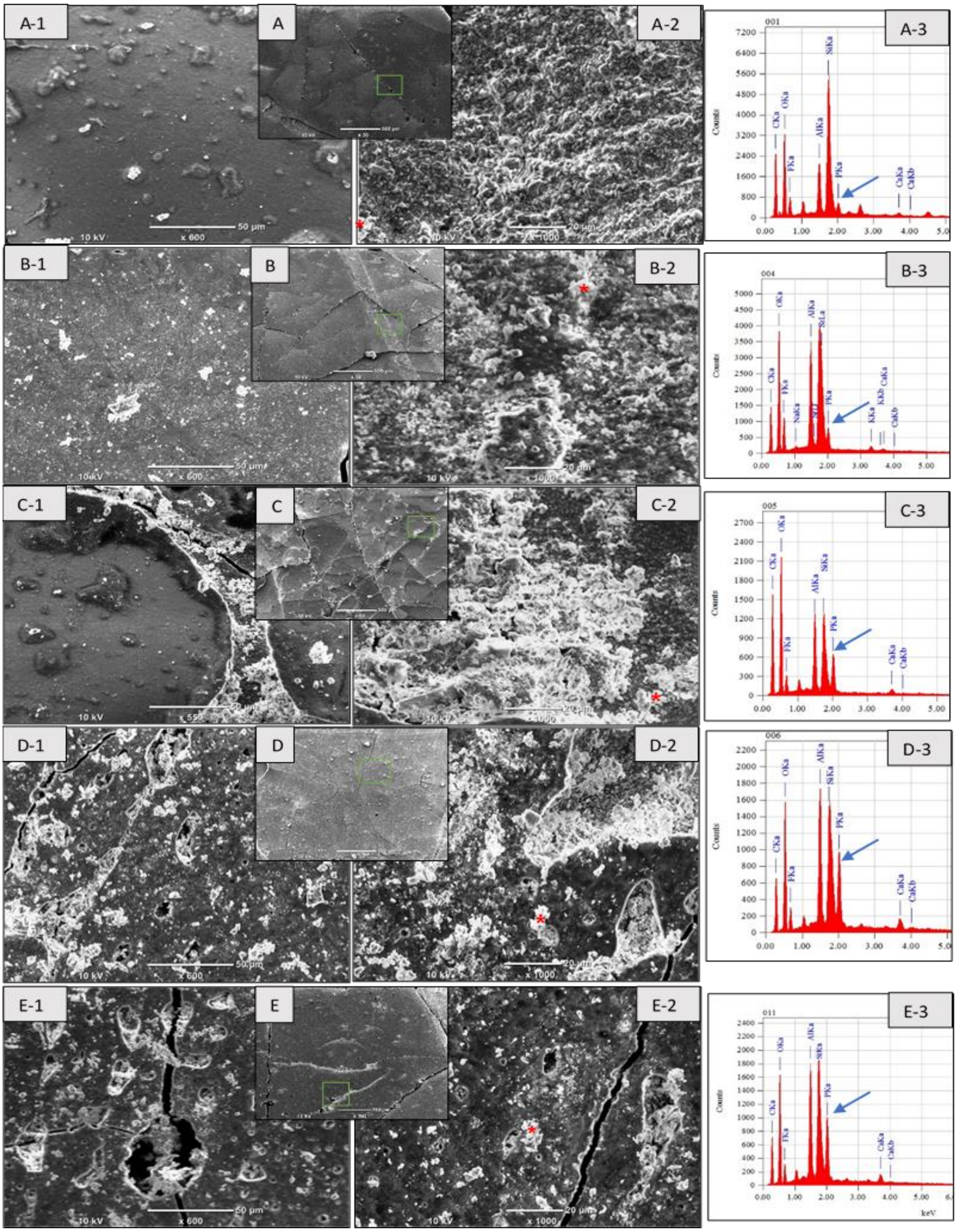


Figure 3-10 SEM-EDX of the fractured surfaces after BFS testing following 4 weeks' immersion in SBF at 37°C at x50, x600 and x1000 magnification power. Fig. A-E represent the surfaces of all groups at x50 as follow; A: EGMP0, B: EGMP10, C: EGMP20, D: EGMP30, and E: EGMP40. These surfaces were further analysed at higher magnifications (x600, x1000), Fig. A-E (1, 2). Cement matrix in the experiment groups in B-E (1, 2) were interspersed by shiny particles irregularly shaped that are not seen in the control group (Fig A-1, 2) that showed no visible changes in the surface morphology after ageing. The red asterisks represent the selected points to be further analysed by EDX (A-E, 3). The phosphate-based experimental cements show a similar chemical composition to the control group A-3 which contain elemental peaks of aluminium, silica, fluoride,

phosphorus, and calcium. But the peaks of phosphorus are recognised to be increased proportionally with increasing the content the monomer with in the cement, as shown by blue arrows

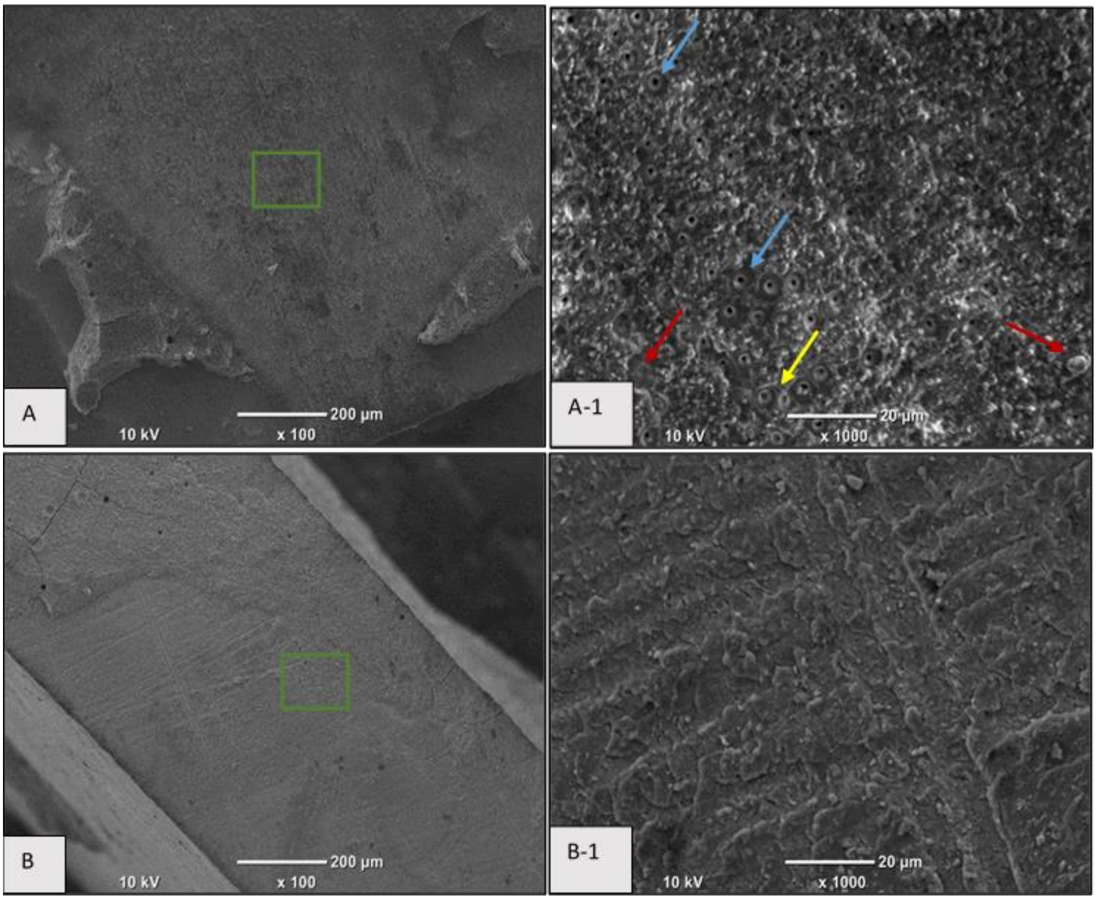


Figure 3-11 Scanning electron micrographs of the debonded interface between Fuji II LC (control) and sound dentine after two weeks storage in SBF at 37°C.They show mixed failure predominantly adhesive in A, while mostly cohesive in B. The selected areas (green box) showed large number of opened dentinal tubules on further magnification (blue arrows) (A-1), with the presence of partially and completely closed tubules (yellow arrow, red arrows, respectively). In Fig B-1, F2LC is completely covered the debonded area.

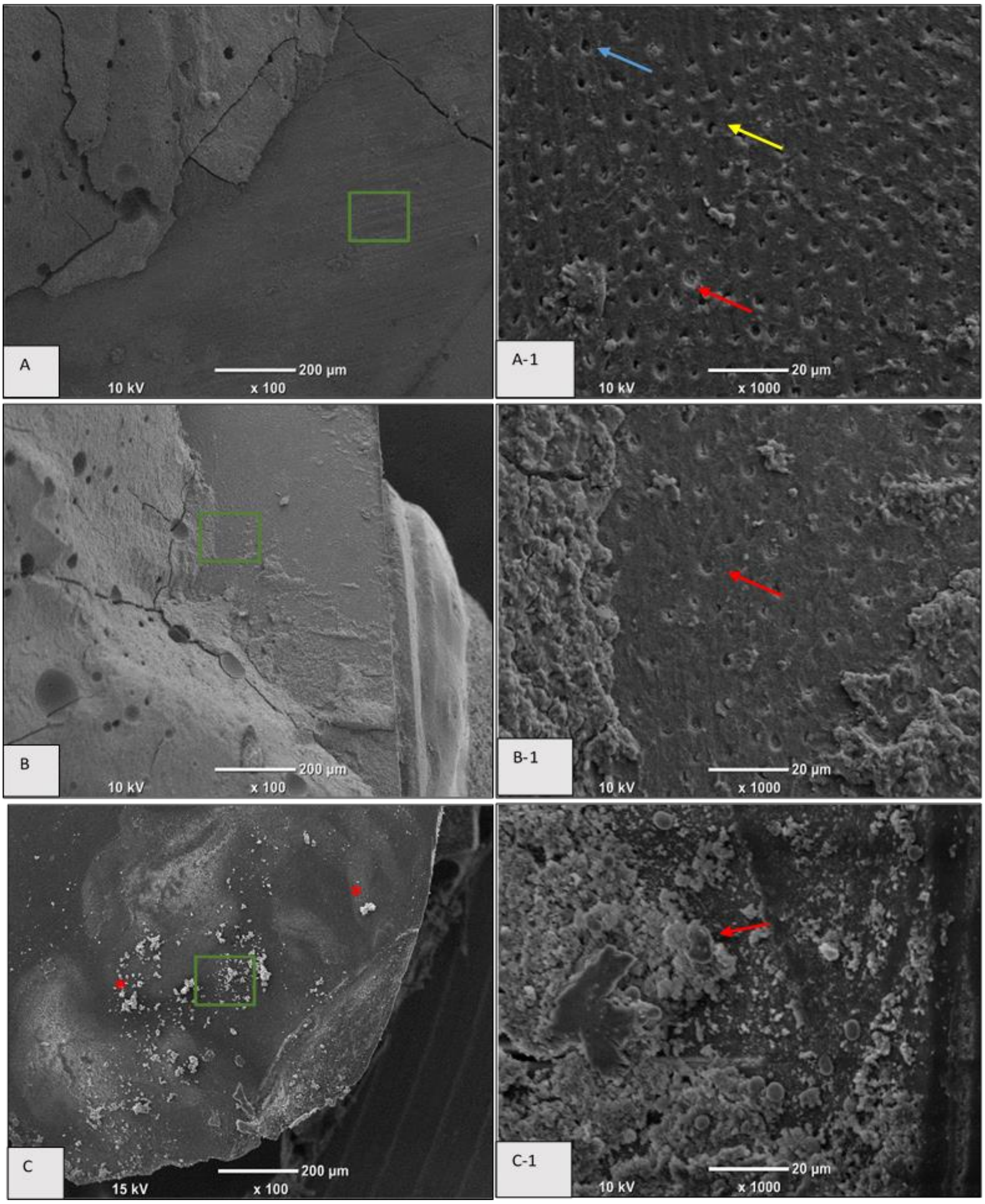


Figure 3-12 Scanning electron micrographs of the debonded interface between EGMP30 and sound dentine after two weeks storage in SBF at 37°C.They show mixed failure predominantly adhesive in A, mostly cohesive in B, and completely adhesive in C. When a selected area in a green box was further magnified to x1000 in (A-1), most of the dentinal tubules are recognised as partially or completely closed (yellow, red arrows, respectively) with some of the cement still attached to the surface with the presence of some opened tubules (blue arrows). In B-1, EGMP30 covered the debonded area with completely closed tubules (red arrow). In C there is an evidence of irregular shaped granular patches distributed (red asterisk) over the adhesively debonded cement-dentine interface indicated the mineral forming potential of the cement. On further magnification (C-1), part of the cement could be observed attached to the dentine surface with complete obliteration of the dentinal tubules.

3.4 Discussion

Ideal reparative material for repairing failed tooth-restoration complexes should be able to adhere to tooth structure and other different restorative substrates, with proper clinical handability and physical properties. A new repair material was developed in this study based on incorporating a phosphate-based monomer to a commercial RMGIC, which has an inherent property to adhere to tooth tissue and further modified to enable adhesion or interaction with other restorative materials. At the same time, can impart in remineralising a defective tooth tissue. Ethylene glycol methacrylate phosphate (EGMP) monomer EGMP is known as adhesion promoters (Yamauchi et al., 1981), it has an acidic functional phosphate moiety with a polymerisable monomer. The incorporation of a resinous water-soluble functional monomer in to the RMGIC system is expected to enhance their physical properties. This can be achieved by the formation of an interpenetrating polymer networks combining the acid-base cross-linking of the metal ion-polyacid and the cross-linking polymerisation of the incorporated monomer. Further interactions might be expected between the phosphate groups of the EGMP monomer and the metallic cations within the cement structure which reinforce the cement matrix itself, reduce its hydrolytic solubility, and maintain the biological requirements of RMGIC. This suggests that these cements are suited as restorative grade glass ionomer cements, and can be additionally used as a reparative material for failing TRCs.

3.4.1 Monomer compatibility and curing parameters

EGMP is miscible and compatible with the co-monomer HEMA (Suzuki et al., 2006) and the liquid phase of Fuji II LC, which was confirmed by the lack of any evidence of phase separation. At lower concentrations of EGMP, no discernible changes were observed in the working time of the cements, however a significant increase ($p<0.05$) resulted with the higher concentrations of 30-40% wt. EGMP in the formulation, (Table 3-2). It is interesting to note that at lower concentrations of EGMP the working time remained unaffected most likely due to the phosphoric acid groups being neutralised and integrated with the calcium ions. At higher concentration, there may a competing reaction of the phosphate and carboxyl groups to interact with the calcium ions, which prolongs the working time significantly, nevertheless it still meets the clinical requirements for water-based cements (ISO, 9917-2, 2010). Importantly, the inclusion of EGMP did not intervene with the setting time of the resultant cements. Since the acid-base reaction occurs due to the presence of the acidic polymer solution and alkaline glass powder irrespective of the presence of EGMP, which undergoes photo-polymerisation, the setting reaction remains unaffected.

3.4.2 Monomer effect on the mechanical properties

Compressive strength, modulus, microhardness and flexural properties are not only a reliable method to estimate the survival probabilities of dental restorations over time, but they can predict changes in the internal structural properties of the materials (Saskalauskaite et al., 2008) which directly influences the behaviour of the restoration when subjected to load. In this study, the addition of functional phosphate-based monomer to the RMGIC enhanced the cement's mechanical strength in most time intervals. Although, early compressive strength values of the EGMP-RMGIC are comparable to the control, all experimental formulations manifested a pronounced enhancement in CS post-ageing. The compressive modulus of the new cements also showed boosted values in comparison to the control. The increase in the values was proportional to the amount of the added monomer in the matrix after all time intervals. The microhardness that reflects changes in the density of the matrix was also enhanced in the new cement. However, the immediate effect was pronounced in the high EGMP-containing cements (EGMP30 and 40), but after cement maturation (180 days), all formulations (EGMP10-40) showed this trend. The addition of pendant free-radical polymerisable double bonds which have methacrylate and phosphate groups are expected to act as an organic space maintaining chemical which assists the movement of the acid groups (COOH) from the rigid polymer backbone providing a greater degree of freedom for the pendent carboxylate anions. This allows the complete conversion of the carboxylic acids to metal carboxylate complexes (salt-bridges formation) during setting reaction and reduces the number of unreacted carboxylic acids due to steric hindrance which in turn encourage the strength of the resultant cement (Culbertson, 2006). Furthermore, EGMP monomer with methacrylate residues can be readily polymerised with free-radical initiation producing a covalently linked matrix of homopolymers or even might be copolymerized with HEMA producing copolymer of EGMP-HEMA, both polymers can reinforce the matrix producing cements with improved properties. Additionally, the reactivity of the acidic functional phosphate group within this covalently linked network and its potential affinity to interact with the metallic cations liberated from the basic glass during the acid-base setting reaction; it thereby creates a network which is more resistant to applied forces. These interactions need to be further investigated. The polarity and the hydrophilicity of the phosphate moiety are supposed that there will be more water available for a more ionic bond formation which might be responsible for the increase in the compressive strength and the microhardness of the final cement matrix. This is consistent with the SEM findings in this study, Figure 3-9, that show a denser and more homogenous microstructure in the modified cements (EGMP20, 30) as compared to the control.

Biaxial flexural strength test (BFS) predicts the elastic qualities of the material which represents a fundamental property of a reparative material when strong and durable adhesion with other substrates is crucial. The interatomic or intermolecular forces of the material are responsible for the property of elasticity. The stronger the basic attraction forces, the greater the values of the BFS and elastic modulus and the more rigid or stiff material. The presence of a strong hydrophilic domains like the EGMP can inhibit the separation of the planes of atoms within matrix (Yamazaki et al., 2005). They act as a hydrophilic centre that support other hydrophilic domains which might increase the bipolar‑bipolar forces and affect the response of the material to BFS test (Xie et al., 2004). This might explain the two-fold increase in BFS values of the experimental cements as compared to the control.

The physicochemical interactions might also affect the strength of the cement matrix since there is a possibility of formation of H-bonds due to the presence of hydroxyl, phosphate, and carbonyl groups within the matrix. Without a doubt, stronger bonds between the organic and inorganic network of the set cement, lead to higher mechanical strength of final set cement. Consequently, they can occupy the empty spaces between the glass-ionomer particles and reinforcing the structure of the glass-ionomer cements.

3.4.3 Monomer effect on bonding to sound dentine

µTBS was performed to measure the efficacy of the incorporated monomer to promote cement adhesion strength to sound dentine. This approach is considered to be more predictable method over the conventional tensile and shear bond strength test since the mean and variance can be calculated per single tooth, hereby enabling statistical comparison on tooth level. Also the higher probability of adhesive than cohesive failures to occur and thus the measurement relies on the interfacial adhesive-tooth strength (Sano et al., 1994). However, in this study, cohesive failures within the cement or dentine was higher than adhesive failure among groups. Several limitations are reported for this technique such as the labour-intensive and technically demanding preparation especially in brittle materials, a difficulty in measuring low bond strengths (<5 MPa) and comparing data to previous studies (De Munck et al., 2012). Additionally, the in homogeneous load distribution as de-bonding stresses may locate within the dentine and material parts near the adhesive interface rather than to be at the interface (Söderholm et al., 2012). Accordingly, this method was used for optimisation purposes and would not applied further in measuring the interfacial adhesion strength of the new cement to different tooth/restorative substrates in the following chapters. All pre-test failures occurred during sectioning before confirming the total number of specimens per group (n=20). Therefore, these specimens were excluded from the analysis. As the premature failures are evenly distributed across all of the groups, their exclusion from the analysis would not have biased the bond test outcomes.

There was 21-27% enhancement in the bond strength of RMGIC to sound dentine when 20-40% wt. EGMP monomer incorporated to the cement ($p<0.05$) after 14 days storage in SBF at 37°C. This is attributed to the presence of acidic functional groups that confer self-etching property of the cement and augment the micro-mechanical interlocking to dentine with a possibility to interact with dentine leading to raised bond strength values. EGMP30 showed high µTBS to dentine (39.5 MPa) than the control (31.1 MPa) ($p=0.001$) and displayed higher percentages of cohesive and mixed fractures, Fig 3-8. Although, the value was comparable to EGMP20 and 40 groups (37.5, and 37.3%, respectively)($p=1.000$), but this percentage allows the availability of pendent phosphate groups for chelation potential with the hydroxyapatite of the tooth structure without compromising the acid base reaction of the original cement. This results was supported by SEM finding that showed more partial and complete closure of the dentinal tubules at the debonded dentine interface (Figure 3-12, A-1), in comparison to the control (Figure 3-11, A-1). Additionally, there is dispersion of irregularly shaped particles at the dentine surface following adhesive failure from EGMP30 (Figure 3-12, C and C-1) indicate an evidence of mineral deposit over the surface post-ageing which might confirm the mineralisation potential of the new cement.

3.4.4 Effect of different monomer proportions (10-40% by weight)

Different proportions of phosphate-based monomer range from10-40% by weight were applied, as a step forward for developing a new reparative material with enhanced mechanical and bonding strength. The addition of 10% wt. EGMP produced cement with comparable properties to the control (CS, MH and µTBS) (p>0.05). In contrast, cement containing higher monomer concentrations 20-40 % exhibited enhanced physical and bonding properties in comparison to the control cement with no significant differences among them (p>0.05). The higher amount of monomer provides more double bonds in cement matrix for effective covalent crosslinking with more expected interactions via functional groups with the metallic cations and tooth surface to form complexes that can reinforce the matrix and enhance bonding to tooth surfaces. Additionally, the higher acidic functional groups, the greater potential to dissociate into ionic form and generate protons that makes the medium acidic (Salz et al., 2006). On the other hand, higher concentrations might interfere with the acid-base reaction in RMGICs, causing the glass particles to play a role as a filler instead of as a reactive species. Accordingly, 30 wt.% EGMP was selected as the optimum amount of monomer to be included in the new cement (pRMGIC) as it showed enhanced mechanical strength that are maintained post-ageing and improved bond strength to sound dentine with less chances of adhesive failures. Nevertheless, increasing the monomer content beyond this limit does not show further improvement in the properties.

3.4.5 Effect of ageing and monomer reactivity

Ageing apparently improved the mechanical strength of all RMGICs (experiment and control) during the first two weeks as compared to their early values. However, no further enhancement in properties have shown up to four weeks, Tables 3-(3-

6). The results are consistent with early studies (Tosaki, 1994; Xie et al., 2004) that supposed the initial strength of the RMGICs relied on photo-initiated covalent crosslinking, with time, there will be more constant ionic salt-bridges formed which offer higher strength values. On prolonged ageing, EGMP-contained cement maintained their strength while the control group lost the gained strength and being comparable to its early value. This agrees with a study conducted by Uno et al., (1996) that reported ageing of RMGIC up to 6 months has little adverse effect on the mechanical properties. The reactivity and hydrophilicity of the phosphate group within the matrix might allow a mature ionic setting and build-up of salt-bridges that endure dissolution and resist the plasticising effect of water with time.

When a polymeric acid with polymerisable monomers mixed with acid-soluble glass, the material sets via a light activated process as well as acid/base reaction. The calcium ions that leached out from the glass due to acid hydrolysis of the glass are cross-linked by PAA molecules forming calcium polyacrylate network (Nicholson, 1998). The pendent phosphate moieties within the cement matrix might interact with the residual calcium ions forming Ca/P complexes that precipitated in the form of minerals within the matrix, which reinforce the cement against stresses and ageing effect. These speculations are supported by previous study that confirms the inductive ability of these negatively charged functional groups for apatite precipitation within the body environment (Stancu et al., 2004). The microstructural observation under SEM shows precipitation of a mineral kind of deposit inside matrix pores after four weeks storage in SBF (Fig 3-9). Under the same circumstances, the IR spectra (Fig 3-3b) also supported this evidence, as the modified formulations displayed a strong, absorbent band at 966 $^{-1}$cm assigned to v1 stretching vibrations of the phosphate PO_4^{3-} in the apatite (Rey et al., 1990; Gadaleta et al., 1996). Further surface examination of EGMP-contained cement using SEM shows morphological variations at the surface, Fig 3-10, characterised by the dispersion of particles with varying size and shape showed evidence of mineral deposit at the surfaces that were distinctly different from the particles of the cement. However, EDX microanalyses did not confirm these deposits as a true apatitic structure, Fig. 3-10, A-E (3). It shows abundant quantities of P within the modified cement that increased proportionally with increasing the content of the added monomer. The Ca peaks are not detected as high as the P peaks in all RMGICs. This might be attributed to the reduction in the oxidation state of these calcium atoms by charge transfer process as calcium atoms are crosslinked by the organic functional groups of the polyacrylic acid and the phosphate-based monomer which expected to change their electronic structure (Gerth et al., 2006). The chelation of the calcium ions from the HAp of the dental tissue with acid functional groups based on the same principle. Accordingly, it will be a powerful method to quantify the Ca peak components when the cement chemically interacted with the inorganic teeth components.

Series of studies (Nuttelman et al., 2006; Chirila et al., 2007; Kemal et al., 2011) are concerned with the functionalisation of the polymer by incorporating phosphate groups to produce surfaces capable of initiating a cascade of events that lead to calcium phosphate (CaP) mineral nucleation and subsequent biomineralisation. In this study, the incorporation of a methacrylate monomer carrying phosphate side groups to the RMGIC allow designating an improved cement with an established mineralisation potential, which needs to be further investigated.

3.4.6 Water uptake, solubility, and diffusion coefficient

Water uptake measurements carried out following ISO 4049 standard for resin-based materials. However, specimens were not desiccated prior to commencing the sorption cycles and immersed immediately in the distilled water at 37°C. Desiccation for the RMGICs might remove either or both the ‘loosely bound’ water which is essential for the progression of the acid-base reaction, and the ‘tightly bound’ water which forms the structural part of the cements. In agreement with literature (Zankuli et al., 2014), all cements are continuously gained water until reaching the equilibrium during the first week. The polarity of the functional groups attract more water in the EGMP-contained cement as the hydration was proportional to the amount of the added monomer and it was significantly higher than the control. The hydrophilic potential of this functional monomer was previously confirmed by Kemal et al., 2011, who reported a proportional correlation between the equilibrium water content of a HEMA-co-EGMP containing hydrogel and the content of EGMP copolymer. Nevertheless, the absorbed water may contribute to the ongoing acid-base reaction and the formation of stable ionic interaction over time, additionally, these functional moieties can chelate with the metallic cations forming complexes which produce dense matrix more homogenous matrix, as evidenced by SEM images (Fig 3-9).

Theoretically, the rate of water uptake and loss depends on the density of the polymeric network, the concentration of polar sites available for hydrogen bonding (Unemori et al., 2003), polymer polarity (Bellenger and Verdu, 1989) and polar interactions within the matrix. The highly crosslinked polymers showed better resistant to degradative reactions (Ferracane, 2006) due to limited space and pathways available for solvent molecules to diffuse in to and out of the polymeric structure. The polarity and interactions of the functional phosphate groups produced dense matrix that imposes a certain resistance to water intrusion to the matrix, decreasing the rate of water diffusion, and significantly reduce the solubility. They are both strongly correlated with the proportion of the EGMP monomer within the matrix, the greater amount of monomer, the slower coefficient of diffusion and less solubility. In agreement to Yap and Lee, (1997), water uptake in all groups follow Fick's low of diffusion. Slopes of the initial part of water uptake showed that water absorbed more rapidly in commercial RMGIC than the experimental cements, adding more monomer associated with a lower diffusion rate. The reinforced cement matrix constringes the elution of the unreacted components and degradative residuals reducing the solubility of the EGMP-contained cement in comparison to the control. Furthermore, mass loss due to dissolution might be compensated by water retention, which in turn be converted from loosely bound to tightly bound within the matrix over time.

3.4.7 Fluoride release

The enhanced ionic interactions within the EGMP-contained cement which reflected by improved mechanical strength produce tightly bonded matrix might be responsible for reducing the amount of fluoride release from the matrix during the first 48 h of the cement life, as the F^- ion diffuses through the microporosities at this stage (Verbeeck et al., 1993; Thanjal et al., 2010). Even so, with time, the pattern of fluoride ion elution is apparently consistent for all tested formulations, Fig. 3-6.

3.5 Limitations and future work

The current study highlighted the possibility of incorporating different proportions (10-40% wt.) of a methacrylated acid-functional monomer (EGMP) to a RMGIC to expand its clinical applications and specifically to be used as a reparative material for failed TRCs. The new material showed enhanced mechanical strength up to 6 months when stored in SBF at 37°C. However, further studies in this area is needed to assess the structural changes in the modified cement and the setting kinetics. Furthermore, it is important to appreciate the necessity for optimum P/L mixing ratio when delivering a new cement, taking in to account the possibility of a change in the early viscosity of the mix after liquid modification. Thus future investigation using different proportional P/L ratios needs to be conducted to establish the effect on the resultant properties.

The early hydration of the new cement did not exhibit appreciable adverse effects on the mechanical strength up to six months’ storage, which this may be attributed to the ongoing acid-base reaction and the formation of stable ionic interactions with time which imposes a certain resistance to water intrusion, decreasing the rate of water diffusion and significantly reducing the solubility and the early fluoride release. However, the long-term effects in terms of water uptake, expansion and the possible hydrolytic degradation need to be investigated for future work.

Another factor must be considered is the possibility of the interference between the acidic monomer and the photoinitiator systems that contain tertiary amines. This may affect the degree of monomer conversion and compromise the properties with a possibility of leaching out the residual monomers (HEMA) which compromise the biocompatibility. However, there is limited information regarding the composition of the photoinitiators that have been used in the RMGICs, since they already have polyacids in their main composition. Nevertheless, further research is necessary to assess the effect of the acidic monomer on the rate of acid-base reaction and the polymerisation reaction with and without light curing.

Finally, the precipitation of mineral kind of deposits at the surface of the modified cements and inside matrix pores after four weeks storage in SBF were recognised. This is further supported via the IR spectra (Fig 3-3b) as a strong, absorbent band at 966 $^{-1}$cm assigned to v1 stretching vibrations of the phosphate PO_4^{3-} in the apatite, which suggested the potential mineralisation ability of the new material.

However, this FTIR peak might be referred to the phosphate groups in the EGMP monomer which increased in its intensity with increasing the amount of added monomer. Furthermore, the EDX microanalyses did not confirm that these deposits are true apatite, as it showed abundant quantities of P within the modified cement that increased proportionally with increasing the content of the added monomer. Accordingly, it is suggested, that further analysis of these mineral deposits using other tools like Raman spectroscopy, XPS and XRD, be carried out.

3.6 Conclusion

Within the limitation of this study, this chapter investigates the possibility of incorporating a monomer with phosphate moieties to the RMGIC to be used as a reparative material for failed tooth-restoration complexes. The mechanical properties (CS & CM, MH and BFS) of the new cements showed that these materials are promising additives for the glass-ionomer systems. The optimal proportion for EGMP is 30% by weight that recorded higher strength values up to 6 months' storage with higher adhesion strength to sound dentine after two weeks storage in SBF. SEM analysis supports strength data associated with the formulations. Minerals precipitation within the EGMP-contained cements showed promise for a repair material and/or workable cements for several dental applications.

Chapter Four

4 Interfacial adhesion strength of pRMGIC to sound and demineralised enamel and dentine surfaces

4.1 Introduction

Minimally invasive dentistry focuses on prevention, remineralisation, and minimal intervention with the primary aim of tissue preservation. Thus, of particular relevance is repairing failing tooth/restoration complexes (TRCs). This encompasses the preservation of the maximum quantity of repairable dental tissues and utilising reparative therapeutic materials that seal and rejuvenate enamel and dentine margins (Green et al. 2015). Despite the considerable differences in properties of the commercial materials used for repair. The key approach remains to utilise a material with close affinity, both physically and chemically to sound and carious tooth margins to minimise the risk of further tooth damage and prevent the ingress of bacteria. At the same time, the ability to bond to variety of restorative materials, with appropriate physio-mechanical properties.

Alterations in the morphological, chemical and physical characteristics of the defective substrate from the healthy tooth tissue compromise the bonding efficiency when adhesive resin restorations are used (William et al., 2006; Erhardt et al., 2008). The poor quality of the hybrid layer at the mineral-depleted surfaces may jeopardise the longevity of these restorations due to the hydrolytic degradation over time (Hashimoto et al., 2000). Clinically, the preservation of repairable demineralised enamel and carious affected dentine surfaces are beneficial for implementing atraumatic restorative treatment (ART), since they serve as a suitable substrate for dentine adhesion (Yoshiyama et al., 2003) and physiologic remineralisation (Watson et al., 2014). This can be achieved using a therapeutic adhesive material with advanced physical properties and ionic supply that able to bond chemically with the damaged enamel or dentine tissue and protect the bonded interface from further damage and increase the longevity of the restoration.

Conventional glass ionomer cement CGIC is considered as a therapeutic alternative to the adhesives/composite restorations. This is attributed to the chemical bonding ability to dental substrates, with a sustained fluoride release (Wilson, 1989; Smith, 1998). They are capable of supporting repair and remineralisation of dentine left after decay and cavity preparation due to the ion exchange concept between hard tissues whether healthy or diseased and the material (Calvo et al. 2014; Toledano et al., 2016). But they do not, as yet, have the wear resistance and the mechanical properties to make them suitable as long-term reparative material.

The addition of resin chemistry to GICs enhanced flexural and tensile strengths, elastic modulus, and wear resistance (Modena et al., 2009). Additionally, RMGIC exhibited higher bonding strength to enamel and dentine than CGIC, since the adhesion to tooth surfaces based on both chemical interactions and micromechanical interlocking of the polymer and polyacrylic acid-conditioned tooth surface (Yiu et al., 2004b). On the other hand, shrinkage on curing and susceptibility to degradation over time could affect the marginal integrity of RMGIC restorations with the dental substrate or other restorative materials (Attin et al., 1995). However, the dual setting mechanisms of the RMGIC encourages a relief from the polymerisation stresses leading to a better tooth seal, despite their lower bond strength compared to the resin adhesives (Mitra et al., 2009).

The concept in this study is to develop an adhesive reparative material based on aqueous-systems by incorporating a polymerisable functional monomer with a reactive phosphate group. The acidic group is expected to promote the self-etching behaviour of the cement and provide an ionic supply which provokes strong chemical interactions with the remaining tooth structure (Münchow et al., 2015), and deposit minerals that strengthen the tooth/restoration interfaces against degradation over time. Thus, by etching/chemisorbing concept (Fu et al.,2005; Suzuki et al.,2006), the new cement is expected to produce a strong and durable bond to tooth surfaces, even higher as compared to the carboxylic acid groups (Yoshida et al., 2004).

Ethylene glycol methacrylate phosphate (EGMP) was utilised for this modification. This monomer contains a reactive phosphate group, which was previously reported as an adhesion promoter in the self-etching bonding systems (Münchow et al., 2015). Additionally, there is a mineralisation potential had previously proved in hydrogels using this functional monomer (Kemal et al., 2011). Accordingly, this monomer was incorporated in different proportions (10-40% wt.) to the liquid phase of the commercial RMGIC (Fuji II LC). This was previously explained in chapter 3. 30% by weight of EGMP was selected for the present study, due to the higher achieved mechanical strength and bonding ability to sound dentine (µTBS), after two weeks' storage in simulated body fluid (SBF). Accordingly, this innovative reinforced cement with controlled sorption/solubility behaviour was investigated as an adhesive therapeutic reparative material which expected to seal and repair tooth margins by producing high and long-lasting adhesion to healthy and diseased tooth substrates.

The success of reparative materials depends upon enhanced adhesion with the dentine/enamel surface to resist breakdown of the interface on loading whilst exhibiting effective flow and integration with the margin of the existing restorative material. The rationale behind this method is that the stronger the adhesion between tooth and biomaterial, the better it will resist stresses generated during oral function. Bond strength measurement and fracture analysis are gross assessing tools for evaluating the efficacy of bonding restorations to dental substrates. Of the various tests, shear bond strength is less technique sensitive technique and yields reproducible results to enable statistical analysis. Whilst bond strengths cannot be considered a material property (Van Noort et al., 1989). It can reveal the potential clinical performance of materials and allow comparison with the currently available products that used for the same clinical purposes. Thus the repair quality was assessed by evaluating the shear bond strength (SBS) of pRMGIC to four different tooth surfaces; enamel, demineralised enamel, dentine, and carious affected dentine, after 24 h and three months storage in SF at 37°C. Then, the results were compared to three commercial restorative grade materials that are used in ART as reparative materials; RMGIC (Fuji II LC (F2LC)) as a control group, CGIC (Fuji IX (F9)), and composite resin (Filtek™ Supreme (FS)) with Scotchbond Universal adhesive bond (SU).

Null Hypotheses

1. There are no statistically significant differences in SBS values of the experimental cement (pRMGIC) from the control (F2LC), F9 and FS to sound versus demineralised enamel and dentine substrates after 24 h and three months' storage.
2. There is no statistically significant difference between the early and delayed adhesion strength values for each material per substrate.

4.2 Materials and Methods

Table 4-1 List of materials used in the study

Materials	**Manufacturers**	**Code**	**Material composition**
Fuji II LC (RMGIC)	GC Corp., Tokyo, Japan	F2LC	Powder: Fluoro-alumino-silicate glass Liquid: polyacrylic acid, 2-hydroxyethyl methacrylate (HEMA) dimethacrylate, camphorquinone, water.
Fuji IX™ GP (CGIC)	GC Corp., Tokyo, Japan	F9	Powder: fluoro-alumino-silicate glass, Liquid: polyacrylic acid, Polybasic carboxylic acid.
Filtek™ Supreme XTE *(universal restorative composite), (Shade A2B)*	3M™ ESPE	FS	The resin-based matrix contains bis-GMA, UDMA, TEGDMA, and bis-EMA resins. The fillers are a combination of a non-agglomerated/non-aggregated 20 nm silica filler, a non-agglomerated/ non-aggregated 4-11nm zirconia filler, and an aggregated zirconia/silica cluster filler (comprised of 20nm silica and 4 to11 nm zirconia particles, with an average cluster particle size of 0.6 to 10 microns. Filler loading is about 78.5% by wt (63.3% by volume)
Scotchbond™ *Universal, (self-etch Adhesive)*	3M, ESPE, USA	SU	MDP phosphate monomer, Dimethacrylate resins, HEMA, Vitrebond™ Copolymer, filler, ethanol, water initiators, silane
Dentin conditioner	GC Corp.,Tokyo, Japan		90% distilled water and 10% polyacrylic acid

4.2.1 Specimen preparation

256 human molars (192 sound teeth and 64 with occlusal caries). Teeth were collected using an ethics protocol reviewed and approved by NHS health research authority (16/SW/0220). The teeth were cleaned using an ultrasonic cleaner and stored in deionised water in the fridge at 4°C. They were used within six months following the extraction. The roots of all teeth were sectioned using a hard tissue microtome (Isomet 1000, Buehler, Lake Bluff, IL, USA) equipped with a slow-speed, water-cooled diamond wafering saw (330-CA/RS-70300; Struers). Sound teeth were randomly allocated into three main groups (n=64 per group) based on the different bonding surfaces; sound enamel, demineralised enamel, and sound dentine. The fourth group was the caries affected dentine substrate from natural carious teeth (n=64). Each tooth was sectioned mesiodistally into two halves using a diamond wafering blade (XL 12205, Benetec Ltd., London, UK) obtaining the buccal half only from each tooth. Surface integrity of each specimen was inspected using microscopy at x40 magnification. Half of the prepared specimens (n=128) were placed face down in a custom-made silicon sample former in a way that the enamel surface was exposed to the acrylic resin block. While the rest 128 specimens were placed face up to expose the dentine (sound and carious) surfaces. All specimens were polished by a silicon waterproof abrasive paper of 120-grit to expose and flatten the selected surfaces.

For the demineralised enamel group, artificial enamel white spot lesions (WSLs) were induced on 64 randomly selected enamel surfaces. This was done following the previously reported bi-layer demineralisation protocol of 8% methylcellulose gel buffered with a lactic acid layer (0.1M, pH 4.6) for 21 days at 37°C (Ten Cate and Duijsters, 1982; White et al., 1988). To create artificial enamel lesions, the outer enamel layer was removed using a water-cooled rotating polishing machine (Meta-Serv 3000, Buehler, USA) with a sequential polishing protocol; 500-grit silica carbide disk (Versocit, Struers A/S, Copenhagen, Denmark) for 5 s, 1200-grit for 10 s, 2000-grit for 30 s and 4000-grit for 2 min. Ultrasonication was carried out for 1 min after each step and 4 min after the 4000-grit to remove the smear layer. All samples were kept in deionised water at 4°C before further treatment. The polished surface was protected by water-resistant stick tape to leave a window approximately 3.5 mm wide and 3.5 mm long. The MC gel was prepared by adding deionised water (100°C) into methylcellulose powder (Sigma-Aldrich, USA) and magnetically stirred up to 24 h at ambient temperature. All enamel specimens (n=64) were placed on the bottom of the 2 L glass beaker, followed by pouring 800 mL MC gel to fully cover the samples and putting a filter paper on the top of the gel. Then samples were kept in the fridge at 4°C overnight for the gel to set before further step. The pH of lactic acid (AnalaR, UK) was adjusted to 4.6 using 1 M NaOH. 800 mL lactic acid was poured over the filter paper. Finally, the beaker was sealed tightly with cling film and placed in the incubator at 37 °C for three weeks. The acid demineralising solution was changed every week. After demineralisation, the acidic gel was removed from sample surfaces which were then rinsed with distilled water for 1 min. All samples were stored in deionised water at 4 °C for the next step. The depth of the created artificial WSLs on enamel surfaces ranged between 100-150 µm depth (n=3) measured using optical coherence tomography (OCT) (Noran Instruments, Middleton, WI, USA), Figure (4-1).

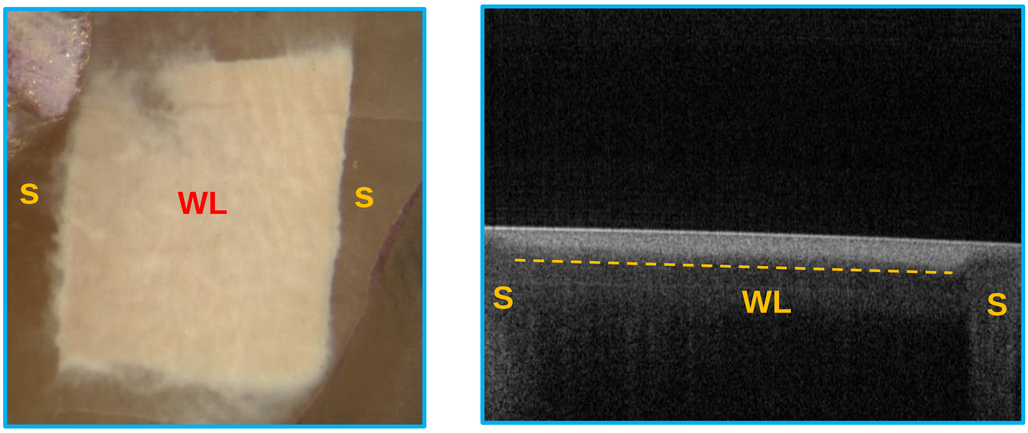
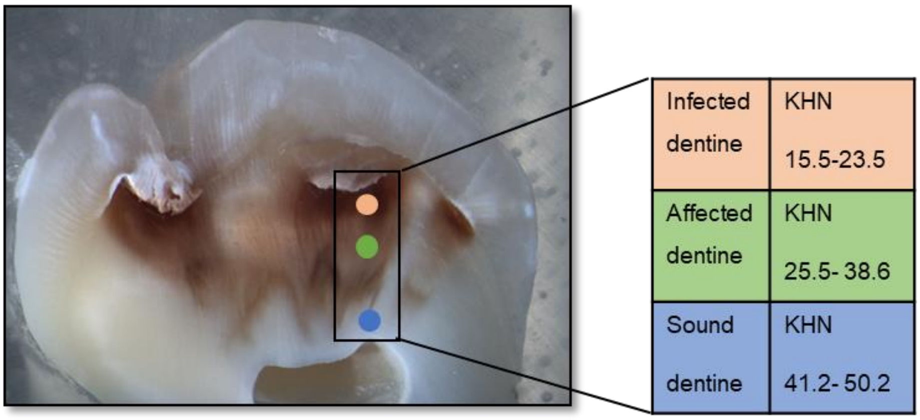


Figure 4-1 A photograph of one of the samples used in this study shows the WSL (WL) in the centre of the enamel surface surrounded by an intact enamel area (S). The OCT image on the right side presents the cross-sectional view of the sample. The lesion in the centre exhibits an increased signal intensity compared to the surrounding sound enamel.

For the natural carious dentine group, sixty-four carious molars were sectioned mesiodistally through the carious lesion. The selected teeth exhibited a deep carious lesion that extended halfway through the dentine without pulp exposure. To identify and select the caries affected dentine area (CAD), specimens were inspected visually to detect the colour change. Surface hardness also implemented using a dental explorer to identify the consistency and moisture of the carious layers. All samples were embedded in acrylic blocks, as previously explained, and polished to expose a flat tissue surface. Further characterisation using Knoop microhardness was carried out, to discriminate the two main carious layers (infected and affected caries dentine). Six measurements were taken from the dentine-enamel junction towards the pulp using Struers Duramin (Struers Ltd., Denmark) after applying a 10 gf load for 15 s. Three zones were identified; caries infected dentine, caries affected dentine (CAD), and sound dentine with the KHN values range 15.5-23.5, 25.5-38.6, 41.2-50.2, respectively, Fig 4-2. The characterisation results for identifying different zones of carious dentine are consistent with what previously reported summarised in Table (4-2).



Infected dentine	KHN 15.5-23.5
Affected dentine	KHN 25.5- 38.6
Sound dentine	KHN 41.2- 50.2

Figure 4-2 A photograph of half a sample with an example of the selected caries lesion displaying caries zones with the range of the recorded range of microharness values (KHN) for each zone

Table 4-2 Summary of the methods used to characterise the carious dentine zones

Techniques	Caries infected dentine	Caries affected dentine (CAD)	Sound dentine	References
Visual inspection	Dark brown	Paler-light brown	Yellowish/ white	Fusayama et al., 1966; Banerjee, 1999
Relative tissue hardness	Wet/ soft	sticky/ scratchy	Hard	Kidd et al., 1993; Banerjee et al., 1999
Microhardness	KHN > 25	25< KHN>40	KHN<40	(Ogawa et al., 1983; Banerjee et al., 2010)

All specimens were rinsed with deionised water and cleaned in an ultrasonic bath for 3 min to remove any surface debris. All bonding substrates were finished with wet 600-grit silicon carbide paper (Microcut; Buehler) for 60 s, to produce a clinically relevant smear layer, followed by the application of 10% polyacrylic acid (Dentin Conditioner; GC Corp) for 20 s to remove the created smear layer. The conditioner was washed with air/water spray for 15 s, dried with a gentle stream of dry compressed air for 15 s. After preparation of the respective surfaces, each bonding substrate (sound enamel, demineralised enamel, sound dentine, and carious dentine) (n=64 per group) was assigned randomly into four experimental groups (n=16) according to the type of the applied reparative material. Three commercial restorative materials were used in this study (F2LC, F9, and FS) to test the bonding effectiveness of the experimental pRMGIC group, Table (4-1).

4.2.2 Shear bond strength test

The modified liquid was hand-mixed with the Fuji II LC glass powder using a P/L ratio 3.2/1.0, at ambient temperature (23±2ºC) and humidity (35±5%) to form a uniform mix. While commercial groups (Fuji II LC, Fuji IX™ GP and Filtek™ Supreme) were dispensed according to the manufacturers' instructions. The materials then transferred into cylindrical silicon moulds (3mm diameter x 4mm height) (Tygon tubing, Saint-Gobain, USA) to ensure a reproducible amount of material over the conditioned surface, Figure (4-3). Due to the difficulty in discriminating the different zones of carious dentine, the mould is placed in such a way that the reparative materials covers at least 70% of the caries affected dentine zone in CAD substrates. Light cured materials were photo-polymerised according to the manufacturers' instructions using a light curing device (Elipar™ DeepCure-S LED, 3M USA) with a light intensity of 1470 mW/cm^2. RMGICs (PRMGIC & F2LC) received 40 s light curing on the top of specimen's surface. Scotchbond™ Universal (SU) self-etching bonding agent that applied before the composite resin was light-cured for 20 s, followed by the application of two increments of FS with 20 s light curing per increment. Samples were stored for an hour in an incubator at 37°C to allow setting of the applied materials. Shear bond strength (SBS) test was carried out at 24 h and the following three months' storage in simulated body fluid (SBF) at 37°C, (n=8 per subgroup). The solution was prepared following Kokubo and Takadamas' formula to examine the reactivity of the added phosphate based monomer for apatite induction in a simulated body fluid with ion concentrations nearly equal to those of human blood plasma (Kokubo and Takadama, 2006). Solutions were replaced on a weekly basis.

After each storage period, pre-test failures (PTFs) were recorded. The specimens were attached to the shear testing device using the universal testing machine (Instron® Model 5569, USA), positioned as close as possible to the tooth/material interface, Fig (4-3, E) A shear force was applied at a crosshead speed of 0.5 mm/min until deboning occurred. The shear bond strength (t) was calculated in MPa using the equation $t=F/(\pi R^2)$ where F was the applied load at failure and R was the radius of the material cylinder.

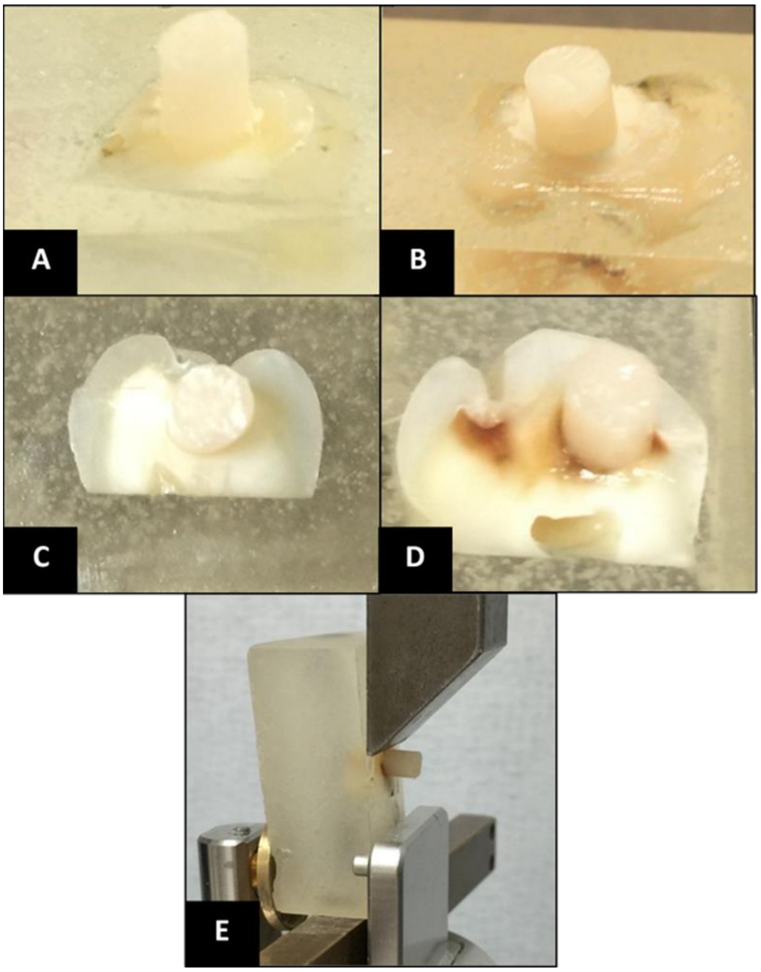


Figure 4-3 Images show the test materials bonded to four different substrates; A: sound enamel, B: demineralised enamel, C: sound dentine, D: CAD, E: a specimen attached to the testing apparatus.

4.2.3 Mode of failures and micro-morphological evaluation of the material/tooth interface

The mode of failures was examined visually using a stereomicroscope (WILD M32; Heerbrugg, Switzerland) at x40. Fractures were classified as; (A) adhesive failure at the reparative material/ tooth substrate interface; (B) cohesive failure within the reparative material; (C) cohesive failure within the tooth substrate; (D) mixed failures that are classified as a combination of partial interfacial adhesive failure and cohesive failures in either the applied material or the dental substrate, (Fig 4-4). Representative debonded aged specimens (n=4 per group) that failed in mixed or adhesive modes were selected to analyse the ultramorphology of the debonded surfaces using SEM. For the SEM, the samples were dried overnight, mounted on aluminium stubs with carbon cement, sputter-coated with gold at 45 mA currents for 2 min and viewed under a scanning electron microscope (JCM-6000 PLUS, NeoScope - Benchtop SEM, USA) at an accelerating voltage of 10 kV and three magnification power (x30, x500 and x2500, and x10000) and working distance (1mm, 50µm, 10µm and 2µm respectively).

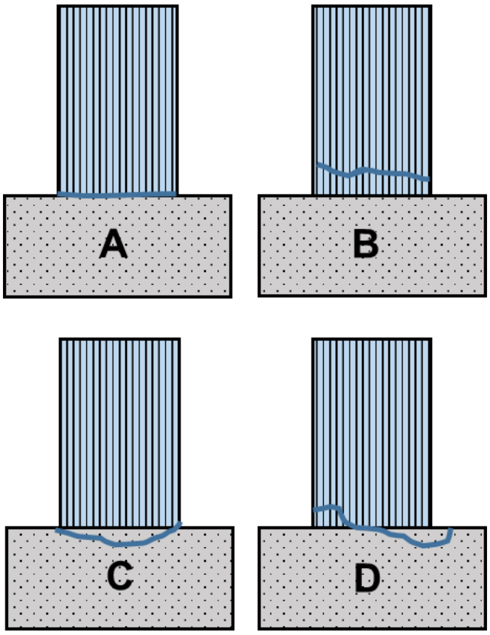


Figure 4-4 Classification of failure modes: (A) adhesive failure at the reparative material/ tooth substrate interface; (B) cohesive failure within the reparative material; (C) cohesive failure within the tooth substrate; (D) mixed failures that are classified as a combination of partial interfacial adhesive failure and cohesive failures in either the applied material or the dental substrate.

4.2.4 Statistical analysis

Descriptive statistics were used to summarise the bond strength characteristics of the selected reparative materials to different tooth surfaces after two-time intervals (24 h & 3 months). Percentages were used to present the failure modes. Data were tested for normality using Q-Q plots and Shapiro-Wilk tests, and they were analysed using parametric analysis as the data followed a normal distribution. Multivariate Analysis of Variance (MANOVA) and Bonferroni HSD post hoc tests were employed to calculate the significant factors (p = 0.05) in SBS values among tested groups within all substrates at each time interval. After analysing the normality of data distribution Independent t-test (for normally distributed values) was used to determine the effects of storage time on the bond strength of each tested material per each substrate. All analyses were conducted using SPSS statistical package (version 24; SPSS Inc., IBM, Chicago, IL, USA).

4.3 Results

4.3.1 SBS to sound and demineralised enamel substrates

The shear bond strength of pRMGIC, Fuji II LC (control), Fuji XI GP (F9) and composite (FS) to both sound and demineralised enamel at 24 h and 3 months ageing in SBF is shown in Table 4-3. The pre-test failures (PTFs) are included in the table and were given a “0” value for statistical analysis according to ISO guidelines (ISO/TS 11405. Dentistry- Testing the adhesion to tooth structure, 3rd edition, 2015).

There was a profound enhancement in the adhesive strength of the pRMGIC to sound enamel (24.3±2 MPa) as compared to the control (18.7±2 MPa) and F9 (6.8±3) (p<0.001) after 24 h, and all applied materials after three months’ storage in SBF at 37°C (p<0.001).

Ageing enhances the bond strength of the pRMGIC to enamel (26.0±3 MPa), however, it was statistically not significant (p=0.233), while the control remained unchanged post-ageing (p=0.252). In contrast, the bond strength of CGIC (F9) to enamel was severely compromised after ageing (2.2±2 MPa) (p=0.003) and associated high pretest failures (PTFs=4). The same trend was noticed in the bond strength of resin composite (FS) to enamel that registered the highest early values among groups (29.1±2 MPa) (p<0.001), but over time the bond strength was minimised to ≃17 MPa.

Analysis of failures is presented in Figure 4-5. pRMGIC presented the same pattern of failures in both times, recognised by predominantly cohesive failures within the cement (62.5%) and mixed mode (37.5%). The same trend was seen in the delayed failure of F2LC that showed more adhesive failures at the early term (62.5%). Adhesive failures were also noticed in FS at either time (87.5%, 62.5%, respectively), and F9 at the first interval (75%), which was shifted to mixed fracture (75%) after storage.

Table 4-3 Shear bond strength SBS (mean [SD]) to sound versus demineralised enamel surfaces after 24 h and three months' ageing

Groups **n=8**	**Sound enamel**				**Demineralised enamel**			
	SBS 24 h	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**	**SBS 24 h**	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**
pRMGIC *Experimental*	24.3 [1.8]*	0	26.0 [3.4]*	0	9.9 [1.0]$^{*+}$	0	13.2 3.5]$^{*+}$	1
F2LC *Control*	18.7 [1.9]	0	17.6 [1.7]^{a}	0	6.7 [1.3]$^{+}$	0	7.4 [4.0]$^{bc+}$	2
F9 *(CGIC)*	6.8 [2.8]	1	2.2 [2.3]$^{\wedge}$	4	5.2 [1.1]	0	4.6 [2.0] $^{c+}$	1
FS *(Composite)*	29.1 [2.0]	0	16.8 2.0]$^{a\wedge}$	0	8.8 [1.4]$^{+}$	0	8.9 [1.4]$^{b+}$	0

(*) a statistically significant difference of the experimental cement (pRMGIC) from the control group. Similar letters in each column indicate no significant differences among groups (Bonferroni test post-hoc tests, an alpha level of 0.05). (^) a significant effect of ageing for the same group from the 24 h values with in each row (Independent t-test), ($^{+}$) significant statistical differences in values in demineralised surfaces from the sound enamel, (n=8)

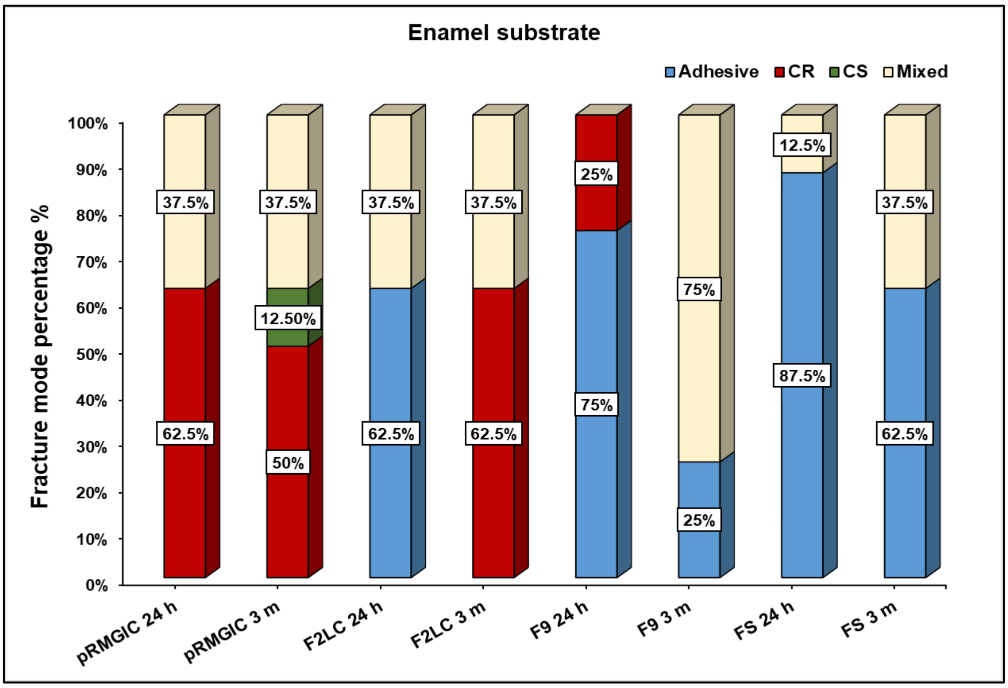


Figure 4-5 Mode of failure of each material from enamel substrates over time

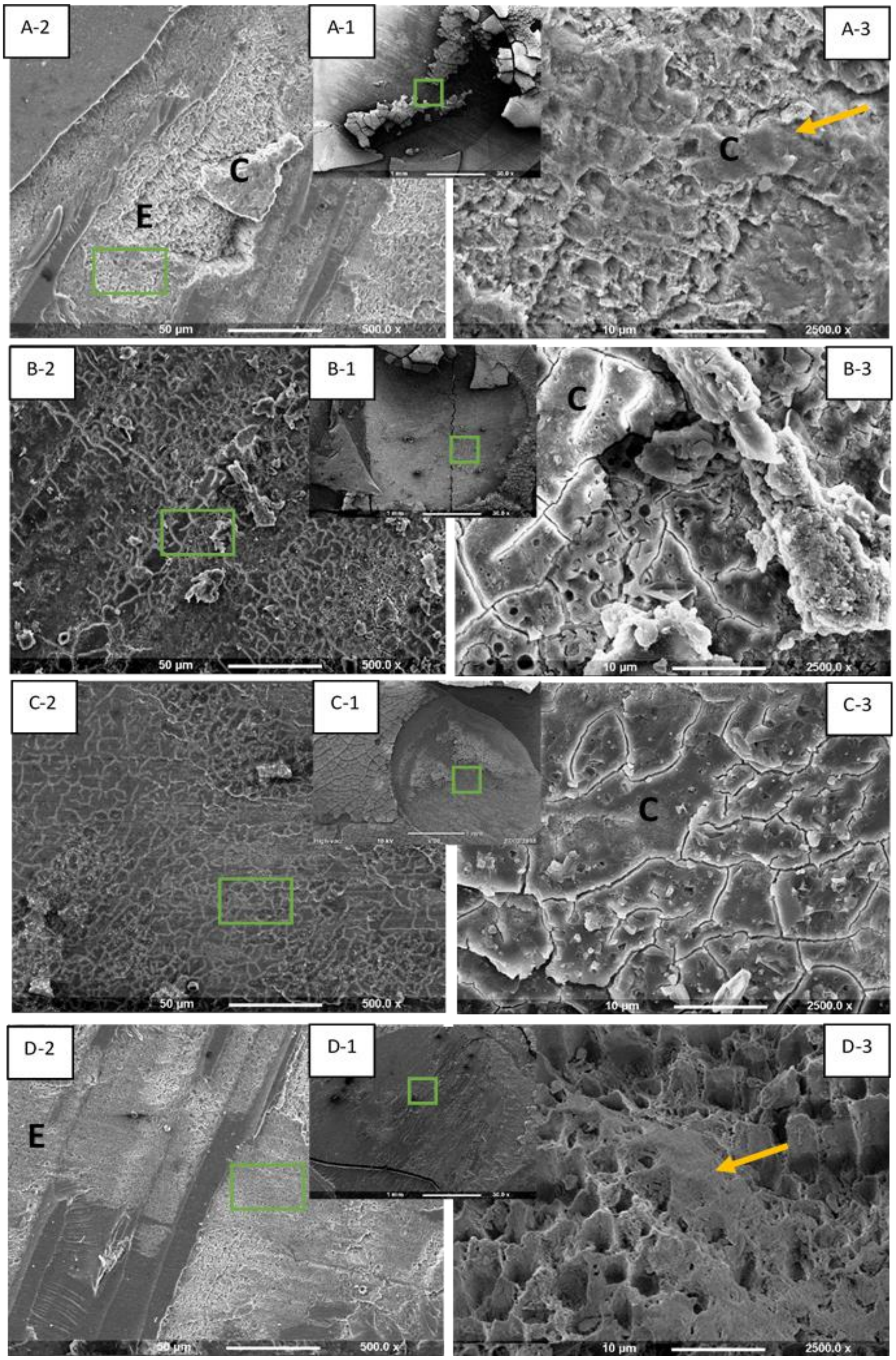


Figure 4-6 Representative scanning electron micrograph of the fractured surfaces along the sound enamel interface of specimens which debonded from pRMGIC (A), F2LC (B), F9 (C), and FS (D) after three months storage in SBF at x30, x500 and x2500 magnifications. All specimens showed mixed failures associated with cohesive failure within the enamel in A-2, and D-2. pRMGIC appeared to be well integrated to the exposed enamel rods in A-3 (yellow arrow) while minimal resin coverage could be recognised in D-3 (yellow arrow). In B and C, the debonded interfaces are covered entirely by porous cements. E-enamel, C-cement.

Interestingly, pRMGIC exhibited a profound enhancement in SBS to demineralised enamel substrate, as it registered the highest values in comparison to all applied materials at either time (9.9±1, 13.2±3 MPa, respectively) (p>0.05). Initially, SBS value of FS to demineralised enamel was statistically significantly higher (8.8±1 MPa) than F2LC and F9 (6.7±1, 5.2±1 MPa, respectively) (p<0.001). However, after three months, the delayed mean value of FS (8.9 ±1 MPa) was comparable to that of F2LC (7.4±4 MPa) (p=1.000) that showed no statistically significant difference from F9 (p=0.866). PTFs were observed in aged F2LC (n=2), and F9 (n=1).

There was a further development in bond strength of pRMGIC (33% increase) to this substrate post storage, but statistically, it was not significant (p =0.177) and associated with pretest failure (PTF=1). A similar trend was recognised in the other groups, as there were no detected changes in their long-term bond strength values in comparison to their early strength values (Independent t-test, p>0.05). The percentage of failure modes are summarised in Fig (4-7). The result indicates that adhesive failure was the predominant mode at the early stage of testing at 62.5% in all groups whilst the rest 37.5% are cohesive and mixed failures. In contrast, mixed and cohesive failures are the predominant mode of failures in the delayed term. Additionally, all groups showed cohesive failure within the white lesion post storage.

When considering the condition of the substrate (healthy versus defective), there was a statistically significant reduction (p<0.001) in the immediate and prolonged adhesion strength of the RMGICs (experimental & control) and the composite resin groups (FS) when bonded to demineralised enamel versus sound surface. In contrast, CGIC group (F9) maintain similar early bond strength values to both substrates (p=0.148), however, the difference was significant post-ageing. F9 showed enhanced bonding strength to demineralised enamel surface in comparison to sound (p=0.040), associated with less number of the PTFs, Table 4-3.

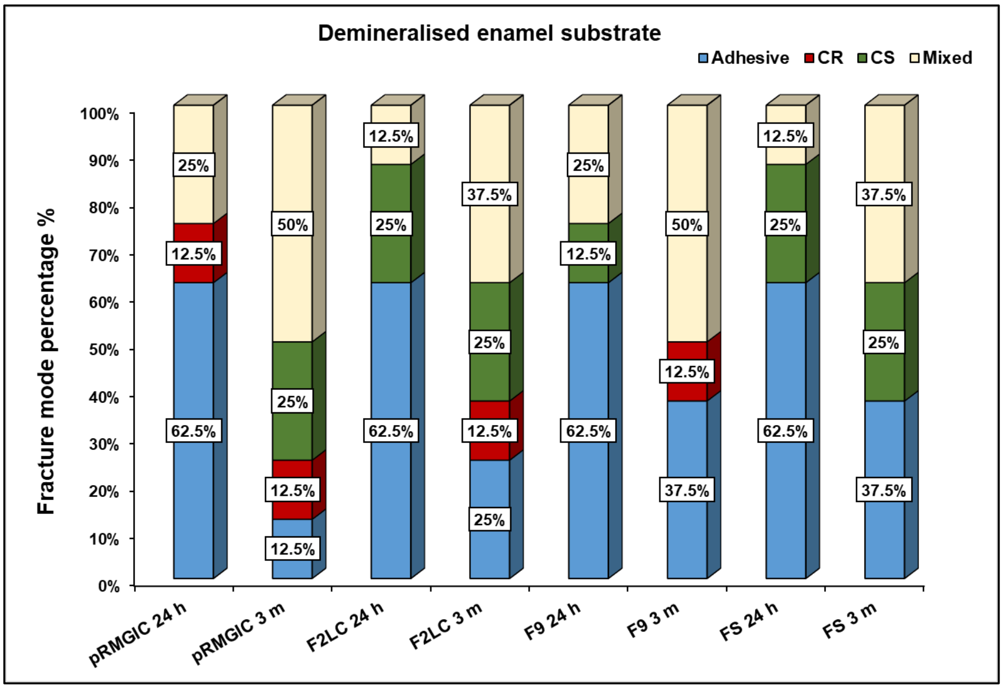


Figure 4-7 Mode of failure of each material from demineralised enamel substrates over time.

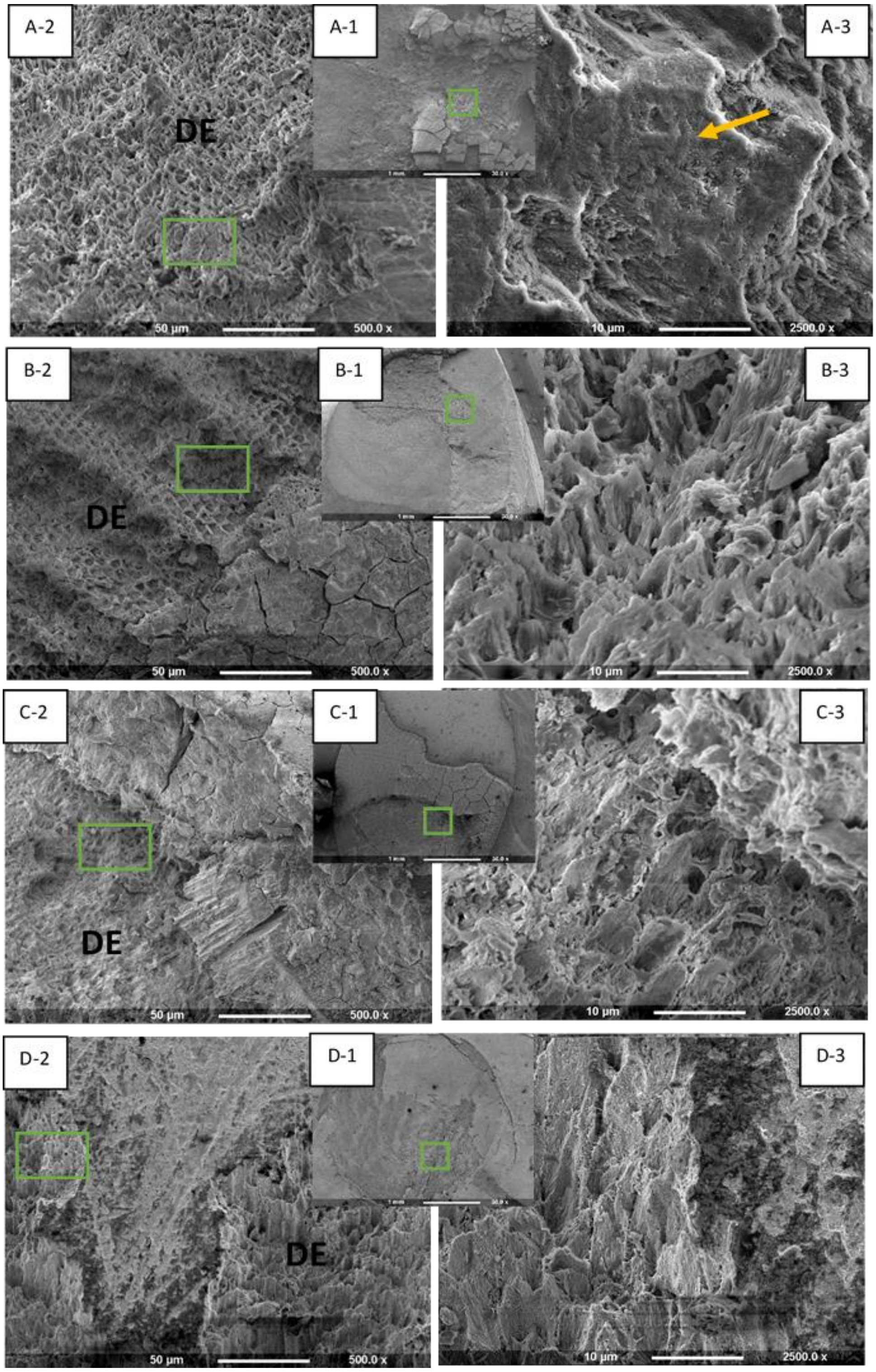


Figure 4-8 Representative scanning electron micrograph of the fractured surfaces along the demineralised enamel interface of specimens which debonded from pRMGIC (A), F2LC (B), F9 (C), and FS (D) after three months storage in SBF at 30x, 500x and 2500x magnifications. Mixed failures in all groups are associated with cohesive failure within the WSLs. At higher magnification, only pRMGIC cement was firmly attached to the lesion surface (A-3) (yellow arrow), which could not be recognised in the other groups. DE-demineralised enamel, C-cement.

4.3.2 SBS to sound and caries affected dentine substrates

The shear bond strength to sound and caries affected dentine are summarised in Table 4-4. PTFs are presented in the table and were given a “0” value for statistical analysis. The incorporation of a phosphate-based monomer to the RMGIC boosted the early and delayed SBS to sound dentine from 8.2±1, 10.7±4 MPa, respectively to 11.7±2, 18.8±2 MPa, respectively ($p<0.05$). Initially, the bond strength of FS group (8.7±1 MPa) was comparable to F2LC (8.2±1 MPa) ($p=1.000$). However, after three months’ storage, the adhesion strength of FS (17.9±2 MPa) was equal to that of pRMGIC (18.8±2 MPa) ($p= 1.000$). Moreover, CGIC registered the lowest immediate bond strength (6.2±1MPa) among all materials ($p<0.05$), with time, the bond strength was comparable to F2LC ($p=0.539$). Three months’ ageing resulted in further increase in the SBS values of all applied materials to sound dentine when compared to their early values which were statistically significant in pRMGIC and FS (60-100%, respectively) ($p<0.001$).

The enhanced strength to dentine was accompanied by a shift in the mode of failure for all tested materials from a predominant adhesive mode in the early term to mixed and cohesive modes on the delayed term. After 24 h, all specimens in FS and F2LC groups, and 75% of pRMGIC and F9 have failed adhesively from the surface. With time, the percentages of adhesive failures reduced to half for all groups, and were substituted by mixed failures (50%) in the GIC/RMGICs, in which a thin layer of the cement was still attached to the conditioned dentine surfaces, whereas the remaining percentages were cohesive within the materials, Fig (4-9).

Table 4-4 Shear bond strength SBS (mean [SD]) with the number of PTFs to sound versus carious affected dentine after 24 h and three months' ageing

Groups **n=8**	**Sound dentine**				**Caries affected dentine**			
	SBS 24 h	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**	**SBS 24 h**	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**
pRMGIC Experimental	11.7 [2.0]*	0	18.8 [2.5]$^{*b\wedge}$	0	13.9 [1.0]$^{d+}$	0	14.8 [2.2]$^{*e+}$	0
F2LC Control	8.2 [1.4]^{a}	0	10.7 [4.0]^{c}	1	13.9 [1.8]$^{d+}$	0	7.7 [3.0]$^{f\wedge}$	1
F9 (CGIC)	6.2 [1.1]	0	7.6 [3.2]^{c}	1	7.7 [1.2]$^{+}$	0	7.0 [2.5]^{f}	0
FS (Composite)	8.7 [1.4]^{a}	0	17.9 [2.4]$^{b\wedge}$	0	16.5 [1.6]$^{+}$	0	14.5 [1.6]$^{e\wedge+}$	0

(*) a statistically significant difference of the experimental cement (pRMGIC) from the control group, Similar letters in each column indicate no significant differences among groups (Bonferroni test post-hoc tests, an alpha level of 0.05). (^) a significant effect of ageing for the same group from the 24 h values with in each row (Independent t-test). ($^{+}$) a significant statistical difference in values of the CAD surfaces from sound dentine, (n=8)

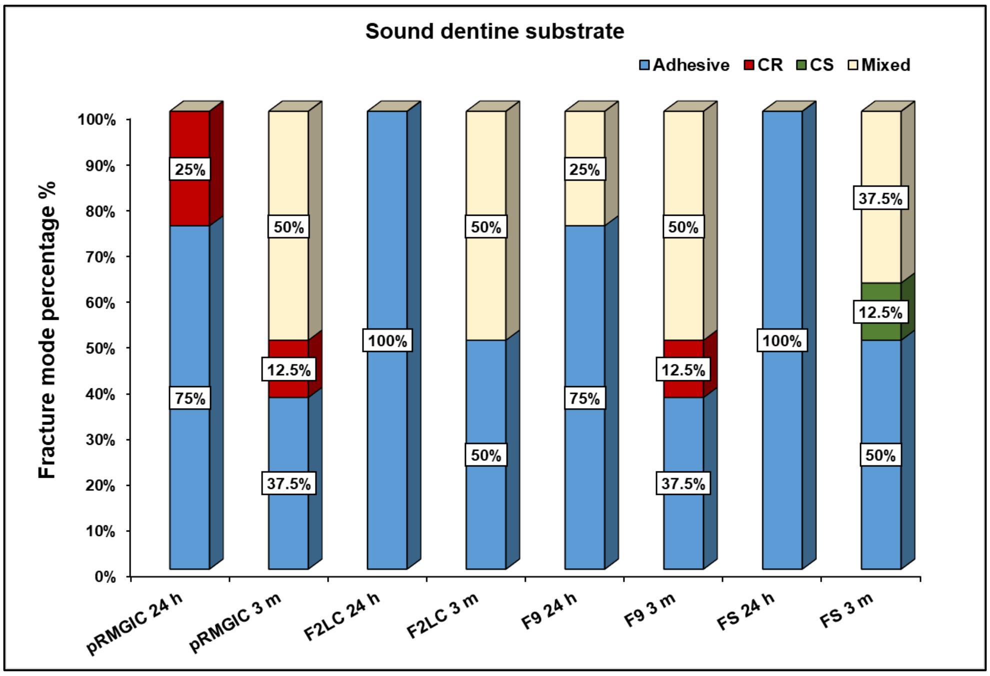


Figure 4-9: Mode of failure of each material with sound dentine substrate over time

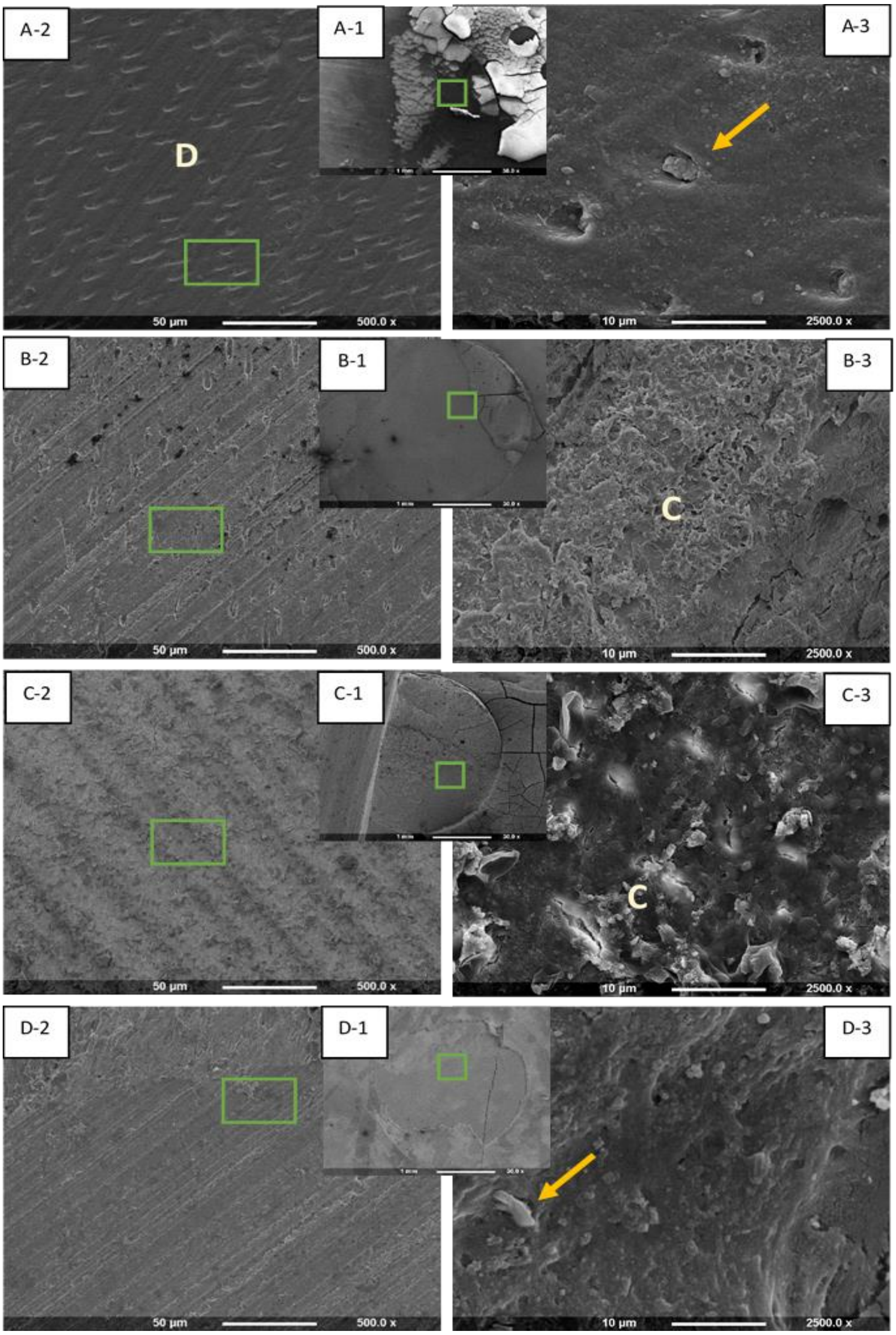


Figure 4-10 Representative SEM observations of the fractured surface along the dentine surface of specimens bonded to pRMGIC (A), F2LC (B), F9 (C), and FS (D) after three months storage in SBF. All failures are mixed located mainly within the hybrid complex. In A-3 and D-3 failures are found within the hybrid layer, showing numerous obliterated dentinal tubules by resinous tags as shown by yellow arrows. The precipitation of mineral formations only allowed a restricted display of the tubule entrances, deposited in progressive strata until the complete sealing of the lumen of tubule (arrow). While failures in B-3 and C-3 are located above the hybrid layer where the debonded interface is entirely covered by the cements. D-dentine, C-cement.

The early mean SBS of pRMGIC to caries affected dentine was similar to the control (13.9 MPa) (*p*=1.000), while resin composite (FS) recorded the highest mean value among groups. After three months, the new cement maintained its bond strength to CAD (≃15 MPa), while the control showed a significant drop in SBS values (7.7±3 MPa) (*p*<0.001). The delayed mean SBS of pRMGIC was comparable to that of FS group (≃14 MPa) (*p*=1.000). CGIC recorded the lowest initial values among groups (*p*<0.05), but the bond strength to CAD remained unchanged over time (*p*=0.469) and was comparable to that of F2LC (*p*=1.000), Table (4-4).

Three months' ageing did not improve the SBS of all materials to CAD. However, F2LC and FS showed deteriorated strength values after storage (*p*<0.001, *p*=0.025, respectively).

The mode of failure of the experimental and control bonded to sound and carious affected dentine are shown in Fig 4-11. Both the experimental and control groups which exhibited similar SBS values after 24 h showed a similar pattern of failure. The predominant mode of failure was cohesive (62.5%) within the carious surface/or the cement, whilst the rest (37.5%) yielded a mixed failure where remnants of the cement were found to smear the carious lesion. On ageing for 3 months, the percentage of cohesive failure was lower in both groups. It is interesting to note that failures in pRMGIC are mostly mixed whilst they are mostly adhesive in F2LC.

Although immediate and prolonged bond strength values of the CGIC to the carious dentine were the same, the mode of failure changed from mostly adhesive to mixed failures with no PTFs recorded at both intervals. Failures in the FS group are predominantly adhesive at both time points.

By comparing the adhesion strength to healthy versus carious dentine surfaces, initial strength values of all applied materials to CAD were statistically significantly higher (*p*>0.05) than sound substrates. However, the bond strength to sound dentine improved post-ageing compared to a drop in strength to CAD. The reduction in strength was statistically significant in pRMGIC and FS (*p*<0.05), but it was not significant in CGIC and F2LC (*p*=0.646, *p*=0.200, respectively).

The pRMGIC cement showed a beneficial effect on maintaining significantly higher prolonged SBS values to both healthy and defective dentine tissue when compared to the control cement (F2LC). The SBS values were comparable to the composite resin (FS) used with self-adhesive bonding agent (SU). However, the mode of early and delayed failures in the experimental cement are either cohesive within the cement or mixed mode where a thin layer of the cement over the carious lesion was observed in comparison to the predominantly adhesive failure in FS.

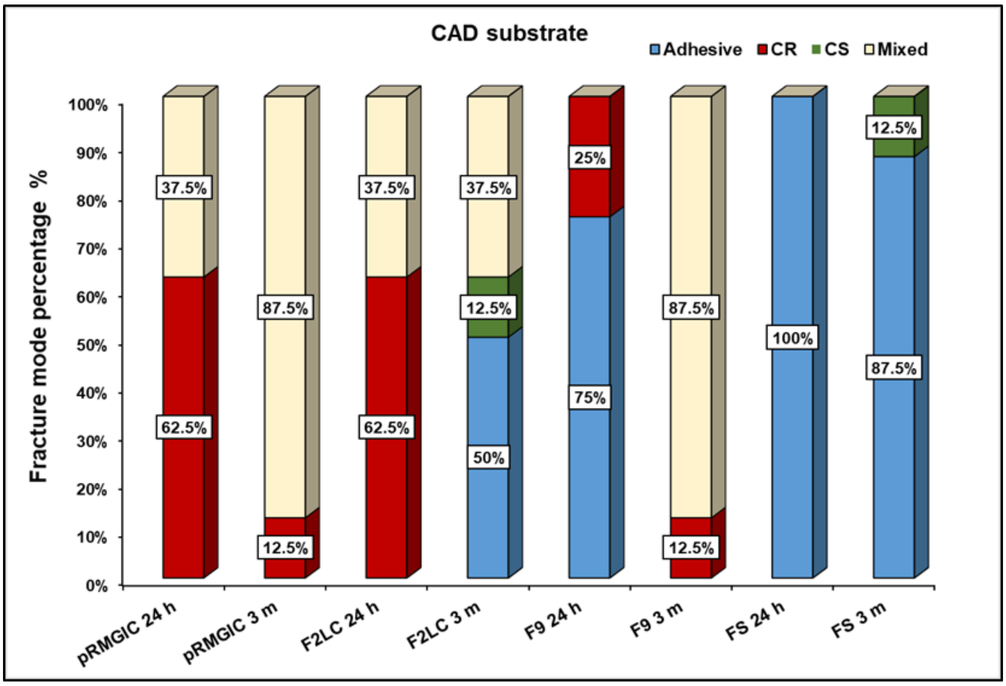


Figure 4-11 Mode of failure of each material with CAD substrate over time.

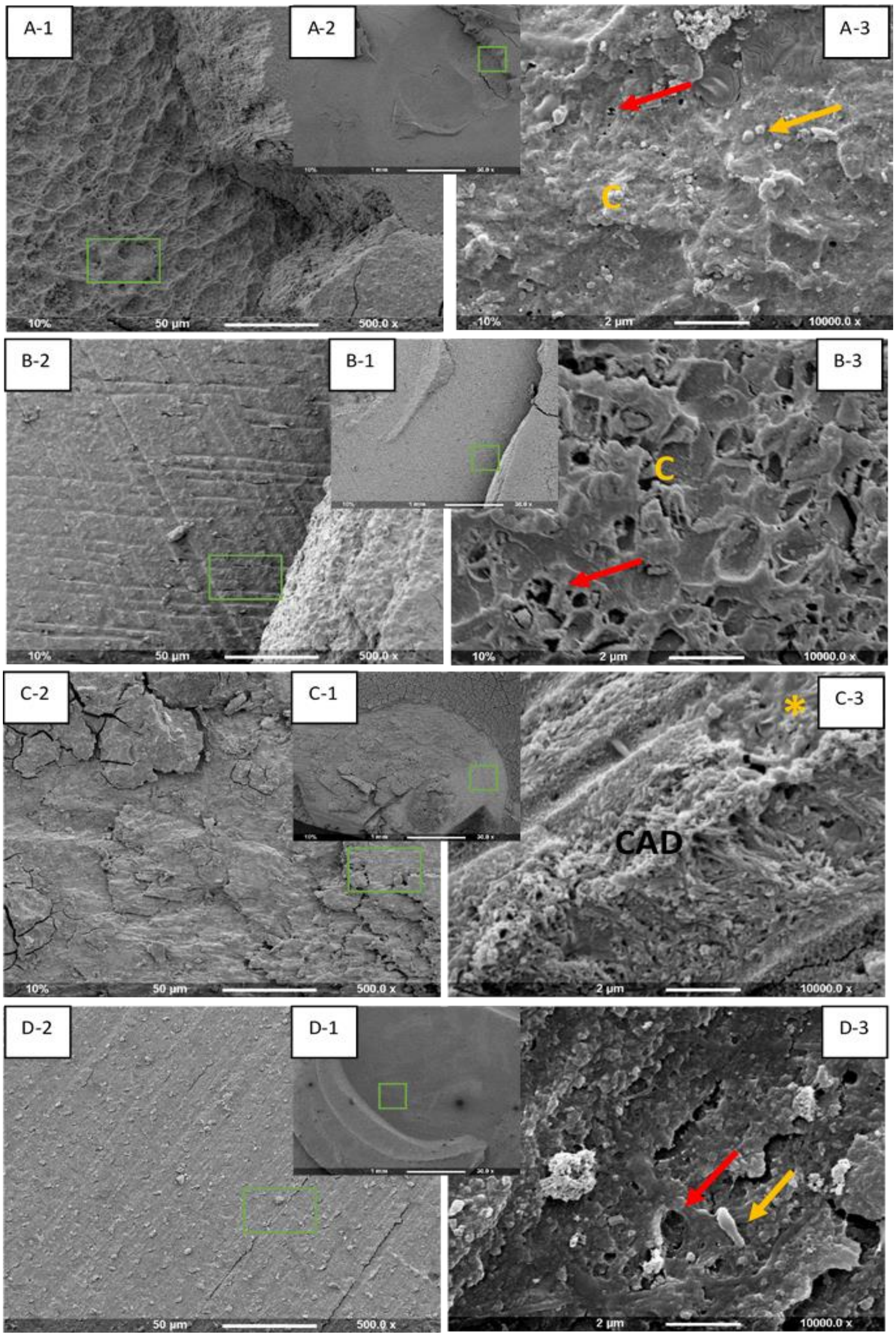


Figure 4-12 Representative SEM of mixed failures in CAD substrates that were treated with pRMGIC (A), F2LC (B), F9 (C), and FS (D) after 3 months' storage in SBF. In A-3, D-3 failures occurred within the hybrid layer where the CAD surface is well infiltrated by the resin components of the pRMGIC and resin composite with some resin tags remain occluding the tubules (yellow arrows), however, there was an evidence of less infiltration area disclosing patent dentinal tubules (red arrows). B-3 failure occurred above the hybrid layer where porous cement sparsely observed covering the lesion exposing parts of the underlying carious dentine with no resin tags could be recognised (red arrow). C-3, F9 specimen shows a cohesive failure within carious dentine in which a minimal cement coverage was seen (yellow asterisk) with apparent naked collagen fibrils, C-cement, CAD-carious affected dentine.

4.4 Discussion

The challenging mission for minimally invasive dentistry is to develop a reparative material with adequate physical properties that can repair failed TRCs by encouraging proper chemical integration with sound and carious tooth surfaces. Thus, to provide a durable marginal seal against degradation and prevent the reoccurrence of secondary caries. Additionally, it must have the ability to repair defective tooth surfaces and strengthen the tooth/restoration interface, thereby enhancing the longevity of TRCs. The efficacy of this new class of material pRMGIC as discussed in Chapter 3 was investigated via detailed analysis of the shear bond strength with different tooth tissues. Presently the lack of a dedicated universal repair material has seen a variety of materials used for repair or replacement, hence comparative studies have been carried out with a resin modified glass ionomer (F2LC), which is also the cement that has been used to develop the pRMGIC's, a conventional glass ionomer and a composite with a bonding agent. The pRMGIC possess pendant acidic phosphate groups prior and post setting since they are not involved in the photopolymerisation confers an ability to further demineralise tooth (healthy and diseased) surfaces and augment chelation with hydroxyapatite simultaneously, which is expected to yield a more durable bond. All selected materials are used in a traumatic restorative treatment (ART). Conventional and resin-modified glass ionomers are considered equivalent to some degree since both are essentially water-based cements and the primary setting reaction remains an acid-base neutralisation reaction. Both are used for similar clinical applications; as they used for repairing and restoring relatively minimal cavity preparations, which exploit their inherent adhesion to tooth tissue. The bonding efficacy of the new cement was also compared to a composite resin with self-etch adhesive bonding agent although the mechanism of the composite/adhesive restorations is entirely different from the GIC systems. Self-etching primers contain acidic resin monomers that can etch hard dental tissues such as enamel, thereby creating a stronger hybrid layer hence the performance of the pRMGIC in comparison is relevant.

4.4.1 Sound enamel substrates

The early shear bond strength of the F2LC to sound enamel was in agreement with previously reported values ≃18.7 MPa (Davidson and Mjör, 1999; Glasspoole et al., 2002; Van Noort, 2014). In comparison the pRMGIC, which has essentially the same alkaline glass exhibited a statistically significant higher values of SBS with a mean value of 24.3 MPa. The pRMGIC, like RMGICs, comprises of acidic polymers which provide free acidic carboxyl groups for ionic interactions, and methacrylate polymerisable groups which form a covalently bonded polymer network. The photopolymerisation yields the initial setting of the pRMGIC similar to RMGIC's however the presence of the phosphate groups in the system might promote the self-etching property of the original cement. The adhesion with enamel is an acid-dependent process, so both acidity and the type of acid-functional monomer influence the bond strength as previously explored by Münchow et al. (2015). The enhanced self-etching effect leads to higher prisms rod exposure in enamel and greater micro-mechanical interlocking which increases the bond strength values. The pendant phosphate units in pRMGIC may partly be neutralised via the interaction of the mineral, however the higher concentration of the EGMP in the formulation leads to higher abundance of the phosphate units which influences the interaction with the enamel that presented by 30-48% enhancement in early and delayed mean SBS values of pRMGIC when compared to F2LC (p<0.001). Of note that this is not related to just the reduction in pH of the liquid phase alone but also the ability of the acidic monomer to reach the surface in sufficient quantity and withstand the buffering effect of the reactive glass and enamel minerals, which needs to be further investigated. On the other hand, the functionality of the acidic monomer with the carboxylic groups is expected to provoke more chemical interactions by complex formation with the calcium of the apatitic substrate through ionic bonding, as reported for polyalkenoic acid interactions (Yoshida et al., 2000). Given the above, pRMGIC demineralises enamel and chemically integrates with the surface simultaneously producing stronger bond when compared to the F2LC cement (p< 0.001), which remained stable after three months' storage in SBF (26 MPa) (T-Test, p=0.233). Failures are predominantly cohesive in pRMGIC in which the cement was shown well-integrated to the exposed enamel rods under SEM, Figure 4-6 (A-3). In contrast, a porous cement structure covers the interface in F2LC as shown in Figure 4-6 (B-3) which might reflect that F2LC was failed cohesively before debonding.

It is not surprising that the higher displayed early bond strength to enamel achieved by composite resin (FS) with Scotchbond™ Universal self-etch adhesive (29.1 MPa). The adhesion of composite to enamel relies fundamentally on the mechanical interlocking of a solidified adhesive into the irregularities of the adherent surface (Kimmes et al., 2010). The lower viscosity and molecular weight of the monomer in the bonding agent in comparison to that of the polycarboxylic-based polymer (5000-30000 g/mol) (Nicholson, 2016), induce more resin infiltration into the microporosities of conditioned enamel surface. This produces higher bond strength when compared to the GIC systems (F2LC & F9) ($p< 0.001$). On the other hand, the potential benefit of additional chemical interaction between the functional monomer (10-MDP) and the enamel tissue might produce a stable bond over time (Yoshida et al., 2004). However, both chemical interactions and hydrophobicity of the Scotchbond™ Universal may not be able to withstand the stresses associated with the polymerisation shrinkage of resin composite or resist the hydrolytic degradation of the resin-enamel interface over time (Santerre et al., 2001), which compromised the prolonged bond strength of FS (16.8 MPa) as compared to its early SBS value ($p< 0.001$). Stresses exerted on immature bond leads to predominant adhesive failure of the composite from the substrate, as shown previously in the literature (Takamizawa et al., 2018). Although there was a reduction in SBS value post storage, but there was a marked increase in the mixed mode with cohesive failure within the substrate. SEM findings confirmed that a remnant of the bonding agent was firmly attached to the underlying enamel rods, Figure, 4-6 (D-3).

There is an agreement in literature that RMGICs exhibit higher bond strength to enamel than CGIC which showed the lowest values in this study amongst the groups at both time intervals (6.8, 2.2 MPa) with higher numbers of PTFs (n=4) after ageing. In fact, the adhesion in RMGICs is partly micromechanical, due to the penetration of polymer tags into the microporosities of the conditioned enamel, with ionic exchange and chemical bonding (Glasspoole et al., 2002). While in CGIC adhesion is mainly dependent on an ionic interaction with the mineral phase. Theoretically, the initial attraction of GIC with the tooth surface is primarily due to polar interaction with week hydrogen bonds which rapidly buffered by the phosphate ions arising from the hydroxyapatite. The continuing development of bond is thought to be caused by further movement of the ionic species in the interface due to diffusion, as the phosphate ions are displaced by the polyalkenoic acids (Wilson and McLean, 1988). Thus, the bond strength of CGICs, unlike resin-based systems, should evolve with time. However, the dimensional instability of the cement due to excess water uptake or desiccation (Watson & Banerjee, 1993; Watson et al., 1998) might restrict the surface attraction and affect the development of the adhesive bond. This might attribute to the low delayed bond strength values associated with premature debonding of the cement before testing (n=4). The first hypothesis was rejected, as the new cement showed a significant difference from all groups in both times (p<0.05). In contrast, the second hypothesis was partially rejected due to the significantly compromised bond strength to enamel over time by F9 and FS groups (p<0.05).

4.4.2 Demineralised enamel substrates

The assumption that adhesive strength to a hypomineralised enamel surface is lower than sound was previously reported by William et al. (2006) and Krämer et al. (2018). In accordance with this assumption, all resin-contained materials in the current study (pRMGIC, F2LC, and FS) showed more than 100% reduction in the immediate and delayed strength to demineralised enamel versus healthy substrate (p<0.001). This is attributed to the morphological and structural alterations in enamel on prolonged exposure to acids. These changes include lower hardness values and irregular apatitic structure as previously explored by Xie et al. (2008); and Chan et al. (2010), and associated with porosities at the resin-enamel interface (William et al., 2006). These factors lead to disruption of the bonded margins or even retention loss. Theoretically, micromechanical retention is a function of surface area and surface energy of the etched enamel (Reis et al., 2003). Hypomineralised enamel exhibited an increase in the surface area but the irregular etch patterns following the application of self-etch adhesives, and the presence of porosities and cracks at the hybrid layer may compromise the resin infiltration into the substrate, as previously illustrated by William et al. (2006). These surface imperfections are expected to reduce the early bonding strength of resin composite in the present study from 29 MPa in the sound group to 8.8 MPa in the demineralised one (p<0.001). The weak bond strength to demineralised enamel yielded cohesive failures within the white lesions in one-fourth of specimens at either time. In contrast, adhesive failures were dominated when applied to sound enamel. On the other hand, the potential chemical interaction of the functional monomer (10-MDP) in SU with the residual hydroxyapatite in the porous enamel might induce stable calcium salts within the hydroxyapatite lattice (Yoshida et al., 2004). This may contribute to the sustained bond strength of FS to demineralised surface after three months' storage (8.9 MPa) (p=0.877). The fracture mode was also changed from mostly adhesive (62.5%) in the first interval to cohesive and mixed failures with time, (Fig, 4-7).

The compromised SBS of resin contained materials is also noticed in RMGICs groups (Experimental and control), which demonstrated a severe reduction in immediate and delayed bond strength (>100%) when compared to sound enamel (p<0.001). However, the ionic interaction with the demineralised enamel produced stable bonds over time for both groups (p=0.117, 0.678, respectively). It was also associated with a shift from mostly adhesive failure to cohesive and mixed failures within the WSLs post-ageing, Figure (4-7). Interestingly, pRMGIC exhibited an enhanced immediate and delayed bonding to this substrate (9.9, 13 MPa) when compared to all applied materials (p<0.05). This might be attributed to the acidity and functionality of the incorporated monomer in addition to the polyalkenoic acid effect, which might induce further demineralisation at the interface. Furthermore, the inductive ability of the phosphate groups encourage CaP complex formation via chelation with the residual calcium ions as indicated in previous findings (Stancu 2004). These might be precipitated in the form of minerals which potentially penetrate the lesion and enhance the remineralisation that reinforces the substrate with higher resistance to shear loading over time (≃13 MPa). Scanning electron micrograph confirms these findings as it showed that pRMGIC is firmly integrated to the underlying exposed enamel rods within the lesion which could not be recognised in the other applied materials, Figure 4-8 (A-3). However, extensive qualitative and quantitative measurements at the interfacial area are needed to confirm these speculations.

The SBS values of the conventional GIC and RMGIC to both sound and demineralised were similar at 24 h. The disorganisation of the hydroxyapatite crystals with lower mineral content in the hypomineralised enamel (Jälevik et al. 2001; Farah et al., 2010) produces a porous structure which permits fluid movement (Bertucci et al., 2008). This moisturised environment facilitates the adaptation of the CGIC and contributes to the neutralisation reaction between the acidic functional groups and the ion-releasing basic filler particles. Additionally, the hydrogen bonds that formed between the free carboxylate groups and the layer of tightly bound water at the hypomineralised surface of the tooth (Hinoura et al., 1991) are gradually replaced by genuine ionic bonds (Yoshida et al., 2000). These interactions produce strong stable strength of F9 (4.6 MPa) significantly higher than the delayed strength to sound enamel (2.2 MPa) in which bonding resists the plasticising effect of water. Interfacial failure analysis support SBS results, as it shows a shift from mostly adhesive failure (62.5%), to cohesive and mixed within the WSLs, Figure (4-7). Accordingly, the first stated hypothesis was rejected, but the second one was accepted as ageing does not affect the bond strength of all materials to WSLs.

4.4.3 Sound dentine substrates

The presence of the phosphate-based monomer in the structure of RMGIC boosted the early SBS to sound dentine from 8.2 MPa to 11.7 MPa (*p*<0.001), which is even higher than the previously reported values for the RMGICs in the literature (≃7 MPa) (Davidson and Mjör, 1999; Nicholson, 2016). After three months, the adhesion strength of pRMGIC reached 18.8 MPa, which is almost double than reported for RMGIC and significantly higher than its immediate value (*p*<0.001), and the control group (10.7 MPa) (*p*<0.001). It is clear from the results that the acidity and reactivity of this functional monomer played a significant role in elevating the initial and delayed bond strength of RMGIC to healthy dentine. EGMP tends to be ionised in the presence of water and produce protons (Lima et al., 2008) which can promote the self-etching property of the RMGIC. This might increase the surface area for adhesion and produce micro-porosities in the surface for more micro-mechanical interlocking or hybridisation (Van Meerbeek et al., 1998). Furthermore, the initial cement hydrophilicity, due to the polarity of the phosphate moiety and the presence of HEMA, with a naturally hydrated substrate facilitated the ionisation of these acidic monomers and improved the infiltration of the polymer into the substrate. Subsequently, water is expected to be reused by reaction between the acidic functional groups and the ion-releasing basic filler particles during cement setting reaction. So it provides improved adaptation to the tooth structure with moisture tolerance. Additionally, the functionality and chelating effect of the phosphate group may impart in more chemical interactions with the residual hydroxyapatite (Fu et al., 2005) that lead to robust and durable bond over time (18.8 MPa). It is accompanied by a shift in the mode of failure from predominantly adhesive (75%) to mixed and cohesive patterns (62.5%) after ageing, (Fig 4-9). SEM supports these findings as it shows mineral formation in the dentinal tubule entrances which deposited gradually until the complete sealing of the lumen of the tubule as shown in figure 4-10 (A-3). In contrast, failure in F2LC and F9 occurs above the hybrid layer where cements' remnant entirely covers the debonded interface, Figure 4-10 (B-3, and C-3, respectively).

The delayed bond strength of pRMGIC was comparable to the composite resin FS (17.9 MPa) (*p*=1.000) which showed an apparent enhancement in SBS from its initial value (8.7 MPa) (*p*<0.001). This might be correlated to the chemical bonding ability of the functional monomer (10-MDP) in SU to the hydroxyapatite which produces stable calcium salts (Yoshida et al. 2004; Inoue et al., 2005). These interactions create an efficient and durable bond to dentine over time, with an adequate seal against nanoleakage as previously reported by Wagner et al. (2014) and Sezinando et al. (2015). The enhanced strength of the composite resin was associated with a decrease in the percentage of the adhesive failure to the half after ageing which were replaced by mixed failures (50%) (Fig 4-9), whereby a thin layer of the adhesive still attached over the dentine surfaces, associated with the presence of numerous resinous tags obliterated dentinal tubules under SEM, as shown in figure 4-10, D-3.

The immediate bond strength achieved by F9 to sound dentine was slightly higher than the previously reported values (> 5MPa) (Berry & Powers, 1994; and Burke & Lynch, 1994). Initially, a polar attraction by weak hydrogen bonds is the predominant interaction, which is rapidly buffered by the phosphate ions from the hydroxyapatite crystals. The good wetting of the dentine substrate encourages this ionic exchange at the interface (Watson et al., 1991). Subsequently, these interactions are gradually displaced by a stronger interaction as the polyalkenoic acid chains diffuse into the dentine and replace phosphate and calcium ions from the hydroxyapatite crystals (Mount, 2005; Van Meerbeek et al., 2006). Therefore, these chemical interactions encourage tooth-glass ionomer bonds to increase over time and become eventually limited by the cohesive tensile strength of the cement. The same trend was noticed in the present study in which the bond strength of F9 increased after storage to 7.6 MPa and associated with a shift towards mixed and cohesive failures (62.5%).

In agreement with previous studies; Van Meerbeek et al. (1998), (2006); Tanumiharja et al. (2000); Coutinho et al.,2007; Falsafi et al., 2014. RMGIC showed higher adhesion strength to dentine (8.2 MPa) than CGICs (*p*=0.029). It arises from the intrusion of short cement tags into the conditioned dentine surface added to chemical bonding ability of the methacrylated copolyalkenoic acid to the calcium in HAP as previously shown via XPS and FTIR (Mitra et al., 2009). The wetting of the substrate and the cement due to the polarity of the HEMA also promoted the penetration efficacy to dentine. This might be responsible for producing an immediate bond strength (8.2 MPa) which was comparable to the composite resin FS with self-adhesive bonding (8.7 MPa) (*p*=0.890). Tay et al. (2004) identified the presence of an absorption layer over the hydrated dentine that was discernible from the hybrid layer when bonded interfaces examined by TEM. This layer was relatively more resistant to dehydration stresses and remained intact over the dentine surface after tensile testing. The absorption layer mediates better bonding of RMGIC to dentine, and functions as a stress-relieving layer to reduce stresses induced by desiccation and shrinkage. Additionally, the hygroscopic expansion of the RMGIC may compensate for the polymerisation shrinkage (Versluis et al., 2011), or the absorbed water aid in the completion of the neutralisation process (Young et al., 2004). All these factors might explain the increase in adhesion strength after three months, even if it was not statistically significant (*p*=0.202) with more mixed failures.

4.4.4 Caries affected dentine substrate

The bonding to natural caries affected dentine is considered as a challenge. This is because of the reduced mineral phase presented by lower Knoop hardness values (Ceballos et al., 2003; Banerjee et al., 2010), poor quality of the hybrid layer as previously showed by Nakajima et al. (2005), and the extra moisture (Ito et al., 2005) that might induce hydrolysis of the resin and collagen fibrils. These factors can jeopardise the micromechanical interlocking of the adhesive polymers producing a lower bond strength as compared to sound dentine (Nakajima et al., 1999; Sattabanasuk et al., 2005; Erhardt et al., 2008).

In this study, the pRMGIC was applied to CAD to improve the bonding potential of the original cement by encouraging more chemical integration with the substrate by virtue of phosphate groups. The principle is to seal the lesion, impede the progression of the carious process and reinforce the tooth-restoration complex. The proposed aim was achieved as the immediate recorded bond strength value (13.9 MPa) was significantly higher than that of sound dentine (11.7 MPa) (p=0.007). This attributed to the promoted self-etching effect which was previously explained, as it might induce further mineral loss from the lesion. However, a thicker hybrid layer might be formed which could jeopardise the micromechanical interlocking with the substrate. But, at the same time, more ions could be available at the interface for more ionic bonding. In addition to the potential ability to form CaP complexes between functional groups and the hydroxyapatite, as discussed earlier, which expected to reinforce the carious substrate and produce a stable bond over time. SEM observation which revealed failure within the hybrid layer supported these results, as it showed that CAD surface is well infiltrated by pRMGIC with some resin tags remain occluding the tubules, Figure 4-12 (A-3).

The same pattern was noticed in the control group at the early interval, as the early strength that was similar to pRMGIC (13.9 MPa) (p=1.000) is significantly higher than that to sound dentine (p<0.001). This might be attributed to the increased hydration in CAD surface which facilitates the adaptation of the cement to the surface with more mechanical interlocking to the substrate. After three months' ageing, the adhesion strength was severely compromised (7.7 MPa) (p<0.001), however, it was comparable to the delayed strength to sound dentine (p=0.200). This coincides with Marquezan et al., 2010, and Toledano et al., (2016) in term of the prolonged bond stability of RMGIC to CAD in comparison to sound dentine.

The reduction in strength over time was also true for the composite resin (FS) which exhibited a compromised adhesion strength (14.5 MPa) after storage when compared to its early value (16.5 MPa) (p=0.025), which was far from that with sound dentine (17.9 MPa) (p=0.004). The result agreed with Erhardt et al. (2008); Osorio et al. (2010); and Marquezan et al. (2010), in the limited bond stability of the adhesives/composite restorations to CAD irrespective to the adhesive systems that are used. The lower bond strength values to this substrate were reported when compared to sound dentine by other researchers; Yoshiyama et al. (2000);

Yoshiyama et al. (2002); and Ceballos et al. (2003). This is attributed to the higher organic component of the smear layer (Taniguchi et al., 2009) and the presence of highly disorganised collagen which forms the ‘collagen smear layer’ (Pashley and Carvalho, 1997). This smear layer is difficult to remove by mild acid-etching (Spencer et al., 2001) which impede resin infiltration into the substrate, thus, affecting the bond quality and increasing the susceptibility of the hydrolytic and enzymatic degradation at the interface when compared to healthy dentine (Hashimoto et al., 2000). When comparing the delayed strength of FS with SU to the new cement, the values were similar ≃ 14 MPa. This is attributed to the presence of reactive functional monomers that are expected to promote a chemical integration to the hypomineralised surface. However, the failure in pRMGIC was mostly mixed with some cohesive pattern as compared to the predominant adhesive failure in FS. This might indicate that the adhesion strength of pRMGIC to CAD was higher than the cohesive strength of the cement. Furthermore, cement was shown well integrated to the lesion under SEM (Figure 4-12 (A-3), which showed minimal porosities and cracks at the interface with more occluded dentinal tubules than that of FS, Figure 4-12 (D-3).

The chemical bonding ability of CGIC to the CAD offers durable adhesion over time. However, in agreement with former studies (Palma-Dibb et al., 2003, Calvo et al., 2014), the low cohesive strength of the cement produces lower SBS values than resin-contained materials. The initial bond strength of F9 to CAD (7.7 MPa) was higher than to sound (6.2 MPa) (p=0.021). After storage, the type of substrate (sound or caries-affected dentine) did not affect the bond strength of F9 (p=0.646). Following the ionic exchange concept of the CGIC with CAD, as previously explored by Ngo et al*.* 2006 and Sennou et al. 1999, there was a possibility of diffusion of calcium/strontium ions into the hypomineralised matrix combined with self-etching effect of the polyalkenoic acids which induce further demineralisation of the lesion (Sennou et al*.*, 1999). This eventually creates an ion-rich layer which might be followed by mineral deposition on the pre-existent nuclei (Atmeh et al., 2012). The mineralisation potential of CGIC was also suggested by previous studies (Yoshiyama et al., 2000; Czarnecka et al., 2007a; Calvo et al. 2014; Toledano et al., 2016) which is expected to strengthen the cement-dentine interface and account for this outcome.

4.5 Limitations and future work

Shear bond strength was chosen to test the ability of the new reparative material to adhere and potentially repair different tooth substrates (sound vs. diseased). It provides valid results that are easily compared to the previously reported values for the materials that are used for the same purpose. However, the variations in the tooth structure, patients' age and the nature of the carious lesions among teeth cannot be avoided. Another limitation was the difficulty to specifically measure the adhesion strength to CAD due to the large surface area of the bonded interface, since there is a difficulty in discriminating the various zones of the carious lesion during testing. The µSBS is expected to be more appropriate than SBS in measuring the small surface area, however, this technique is not suited in case of GIC/RMGIC due to difficulty of initial retention of the conventional GIC that recommends large bonded surface areas. The use of other tests like the µTBS and the interfacial fracture toughness (iFT) can be more effective in the measurement of the interfacial adhesion-tooth strength. The µTBS reduces the variability by using single tooth with a lower probability to incorporate the interfacial defects due to small surface bonded areas (Pashley et al., 1995; Smith, 1997). However, limitations such as the technically demanding specimen preparation, the difficulty to measure the low bond strengths (<5 MPa) and the potential of specimens' dehydration cannot be avoided. Moreover, the main stress is located within the tooth or the material near the interface rather than at the interface (Söderholm et al., 2012). Interfacial fracture toughness (iFT) has been proposed as an alternative effective tool to measure the efficacy of bonding via resistance to crack propagation. It promotes true interfacial failure with minimal cohesive fractures within dentine or cement (Souza et al., 2016). However, the specimens' preparation is time-consuming and very sensitive to technical errors. The limited data provided by this test in the literature makes the comparison of a new material to others quite difficult. Nevertheless, for future work both µTBS and iFT can be performed for more validated interfacial bond strength to different tooth surfaces.

The results of the current study showed significantly improved early and delayed SBS of the phosphorylated RMGIC to different tooth surfaces in comparison to the control. The possible reasons are partly due to the acidity of the added monomer which might confer the demineralising effect of the RMGIC, which is expected to enhance the mechanical interlocking of pRMGIC to substrates and provide more ions for chemical bonding. However it difficult to confirm whether this acidic monomer can reach the surface in sufficient quantity and withstand the buffering effect of the reactive glass and tooth minerals which might need to be further investigated. But at least this monomer does not reduce the self-etch property of the RMGIC.

The other possible reason is referred to the functionality of the acidic monomer and the carboxylic groups which expected to provoke more chemical interactions via complex formation with the calcium of the apatitic substrate, as reported for the polyalkenoic acid interactions (Yoshida et al., 2000). However, this is based on the ability of both functional groups to form hydrolytic stable calcium-phosphate complexes which might be difficult to assess due to the complex structure of the cement matrix, and the contribution of these functional groups in the cement formation process. Additionally, the improved strength post-ageing reflect either a reinforcement effect for the tooth structure due to mineral precipitation, or an enhancement in the cement properties due to maturation or both. This needs to be further investigated via measuring the change in the mineral profile of treated teeth at the interface using Raman microscopy, XPS, XRD, and surface microhardness. The failure analysis and SEM support the potential integrity of the pRMGIC to different tooth substrates. However, the chemical adhesion can be effectively evaluated by analysing the interaction of pRMGIC to pure HAp via FTIR analysis of treated HAp powder and XPS for the treated HAp disks. By this way, the effects of chemical bonding of pRMGIC to HAp crystals in enamel and dentine can be isolated via eliminating the biological variability in tooth structure.

Finally, the hydrophilicity of the developed cement may promote surface wetting that is necessary for adhesion to tooth substrate which might be responsible for the enhanced early strength values of pRMGIC and provide more water that helps in the completion of neutralisation reaction which enhanced the adhesion strength after three months. However, the increased hydrophilicity of this cement may jeopardise the adhesion strength over time due to the hydrolytic degradation. Accordingly, it advocated to measure the bond strength after long-term storage, thermocycling, and by using the cyclic loading which will reflect the adhesion strength under similar oral conditions.

4.6 Conclusions

Within the limitations of this study the conclusions are:

1. The first stated hypothesis was rejected, pRMGIC showed statistically significant higher SBS values to sound & demineralised enamel at both intervals, while the second hypothesis was accepted in pRMGIC & F2LC as there was no statistically significant difference in bond strength to sound/demineralised enamel after storage.
2. The first stated hypothesis was partially rejected when pRMGIC bonded to sound dentine at both time periods, as it showed statistically significant higher SBS than F2LC and F9 only, while in CAD, the hypothesis was partially rejected at the delayed interval only. The second hypothesis was rejected in pRMGIC and FS when bonded to dentine, and in F2LC and FS when bonded to CAD.
3. Failure analysis and scanning electron micrographs reflect the potential integrity of pRMGIC to either healthy or diseased substrates.

Chapter Five

5 Interfacial bond strength of the pRMGIC to different restorative substrates

5.1 Introduction

Complete replacement of failed restorations due to caries associated with restorations and sealants (secondary caries, CARS) (Green et al., 2015), fractures and marginal defects may not be the treatment of choice for defective restorations. Refurbishment, resealing and repair offer alternative forms of conservative intervention, especially when the impaired tissues or defective restorative margins are adequately accessed (Moncada et al., 2009). Although the effectiveness of repair vs. replacement is not fully established, the evidence as it currently stands seems to favour repair (Sharif et al., 2010a, Casagrande et al., 2017), since this minimally invasive approach preserves tooth structure and increases the longevity of restorations with a cost-benefit ratio for the patient and healthcare system.

Different repair materials and protocols have been suggested by researchers to enhance the functionality, longevity, aesthetics and the integrity of the restorative interfaces. However, currently no repair system guarantees a favourable clinical outcome. Generally, the repair strength is influenced by the type of the substrate material, surface conditioning protocol, ageing condition, and the reparative system used. Surface conditioning is greatly recommended to improve the micromechanical attachment between joint interfaces. They include micromechanical roughening (grinding or sandblasting), chemical adjuncts (different adhesive systems, resin-modified glass ionomers, alloy primers), or a combination of both (sandblasting followed by primers/adhesives) (Özcan et al., 2011; Özcan and Salihoğlu-Yener, 2011).

The higher mechanical strength, wear resistance and improved aesthetics of the adhesive resin composite restorations combined with the effective bonding ability to different restorative interfaces via the use of adhesive systems (Zhang et al., 2011), make them the preferable choice for repair. Resin composite restorations exhibit a predictable adhesion strength to air-abraded amalgam following the application of bonding agents (Blum et al., 2012; Çehreli et al., 2010) with lower incidence of microleakage between the repair-tooth/or restoration interfaces as compared to the amalgam-amalgam repair (Özcan et al., 2010; Popoff et al., 2011), even though the microleakage at these interfaces is still exist. For resin composite repair, the absence of oxygen inhibition layer of un polymerised resin combined with the surface changes due to ageing (Ferracane, 2006) compromise the chemical and micromechanical bonding to aged resin composite restorations (Papacchini et al., 2007c; Dall'Oca et al., 2007; Özcan et al., 2013). Accordingly, surface roughening has been recommended to increase the surface area through the creation of surface irregularities to promote the micro-mechanical interlocking into aged composite substrate. This can be done using different techniques; diamond bur, air-abrasion with aluminium oxide, bioactive glass particles, silica oxide (tribochemical particles), etching with 37% phosphoric acid or etching with 9.6% hydrofluoric acid (Özcan et al., 2005), followed by the application of an adhesive agent (Staxrud and Dahl, 2011). Resin composites generally offer reliable immediate bond strength to the restorative/dental substrates via the use of bonding agent. However, the dimensional instability during curing and hygroscopic expansion might generate detrimental stresses at the tooth /or restorative margins (Atai and Watts, 2006). In addition, the hydrolytic degradation of some composite resin restorations over time (Wei et al., 2011) can lead to de-bonding, micro-leakage followed by CARS, thus compromising the seal in long-term repair of failling restorations (Woolford, 1993; Taha et al., 2012).

Polyalkenoate glass-ionomer cements (GICs) offer a reliable, long-term chemical adhesion to tooth tissues, low coefficient of thermal expansion, good tissue biocompatibility and fluoride release with a potential reduction in the incidence of CARS (Yoshida et al., 2000; Van Meerbeek et al., 2003; Peumans et al., 2005). They provide an adequate seal against microleakage as compared to resin composite-composite restorations (Welsh and Hembree, 1985) with a possibility to remineralise affected caries dentine surfaces (Smales et al., 2005). GIC can adhere to mechanically-roughened amalgam (Aboush and Jenkins, 1989) and resin composite restorations. However, the use of bonding agent intermediated the GIC substrate and resin composite is recommended for better seal against microleakage (Mount 1989; Hinoura et al., 1989; Wooford and Grieve, 1993). In GIC-GIC repair, surface pretreatment by chemical and mechanical approaches including phosphoric acid or bur-roughening followed by etching, can enhance the repair strength (Jamaluddin and Pearson, 1993). Furthermore, the chemical bonding of the free polyacrylic acid from freshly mixed cement with the exposed glass components of aged GIC (Pearson et al., 1989) can stabilise the strength over time (Jianguo et al., 1996). However, the slow maturation of the GIC, the sensitivity to the clinical conditions, low cohesive strength and low wear resistance limit their use for long-term repair protocols (Pearson et al., 1989; Zoergiebel and Ilie, 2013). Thus, failed GIC restorations are repaired by resin composite restorations which are preferred by many clinicians (McLean et al., 1985; Green et al., 2015).

Resin-modified glass-ionomer cements (RMGICs) retain the clinical advantages of the conventional GICs regarding the chemical union with the tooth surface and fluoride release. They exhibit better aesthetics, an extended working time, low moisture sensitivity (Sidhu & Watson, 1995), with significantly improved diametral tensile and flexural strengths as well as fracture toughness as compared to GICs. Furthermore, RMGICs can adhere effectively to roughened amalgam surfaces without intermediary adhesive between them, as previously stated by Aboush & Jenkins, (1991). Other studies (Fruits et al., 1998; Pilo et al., 2012) suggested that RMGIC showed higher adhesive strength when sandwiched between resin composite and set amalgam, even better than the use of an adhesive. Surface roughening with a coarse diamond bur followed by etching using phosphoric acid, or polyacrylic acid can enhance the repair strength and modify the fracture pattern to entirely cohesive as shown by Maneenut et al. (2010), and Camilleri et al. (2013). The chemical conditioning with an adhesive can further enhance the RMGIC-RMGIC bond strength (Shaffer et al., 1998), as it can flow into surface irregularities and promote the micromechanical attachment to the underlying roughened cement, with chemical bonding ability to the exposed glass particles which is expected to enhance the bond strength over time. The modest bonding strength of GIC/RMGICs to different restorative interfaces added to the general weakness of the cements lead occasionally to fracture or wear with time (Wu and Smales, 2001; Maneenut et al., 2010) which could limit their use as a definitive restoration or to be used for long-term repair.

The use of a reparative system contains reactive polar groups which are capable of chelating to the inorganic crystal lattice (apatite) of the tooth from one side and anchoring mechanically and/or chemically to an existing restoration on the other side, is considered to have a clinical significance. Accordingly, a photoreactive phosphate-based monomer (ethylene glycol methacrylate phosphate, EGMP) was incorporated into a RMGIC system to improve its adhesion strength to resin composites, RMGIC/GIC's and amalgams by virtue of the polar phosphate groups. Based on the previously recoded findings, pRMGIC showed improved adhesion strength values to different tooth substrates (sound vs. diseased) in comparison to the control cement (detailed in Chapter four). In this chapter, the bonding efficacy of pRMGIC to different restorative interfaces was assessed to investigate whether the new cement can be used for long-term repair of failing TRCs.

Although in vitro investigations cannot be directly translated to in vivo performance, they can be considered as a useful predictor of the potential clinical application of the materials and allow comparison with the currently available products that are used for the same clinical purpose. There are many methods used to assess interfacial bond strength between similar/dissimilar restorative materials. Statically, a macro or micro-test can be used depending on the area of the tested interface. The macro-shear test was used due to the simplicity and popularity allowing comparison to the previously reported values in the literature. It reduces the risk of specimen damage due to the brittleness of the GICs in thin cross section and can be used only in bulk to avoid sample damage (Van Meerbeek et al., 2010; De Munck et al., 2010). Accordingly, this in vitro study aimed to evaluate the immediate and delayed shear bond strength (SBS) of pRMGIC to four conditioned substrates (amalgam, resin composite, RMGIC, and CGIC) with and without the use of adhesive bonding agent (Scotchbond™ Universal, 3M ESPE, USA). Results were compared to three commercial restorative grade materials; RMGIC (Fuji II LC (F2LC)) as a control group, CGIC (Fuji IX (F9)), and resin composite (Filtek™ Supreme (FS)), See table 5-1 for manufacturers' details.

The null hypotheses investigated in this study were**:**

1. There are no statistically significant differences in SBS values of the experimental cement (pRMGIC) from the control (F2LC) and the other applied materials (F9 and FS) within each restorative substrate (amalgam, resin composite, RMGICs and GIC) at each time interval.
2. There are no statistically significant differences in SBS values within each group with and without application of an adhesive.
3. There is no statistically significant difference between the early and delayed adhesion strength for each material per substrate.

5.2 Materials and Methods

Table 5-1 List of materials used in the study

Material	Manufacturer	Code	Material composition
Fuji II LC (RMGIC)	GC, Corp., Tokyo, Japan	F2LC	Fluoro-alumino-silicate glass, polyacrylic acid, HEMA, camphorquinone, water.
Fuji IX™ GP (CGIC)	GC Corp., Tokyo, Japan	F9	Fluoro-alumino-silicate glass, polyacrylic acid, Polybasic carboxylic acid.
Filtek™ Supreme XTE (universal restorative composite), (Shade A2B)	3M™ ESPE USA	FS	The resin-based matrix: bis-GMA, UDMA, TEGDMA, and bis-EMA resins. Fillers: non-agglomerated/non-aggregated 20 nm silica filler, a non-agglomerated/ non-aggregated 4-11 nm zirconia filler, and an aggregated zirconia/silica cluster filler (comprised of 20 nm silica and 4 to11 nm zirconia particles, with an average cluster particle size of 0.6 to 10 microns. Filler loading is about 78.5% by wt (63.3% by volume)
Scotchbond™Universal, (self-etch Adhesive)	3M, ESPE, USA	SU	MDP phosphate monomer, Dimethacrylate resins, HEMA, Vitrebond™ Copolymer, filler, ethanol, water initiators, silane
SDI, GS-80 Non-gamma 2, admix alloy			Alloy particles are spherical and lathe cut, it composes of 40% Ag, 31.3% Sn, 28.7% Cu, with 47.9% mercury Hg. Grey, regular setting, and the alloy to mercury ratio is 1/0.9.
Scotchbond™Etchant	Universal, 3M, ESPE, USA		37% phosphoric acid by weight

5.2.1 Sample preparation

An acrylic resin block (30 mm x 20 mm) was prepared using a stainless steel mould after the application of a separating medium. After setting, an 8.5 mm diameter and 4 mm deep recess was drilled centrally on one surface with a slight undercut using an inverted cone bur (Size 2.0, US No 38, Diaswiss S.A. Switzerland) in a slow-speed hand-piece to facilitate the mechanical retention of the repaired substrate. Multiple negative replicas for the prepared cavity were obtained using Vinyl Polysiloxane (VPS) impression material (Express STD, putty regular set heavy body 3M ESPE, Germany) which were then recast with acrylic resin to gain similar patterns of the prepared cavity for all repaired substrates (n=512).

The selected substrate materials (Amalgam, Filtek™ Supreme, Fuji II LC, Fuji IX GP), were dispensed, mixed and cured according to the manufacturers' instructions. Each cavity was overfilled by a selected material and covered with a glass slide that was firmly pressed against the restorative interface to get a smooth standardised surface for all specimens. Resin composite were placed in two layers (2 mm each) and photocured for 20 s per layer using a LED curing light (Elipar^{TM} DeepCure-S LED, 3M, USA) with a light output intensity of 1470 mW/cm^2. All surfaces were checked for the presence of a uniform surface without voids using a stereomicroscope (WILD M32; Heerbrugg, Switzerland) at x40 and then placed for an hour in an incubator at 37°C to allow setting of the substrates. After that, all specimens were stored in distilled water at 37°C for two months to simulate a clinically-aged scenario. Surface conditioning for all substrates was carried out to increase the surface area and enhance the interfacial bond strength between the old substrate and the bonded reparative materials (pRMGIC, F2LC, F9 and FS). This was achieved by the creation of retentive features using mechanical (roughening/ sandblasting), chemical (acid etching, adhesive agent) or a combination of mechanical and chemical approaches. Amalgam substrates were air abraded using An Aquacut™ air-abrasion unit (Velopex, Harlesden, UK) with a rounded nozzle (internal diameter 600 µm) that fixed by a micro-positioning device allowing standardised nozzle-substrate distance (10 mm) and angle (90°). Air borne particle abrasion was performed by 50 µm Al_2O_3 at a pressure of 60 psi bar for 4 s for each specimen (Blum et al., 2012). In contrast, the aged resin composite, RMGIC, and GIC surfaces were polished using a wet 600-grit silicon carbide disc (Struers RotoPol 11, Struers A/S, Rodovre, Denmark) for 60 s at 300 rpm to simulate the mechanical roughening that produced by a fine diamond bur (Anusavice, 1996). Then all specimens were cleaned for 10 min in an ultrasonic bath containing deionised water to eliminate any contaminants. Scotchbond™ etchant (37% phosphoric acid) was applied for 20 s to remove surface contamination and increase the reactivity of the underlying surfaces. Each conditioned substrate was divided randomly into two main groups (n=64, per group within each substrate). In the first group, Scotchbond™Universal bonding agent was applied over the conditioned surfaces and photocured for 10 s before the placement of the selected reparative materials. While in the second group, the reparative materials were placed directly on the conditioned surfaces without bonding agent. Each group was further subdivided in to four subgroups (n=16 per each) to receive four different reparative materials; pRMGIC (experimental), F2LC (control), F9 (CGIC), and FS (universal composite resin). The experimental procedures are represented in Figure 5-1.

5.2.2 Shear bond strength (SBS) test

The pRMGIC (30% by weight EGMP) was used as the experimental cement, while the commercial materials (Fuji II LC, Fuji IX™ GP and Filtek™ Supreme) were dispensed according to their manufacturer instructions. These materials were transferred into cylindrical silicon moulds (3 mm diameter x 4 mm height) (Tygon tubing, Saint-Gobain, USA) to ensure a reproducible amount of material over the conditioned surfaces. Light cured materials (pRMGIC, F2LC, and FS) were photo-polymerised for 40 s using a light curing device (Elipar™ DeepCure-S LED, 3M USA) with a light intensity of 1470 mW/cm^2. Samples were stored for an hour in an incubator at 37°C to allow setting of the applied materials. Shear bond strength (SBS) test was carried out at 24 h and the following three months' storage in simulated body fluid (SBF) at 37°C, (n=8 per subgroup). The solution was prepared following Kokubo and Takadamas' formula (Kokubo and Takadama, 2006). Solutions were replaced on a weekly basis. After storage, pre-test failures (PTFs) were recorded, then specimens were attached to the shear testing device using the universal testing machine (Instron® Model 5569, USA). The shearing blade was positioned in a way that the 'new' material was loaded in shear against the ‘old’ material, Fig (5-2). A shear force was applied at a crosshead speed of 0.5 mm/min until debonding occurred. The shear bond strength (t) was calculated in MPa using the equation $t=F/ (\pi R^2)$ where F was the applied load at failure and R was the radius of the material cylinder. Specimens which failed prior testing were assigned a notional SBS of 0 MPa.

5.2.3 Interface examination

After debonding, fractured surfaces were analysed using a stereomicroscope (WILD M32; Heerbrugg, Switzerland) at x40. Failure modes are classified into: (A) adhesive failure at the reparative material/ restorative interface; (B) cohesive failure within the reparative material; (C) cohesive failure within the restorative substrate; (D) mixed failures that are classified as a combination of partial interfacial adhesive and cohesive failures in either the applied material or the restorative substrate, Fig 5-3.

5.2.4 Statistical analysis

Descriptive statistics were used to summarise the bond strength characteristics of the selected reparative materials to different restorative surfaces after two-time intervals (24 h & 3 months). Percentages were used to present the failure modes. Data were tested for normality using Q-Q plots and Shapiro-Wilk tests and were analysed parametrically as the data followed a normal distribution. Multivariate Analysis of Variance (MANOVA) and Bonferroni HSD post hoc tests were employed to calculate the significant factors (p = 0.05) in SBS values among tested groups within all substrates at each time interval. After analysing the normality of data distribution, an independent t-test was used to determine the effects of storage time on the bond strength of each tested material per each substrate. All analyses were conducted using SPSS statistical package (version 24; SPSS Inc., IBM, Chicago, IL, USA).

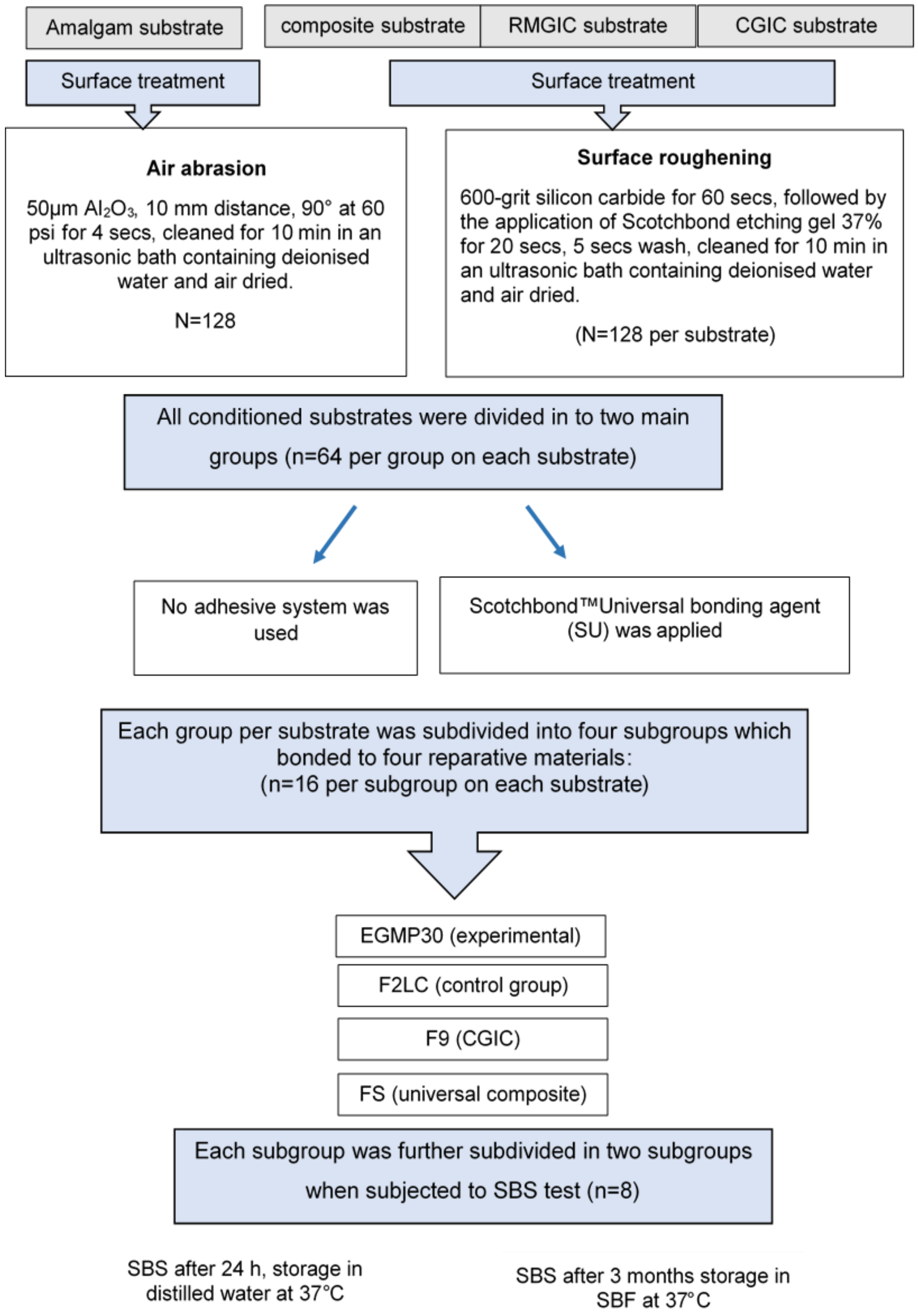


Figure 5-1 A schematic representing the experimental groups according to the experimental procedures.

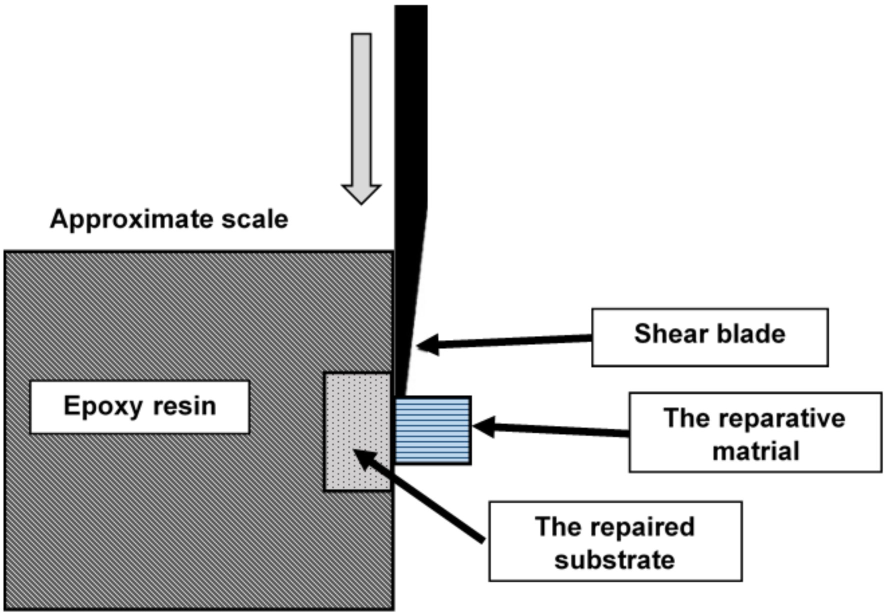


Figure 5-2 Diagrammatic representation of shear jig apparatus and shear force applied

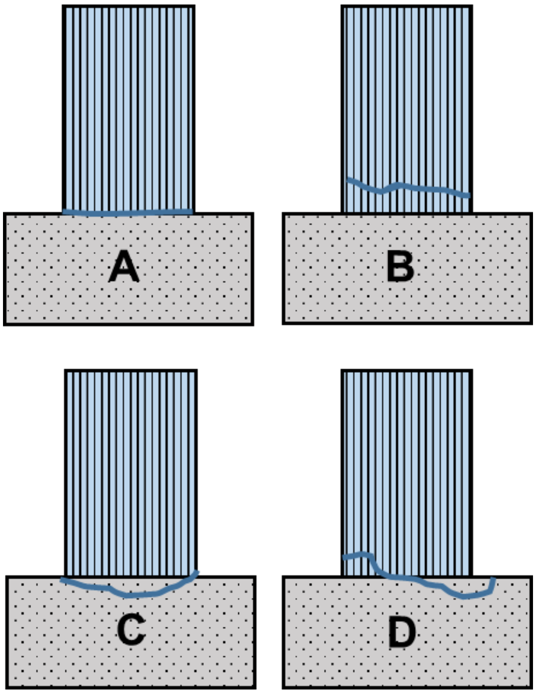


Figure 5-3 Classification of failure modes: (A) adhesive failure at the reparative material/ restorative interface; (B) cohesive failure within the reparative material; (C) cohesive failure within the restorative substrate; (D) mixed failures that are classified as a combination of adhesive and cohesive failures in either the applied material or the restorative substrate.

5.3 Results

5.3.1 SBS to amalgam substrate

The results of the SBS to amalgam substrate with and without an adhesive agent are presented in Table 5-2. There was no enhancement in the repair strength of pRMGIC when placed over an air-abraded amalgam substrate immediately and after three months' storage (2.9±0.5, 1.8±0.3 MPa, respectively) in comparison to the control F2LC (1.8±0.3, 1±0.4 MPa, respectively) (*p*=0.175, 0.295, respectively), which itself showed no significant differences from CGIC (F9) at both time points (*p*=1.000, 0.367, respectively). The resin composite (FS) exhibited the highest adhesion strength at both time intervals (6±1.4, 4± 0.8) (*p*<0.001).

The use of Scotchbond Universal adhesive intermediated pRMGIC and sandblasted amalgam substrate had a significant impact on enhancing the bond strength at both time periods (3.8±0.5, 5.2±0.5 MPa, respectively) as compared to F2LC (2.4±0.3, 1.1±0.4 MPa, respectively) (*p*=0.024, *p*<0.001, respectively), and its values without applying adhesive (*p*=0.029, *p*<0.001, respectively). Although, the adhesive strength of F2LC was enhanced at 24 h (*p*<0.001) when used with an adhesive, it was compromised post-ageing and being comparable to its value without adhesive (*p*=0.900). For GIC, the use of a bonding agent before F9 placement was not applicable, since all samples were detached immediately after cement setting (0 MPa).

For resin composite, initially, the presence of a bonding agent had no effect on SBS to a conditioned amalgam as compared to its value without adhesive (*p*=0.712), even though, it was the highest value among groups (6.3±1.2 MPa) (*p*<0.001). The SBS of resin composite was reduced after storage (5.2±0.5), however, it was higher than that without adhesive (*p*=0.030) with no significant difference from the delayed strength of the pRMGIC (*p*=1.000).

Ageing generally reduced the repair strength of all materials to this substrate whether the adhesive was applied or not (*p*<0.05), except the pRMGIC which showed improved bond strength when used with adhesive (*p*<0.001). Fracture analysis revealed a 100% adhesive failure of all reparative materials from sandblasted amalgam at both time periods regardless the presence of adhesive or not, except the pRMGIC group with adhesive which exhibited 50% mixed failures immediately with one sample failing cohesively after storage, Figure 5-4, and 5.

Table 5-2 Shear bond strength SBS (mean [SD]) to amalgam substrate with and without the use of an adhesive after 24 h and three months' ageing.

Groups **n=8 per group**	**Air-abrasion only**				**Air-abrasion + Adhesive SU**			
	SBS 24 h	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**	**SBS 24 h**	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**
pRMGIC *Experimental*	2.9 [0.5]^{a}	0	1.8 [0.3]$^{b\wedge}$	0	3.8 [0.5] $^{*+}$	0	5.2 [0.5]$^{*d\wedge+}$	0
F2LC *Control*	1.8 [0.3]^{a}	0	1.0 [0.4]^{bc} $^{\wedge}$	0	2.4 [0.3] $^{+}$	0	1.1 [0.4]$^{\wedge}$	0
F9 *(CGIC)*	1.7 [0.5]^{a}	0	0.4 [0.7]$^{c\wedge}$	4	0.0 [0.0]$^{+}$	8	0.0 [0.0]	8
FS *(Composite)*	6.0 [1.4]	0	4.2 [0.8]$^{\wedge}$	0	6.3 [1.2]	0	5.2 [0.5]$^{d+}$	0

(*) a statistically significant difference of the experimental cement (pRMGIC) from the control group. Similar letters in each column indicate no significant differences among groups (Bonferroni test post-hoc tests, an alpha level of 0.05). (^) a significant effect of ageing for the same group from the 24 h values with in each row (Independent t-test), ($^{+}$) significant statistical differences in values between bonding with and without the use of bonding (SU) agent, (n=8)

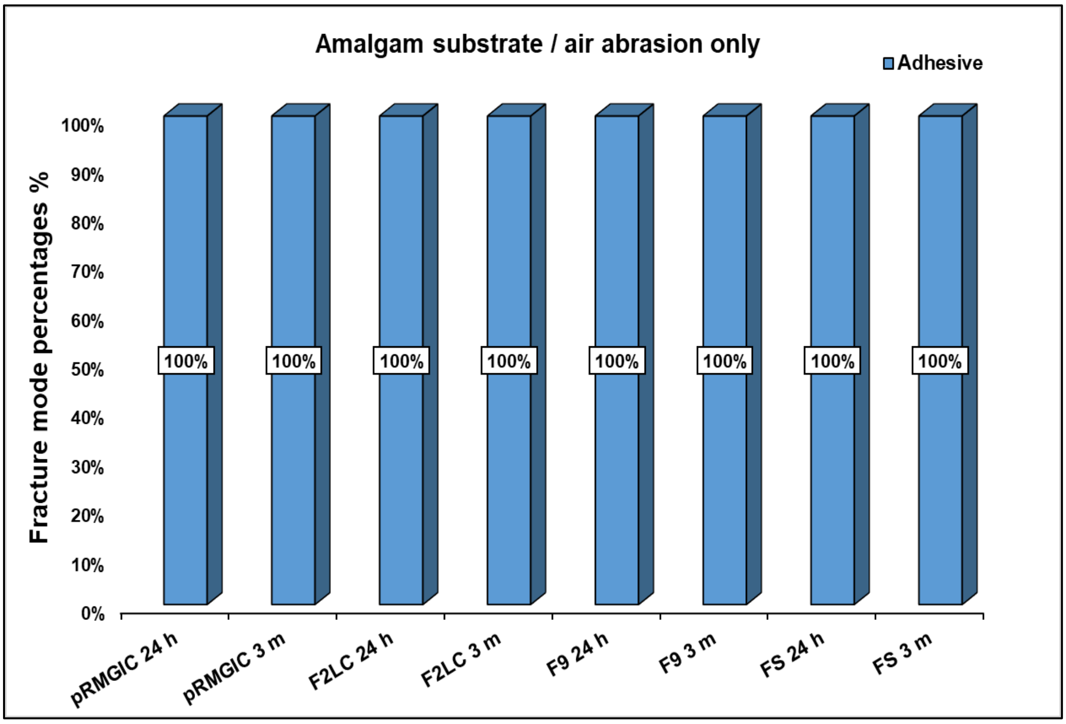


Figure 5-4 Mode of failure of each material from an air-abraded amalgam surface over time

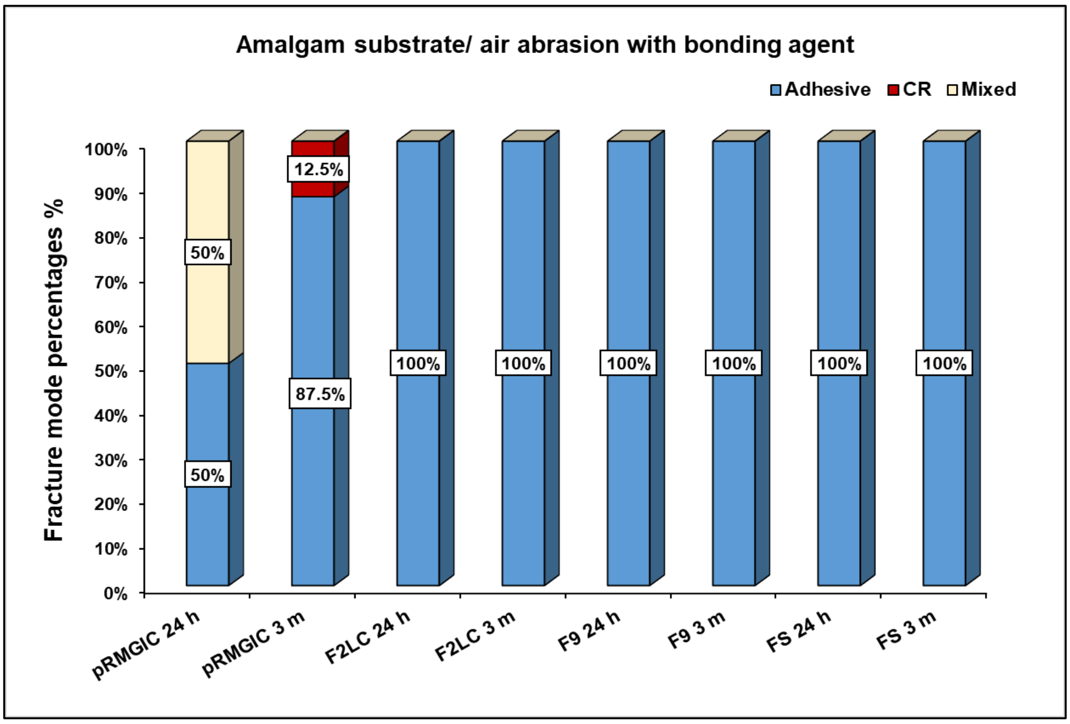


Figure 5-5 Mode of failure of each material from an air-abraded amalgam surface after an adhesive application over time

5.3.2 SBS to resin composite substrate

Early and delayed SBS values to the composite substrate with and without the use of adhesive are shown in Table 5-3. The effect of cement modification in composite repair was not apparent immediately as the initial SBS value was similar to the control (10.5, 10.6 MPa, respectively) (*p*= 1.000). However, on long-term, the enhancement in repair strength was evident and the delayed strength of pRMGIC was the highest among all materials (14±2.6 MPa) (*p*<0.001). F2LC showed no significant differences in SBS from the resin composite groups at both time periods (*p*=1.000), whist F9 recorded statistically significantly the least values in both intervals (*p*<0.001).

The use of adhesive prior to pRMGIC placement over a conditioned resin composite substrate was beneficial since the delayed bond strength of pRMGIC was the highest among all materials (17.3 MPa), however, it was statistically not significant from the control (*p*=0.357). Alternatively, FS which was recorded the highest initial value (16.8±1 MPa) (*p*<0.001) revealed a reduction in strength after ageing (11.2±1 MPa), it was even lower than RMGICs (experimental and control) (*p*<0.001). Nevertheless, the presence of adhesive was necessary at either interval as values were significantly higher than that without (*p*<0.001). Again on this substrate, F9 could not adhere to this substrate with adhesive resulting in 0 MPa SBS values recorded.

Ageing significantly improved the SBS of the pRMGIC to a resin composite substrate irrespective of the application of an adhesive (*p*<0.05), whilst the control showed enhanced strength only when used with an adhesive (*p*<0.001). Composite-composite adhesion strength was stable over time without bonding agent involvement (*p*=0.060), however, with adhesive, the adhesion was compromised post-ageing (*p*<0.05), even though, it was still higher than that without.

The early failure of the pRMGIC was entirely cohesive within the cement when placed directly on a conditioned composite substrate, after storage, cohesive failure was reduced to the half and replaced by mixed failure (37.5%). In contrast, the delayed failures for all groups were entirely adhesive (100%), (Fig 5-6). Moreover, failures in RMGICs are mixed and cohesive (62.5%, and 37.5%, respectively) when adhesive was applied, however, the delayed failure of pRMGIC was mostly cohesive (75%). The significant decrease in composite-composite adhesion strength over time with adhesive, accompanied a shift in mode of failure from mostly cohesive within the substrate (87.5%) to an absolute adhesive failure (62.5%), Figure 5-7.

Table 5-3 Shear bond strength SBS (mean [SD]) to the composite substrate with and without the use of adhesive after 24 h and three months' ageing

Groups **n=8**	**Surface roughening only**				**Roughening + Adhesive**			
	SBS 24 h	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**	**SBS 24 h**	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**
pRMGIC *Experimental*	10.5 [1.2]^{a}	0	14.2 [2.6]$^{*\wedge}$	0	7.7 [0.8]$^{*+}$	0	17.3 [2.4]$^{c\wedge}$	0
F2LC *Control*	10.6 [1.2]^{a}	0	6.9 [2.2]$^{b\wedge}$	0	6.3 [1.0]$^{+}$	0	15.5 [1.8]$^{c\wedge+}$	0
F9 *(CGIC)*	1.4 [0.4]	0	0.5 [0.6]$^{\wedge}$	4	0.0 [0.0]$^{+}$	8	0.0 [0.0]	8
FS *(Composite)*	9.9 [1.0]^{a}	0	8.4 [1.4]^{b}	0	16.8 [1.1]$^{+}$	0	11.2 [0.8]$^{\wedge+}$	0

(*) a statistically significant difference of the experimental cement (pRMGIC) from the control group. Similar letters in each column indicate no significant differences among groups (Bonferroni test post-hoc tests, an alpha level of 0.05). (^) a significant effect of ageing for the same group from the 24 h values with in each row (Independent t-test), ($^{+}$) significant statistical differences in values between bonding with and without the use of bonding (SU) agent, (n=8)

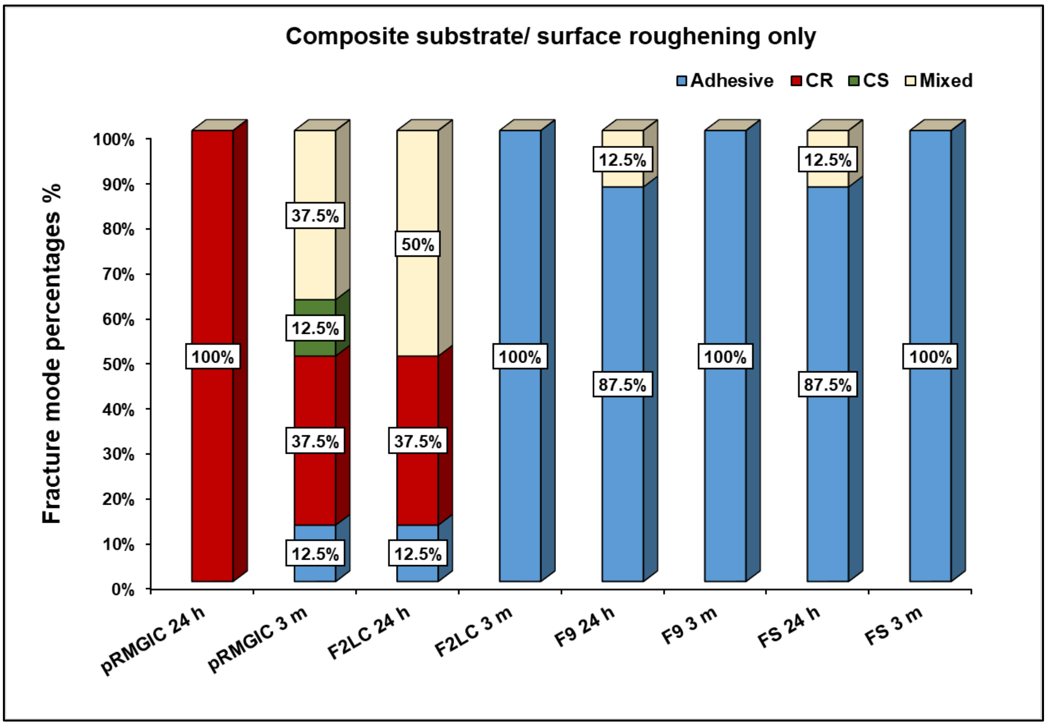


Figure 5-6 Mode of failure of each material from a roughened resin composite surface over time

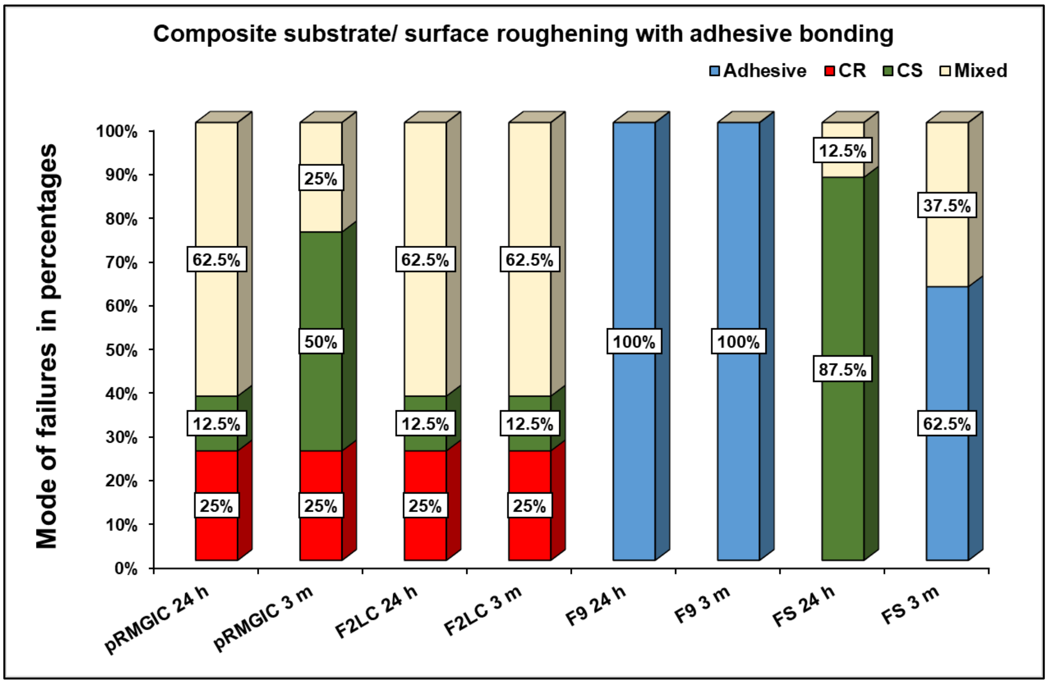


Figure 5-7 Mode of failure of each material from a roughened resin composite surface after adhesive application over time

5.3.3 SBS to RMGIC substrate

The shear bond strength values of the applied reparative materials to conditioned RMGIC substrates with and without using an adhesive are shown in table 5-4. The pRMGIC showed enhanced delayed bond strength to RMGIC substrate without adhesive involvement, as the mean value was statistically higher (11.7±1 MPa) than all applied materials ($p<0.001$). However, the initial value was comparable to F2LC and FS (≃ 5 MPa) ($p=1.000$). Conventional GIC recorded the lowest bond strength to this substrate ($p<0.05$).

pRMGIC showed enhanced SBS values to RMGIC substrate after bonding application at both time intervals (7.7±1, 16.5±1 MPa, respectively) in comparison to the control (4.4±1, 8.0±1 MPa, respectively) ($p<0.001$) and the values without bonding ($p<0.001$). In contrast, F2LC exhibited reduced SBS values when used with adhesive in comparison to values without adhesive at both time periods ($p=0.016$, 0.021, respectively). The GIC samples were detached immediately from this substrate when placed over an adhesive.

It was not surprising that the adhesive agent had a significant effect in improving the bond strength of resin composite group to aged RMGIC substrates immediately and after storage (8.2±1, 13.3±2 MPa, respectively) as compared to its values without (4.9±2, 6.8±1 MPa) ($p<0.001$).

The repair strength of the new cement was doubled after three months' storage in SBF at 37°C irrespective using an adhesive or not ($p<0.001$). However, the same trend was seen in both the F2LC and FS groups which showed a 40-80% increase in SBS over time, apart from the F9 group which lost part of its bond strength after storage when applied without adhesive. The two-fold increase in the SBS of pRMGIC over time was associated with 100% cohesive failures (cement/substrate) with and without adhesive. In contrast, this pattern of failure was prominent in F2LC and FS only when used with adhesive which showed more adhesive failures when placed directly on a roughened RMGIC substrate without a bonding agent, Figure 5-8, 9.

Table 5-4 Shear bond strength SBS (mean [SD]) to RMGIC substrate with and without the use of an adhesive after 24 h and three months' ageing

Groups **n=8**	**Surface roughening only**				**Roughening + adhesive**			
	SBS 24 h	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**	**SBS 24 h**	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**
pRMGIC *Experimental*	5.3 [0.4]^{a}	0	11.7 [1.0]$^{*\wedge}$	0	7.7 [0.8]*$^{b+}$	0	16.5 [1.3]$^{*\wedge+}$	0
F2LC *Control*	5.7 [0.8]^{a}	0	9.4 [1.1]$^{\wedge}$	0	4.4 [0.9]$^{+}$	0	8.0 [0.6]$^{\wedge+}$	0
F9 *(CGIC)*	2.6 [0.2]	0	2.0 [0.4]$^{\wedge}$	0	0.0 [0.0]$^{+}$	8	0.0 [0.0]$^{+}$	8
FS *(Composite)*	4.9 [1.6]^{a}	0	6.8 [1.0]$^{\wedge}$	0	8.2 [0.7]$^{b+}$	0	13.3 [1.8]$^{\wedge+}$	0

(*) a statistically significant difference of the experimental cement (pRMGIC) from the control group. Similar letters in each column indicate no significant differences among groups (Bonferroni test post-hoc tests, an alpha level of 0.05). (^) a significant effect of ageing for the same group from the 24 h values with in each row (Independent t-test), ($^{+}$) significant statistical differences in values between bonding with and without the use of bonding (SU) agent, (n=8)

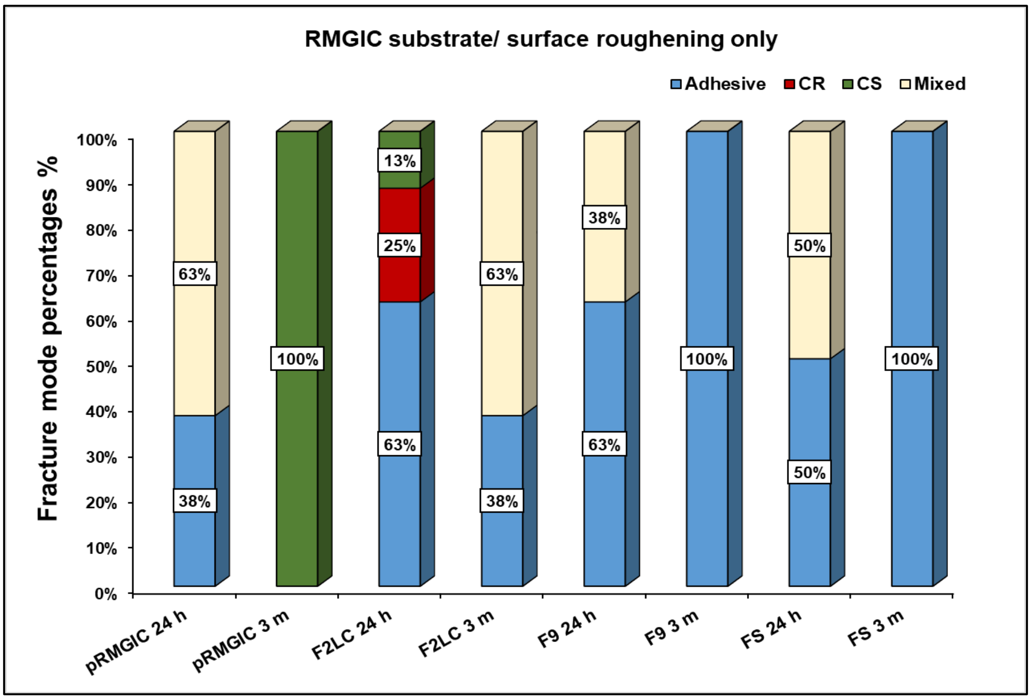


Figure 5-8 Mode of failure of each material from a roughened RMGIC substrate over time

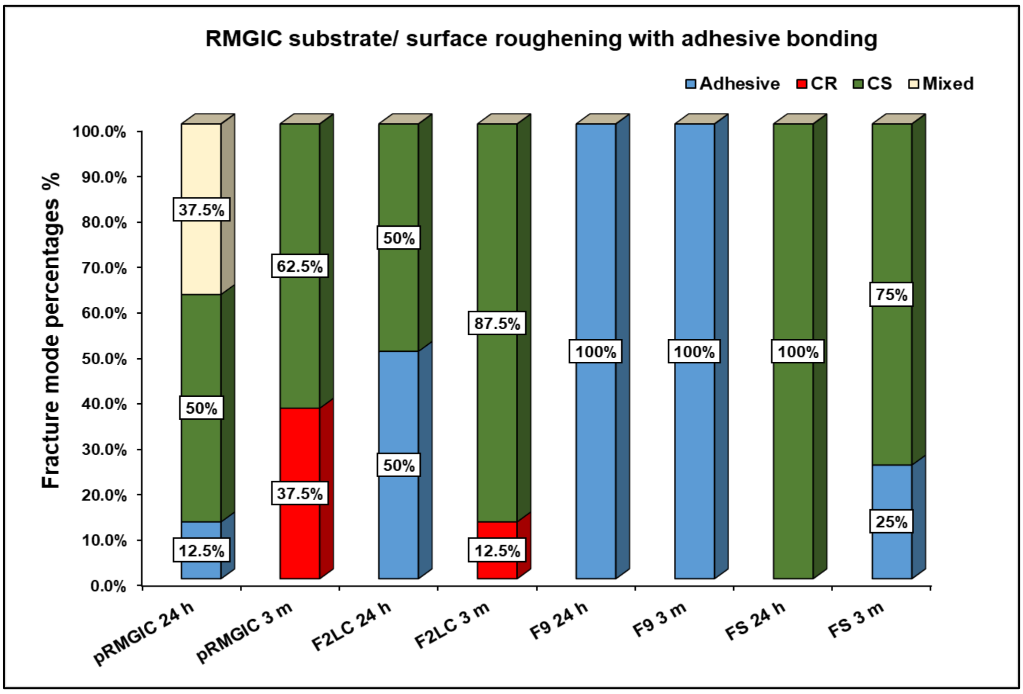


Figure 5-9 Mode of failure of each material from a roughened RMGIC after adhesive application over time

5.3.4 SBS to GIC substrate

The early and delayed shear bond strength to GIC substrate with and without the use of adhesive are presented in table 5-5.

pRMGIC showed statistically significantly higher early and delayed SBS values (6.7±1, 8.7±1 MPa, respectively) as compared to F2LC (5.4±1, 6.2±1 MPa, respectively) (*p*=0.024, *p*<0.001, respectively). F2LC recorded comparable values to resin composite (FS) at both time intervals (5.7±1, 6.9±1 MPa, respectively) (*p*=1.000, 0.734, respectively). Meanwhile, GIC-GIC repair showed the lowest values at both time periods (*p*<0.05).

The presence an adhesive intermediated the roughened GIC substrates and the reparative materials did not enhance the immediate and prolonged SBS with no significant differences among pRMGIC, F2LC, and FS after storage (*p*>0.05). However, the initial value of F2LC was significantly higher than that without adhesive (9.0±1, 5.4±1 MPa, respectively) (*p*<0.001) and that of pRMGIC, and FS (7.3±1, 6.4±1 MPa, respectively) (*p*=0.003, *p*<0.001, respectively). F9 failed to adhere to old GIC substrates when the adhesive was applied.

Ageing improved the bond strength of all reparative materials to roughened CGIC surfaces without adhesive. However, the enhancement was not statistically significant in F2LC (*p*=0.138). In contrast, ageing showed a variable influence on the SBS values when bonding agent was used. FS showed a better adhesion strength (*p*=0.010), F2LC showed lower strength (*p*=0.001), while pRMGIC maintain its strength post-ageing (*p*=0.204).

The adhesion strength of pRMGIC to GIC was higher than the cohesive strength of the substrate as manifested by higher percentages of cohesive failure within the substrate with and without using SU, while this pattern was noticed in F2LC and FS when they used in conjunction with bonding agent immediately and after storage. However, failures in FS group are mostly adhesive without applying an adhesive, Figure 5-10, 11.

Table 5-5 Shear bond strength SBS (mean [SD]) to GIC substrate with and without the use of an adhesive after 24 h and three months' ageing

Groups **n=8**	**Surface roughening only**				**Roughening + adhesive**			
	SBS 24 h	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**	**SBS 24 h**	**PTFs 24 h**	**SBS 3 m**	**PTFs 3 m**
pRMGIC *Experimental*	6.7 [0.8]^{*a}	0	8.7 [0.7]$^{*\wedge}$	0	7.3 [0.7]^{*d}	0	8.0 [1.0]^{e}	0
F2LC *Control*	5.4 [0.8]^{b}	0	6.2 [0.9]^{c}	0	9.0 [0.9]$^{+}$	0	6.9 [0.7]$^{e\wedge}$	0
F9 *(CGIC)*	3.1 [0.4]	0	4.8 [0.9]$^{\wedge}$	0	0.0 [0.0]$^{+}$	8	0.0 [0.0]$^{+}$	8
FS *(Composite)*	5.7 [0.7]^{ab}	0	6.9 [0.7]$^{c\wedge}$	0	6.4 [0.7]^{d}	0	7.9 [0.9]$^{e\wedge}$	0

(*) a statistically significant difference of the experimental cement (pRMGIC) from the control group. Similar letters in each column indicate no significant differences among groups (Bonferroni test post-hoc tests, alpha level of 0.05). (^) a significant effect of ageing for the same group from the 24 h values with in each row (Independent t-test), ($^{+}$) significant statistical differences in values between bonding with and withous the use of bonding (SU) agent, (n=8)

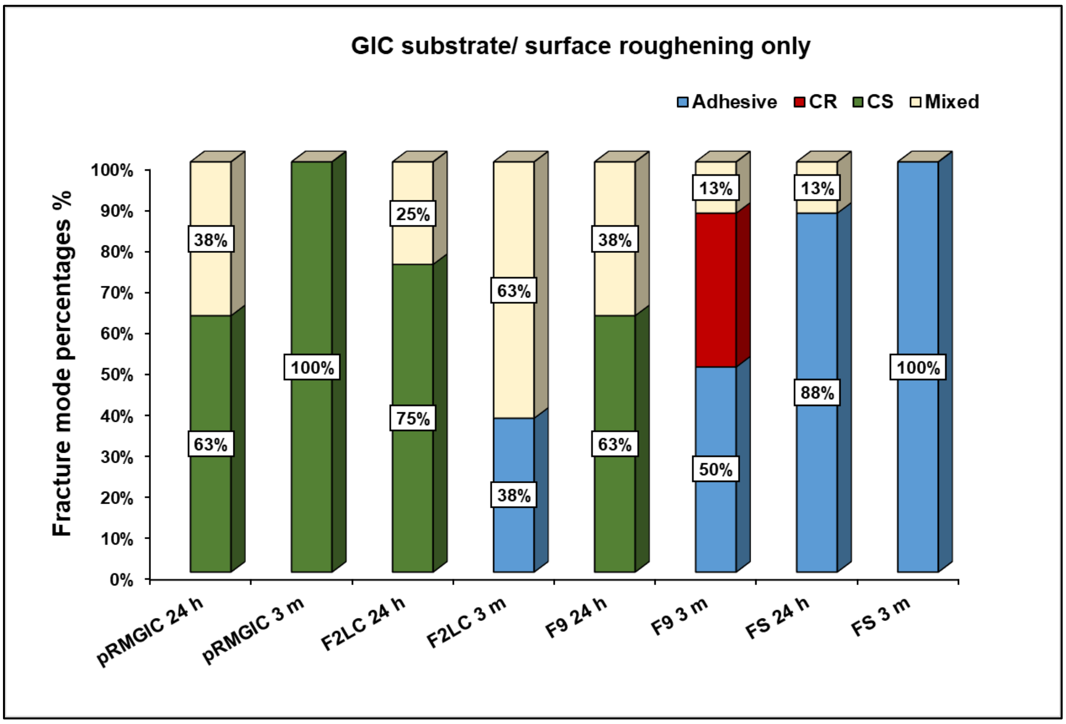


Figure 5-10 Mode of failure of each material from a roughened GIC substrate over time

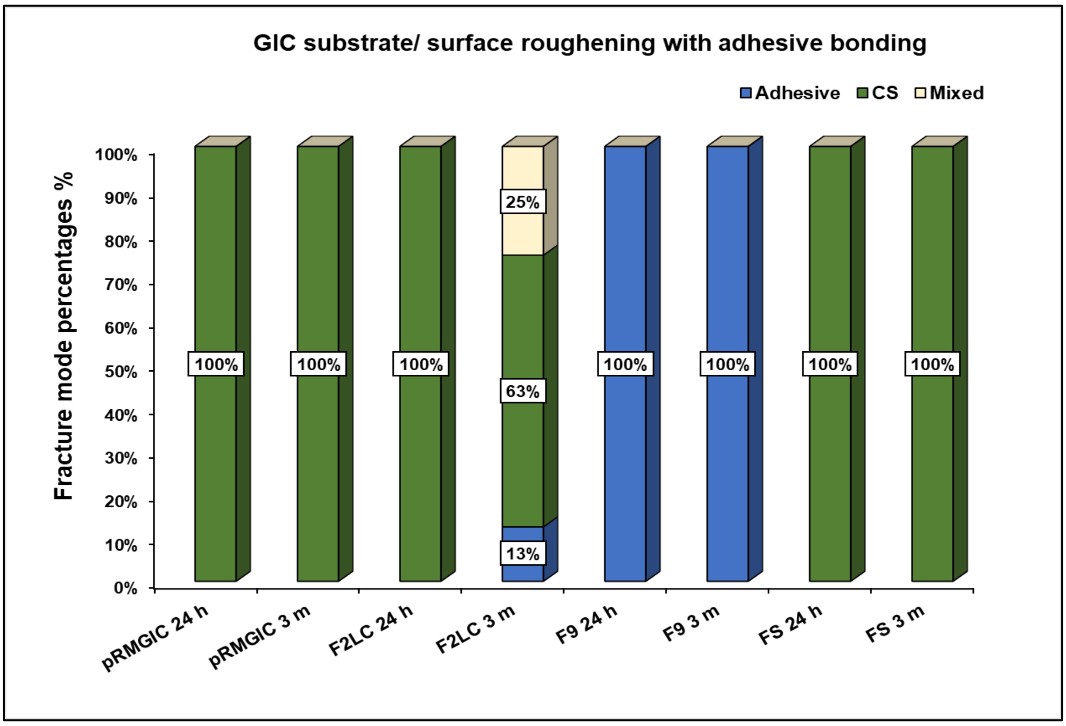


Figure 5-11 Mode of failure of each material from a roughened GIC after adhesive application over time

5.4 Discussion

5.4.1 SBS to amalgam substrate

Surface roughening of aged amalgams were carried out by air-abrasion to create micro retentive features for better interfacial bond strength. The comparatively mild mechanical surface treatment using alumina yields a uniform roughened surface free from major asperities, defects and stress concentrations. In contrast, excessive mechanical roughening produced by burs might induce surface defects and areas of stress concentrations that initiate crack propagation at the interface and interfere with the full penetration of the adhesive agent leading to a weaker interfacial bond strength as previously reported by Machado et al., (2007) and Blum et al., (2012).

The presence of a polymerisable functional monomer within the RMGIC is supposed to improve the chemical and/or physical adhesion to a roughened amalgam surface by virtue of the polar phosphate moieties. However, such a property was not observed when the cement was placed directly over the substrate without an adhesive intermediated the approximate interfaces. The low initial SBS value of pRMGIC ($\simeq$3 MPa) was declined after storage to 2 MPa, and being comparable to that of F2LC (p=0.175, p=0.295, respectively) with entirely adhesive failure for either group, Fig 5-4. This might indicate that the adhesion of the pRMGIC to a roughened amalgam is based on mechanical interlocking and an adherence mediated by Van der Waals forces rather than a possibility of chemical interactions with the substrate. This is agrees with studies by Pilo et al. (1996) and Pilo et al. (2012) who doubted the existence of a chemical bond between the amalgam and resin-based restorations. They reported higher adhesion strength of RMGIC over time when intermediated a composite resin and air-abraded amalgam substrate.

The first, second and third hypotheses were rejected when pRMGIC was placed over amalgam surface conditioned mechanically and chemically with SU adhesive as it shows a significant enhancement in early SBS in comparison to the control and its value without bonding agent (p<0.05). This might be attributed to the low viscosity of the bonding agent which enhances surface wetting enabling better infiltration with improved physical interlocking into the micro-retentive amalgam surface. The repair strength of the pRMGIC was further enhanced after three months' storage and was similar to that of the resin composite group (≃5 MPa) (*p*=1.000). This might be attributed to compatibility in the resin components of the pRMGIC and the adhesive agent that contains phosphate ester monomer (10-MDP) which are covalently copolymerised with the monomer through its long carbon chain, and bonds chemically to the positively charged metallic ions at the alloy surface (Souza et al., 1994) producing high SBS over time. The enhanced strength of pRMGIC after storage was associated with a shift in the mode of failure from entirely adhesive to 50% adhesive when combined with an adhesive, Fig. 5-5. In contrast, F2LC showed high early strength when used with a bonding agent, but it lost more than half of its initial strength due water absorption and hydrolytic degradation as suggested by Beriat and Nalbant, (2009). The delayed adhesion strength of F2LC to amalgam with and without adhesive was the same (≃1 MPa) which might suggest that the presence of an intermediary is unnecessary. This fact coincides with an early study of Aboush and Elderton, (1991) who suggested that the use of a bonding agent for RMGIC-amalgam repair compromises the bond strength as it produces weak linkage between joint interfaces. Former studies (Warren, and Söderholm, 1988; Aboush and Jenkins, 1989) suggested the possibility of repairing clinically ditched amalgam restorations with GIC due to the affinity of carboxyl group anions of polyacrylic acid to chemically interact with the metallic cations, especially tin and silver (Negm et al., 1982). Although the initial SBS of F9 was comparable to that of F2LC ≃2 MPa (*p*=1.000), with time the bond strength was severely compromised (0.4 MPa) with half of the samples were failed before testing (PTFs=4), which indicates that F9 might attach mechanically to a roughened substrate without any possibility of a chemical union to the substrate. These findings agree with Pilo et al., (2012) who excluded GIC from the study since all specimens debonded from the conditioned amalgam substrate spontaneously on prolonged ageing.

The use of adhesive with resin composites in amalgam repair has been preferred clinically as it enhances the integrity of the amalgam-resin composite interface (Matsumura et al., 1997). The reported SBS values in the literature ranged from 3.0 - 7.5 MPa when universal resin composites used in conjunction with MDP-containing adhesive (Zachrisson et al., 1995, Blum et al., 1996; Blum et al. 2012).

In the present study, the effect of using an adhesive was not noticed immediately (6 MPa) (*p*=1.000). Hence, the micromechanical retention of resin composite to sandblasted amalgam is considered the most likely mechanism of adhesion without bonding agent. Thereby, it may not able to counteract the hydrolytic degradation after storage which compromised the delayed bond strength. The use of an adhesive contains 10-MDP with an affinity to interact chemically to the metallic ions at the alloy surface (Souza et al., 1994) produced stable SBS over time (5 MPa). In accordance with Blum et al., 2012, the interfacial failure between amalgam and resin composite was adhesive, irrespective of the repair protocol employed.

5.4.2 SBS to resin composite substrate

Repairing aged resin composite is considered a challenge due to firstly the lack of the oxygen inhibition layer of polymerised resin which dramatically influences the chemical bonding to old resin composite (Papacchini et al., 2007c). Secondly, the surface changes associated with water sorption including matrix softening, microcracks, resin degradation and filler debonding (Ferracane, 2006), are expected to compromise the micromechanical interlocking to this substrate. Surface roughening of aged resin composite was carried out in this study using 600 grit SiC paper for 60 s to mimic the clinical mechanical roughening obtained by fine diamond bur (Anusavice, 1996). This was done to eliminate the superficial layer altered by water storage exposing a clean higher energy resin composite surface, and increase surface area through the creation of surface irregularities to promote micro-mechanical interlocking into the composite substrate. Then, surfaces were acid etched for 60 s using 37% phosphoric acid to remove the surface contamination exposing the underlying surface and fillers at the resin composite surface, as its contribution to the micromechanical adhesion with resin composites has not been proven (Bonstein et al., 2005; Loomans et al., 2011a).

Without adhesive, the initial bond strength of RMGICs (pRMGIC and F2LC) to resin composite was similar to composite-composite bond strength. This findings might be attributed to the chemical affinity of the resinous components of both materials (Jianguo et al., 1996) and the hydration of aged composite surfaces that enhances the penetration of the hydrophilic adhesive systems (Teixeira et al., 2005). The initial hydrophilicity of RMGICs facilitates the infiltration into the irregularities created by surface roughening, producing an initial strength (10 MPa) that was significantly higher than that with adhesive. This coincides with previous findings which revealed the use of identical resin matrix chemistry for composite repair is not prerequisite for higher bond strengths than those of different matrix chemistry (Gregory et al., 1990; Papacchini et al., 2007a).

Functionalisation of RMGIC showed an enhanced bond strength to resin composite substrates after three months' storage (14 MPa) (*p*=0.011) which might be due to cement maturation. The higher bond strength of pRMGIC than that of F2LC and FS (7, 8 MPa, respectively) (*p*=0.001) was associated with predominant cohesive and mixed failures than entirely adhesive in both groups, Figure 5-6. The presence of a photoreactive monomer with pendant phosphate groups may not only create a network of covalently linked phosphate groups that enhanced the cohesive strength of the cement as previously demonstrated (Chapter 3), but additionally can bond to the cross-linked polymer matrix and filler particles of composite by virtue of the polar phosphate groups producing durable bond higher than the cohesive strength of the cement.

In agreement with the literature (Özcan et al., 2005; Teixeira et al. 2005; Cavalcanti et al., 2007; Loomans et al., 2011b; Staxrud and Dahl, 2011) the use of adhesive following surface roughening produces predictable bond strength in resin composite repair. In the present study, composite-composite intermediated by SU produced the highest early repair strength (16.8 MPa) in comparison to their values without bonding agent (*p*<0.001) and the RMGICs (*p*<0.001) with 88% cohesive failures within the substrate, Figure (5-7). Initial hydrophilicity and lower viscosity of SU enhance wettability that promote monomer penetration into microcracks thus enhancing the micromechanical retention (Andrade et al., 2017), added to the possibility of chemical coupling with the aged resin matrix or the exposed filler particles (Kallio et al., 2001).

However, in term of bond durability, and in concurrence with previous studies (Papacchini et al., 2006b; da Costa et al., 2012), composite-composite repair strength was reduced to 11.2 MPa after storage. This is mainly due to water sorption and the hydrolytic degradation of the resinous polymer at the adhesive-resin interface as it was associated with higher percentage of adhesive failure

(63%). Nevertheless, it is still recommended to use a bonding agent as an intermediary in composite-composite repair. In contrast, RMGICs produced more than two-fold increase in bond strength after ageing (17, 15 MPa, respectively) which match the clinically accepted repair strength values for resin composite (15-25 MPa) (Teixeria et al., 2005; Maneenut et al., 2011) associated with cohesive and mixed failures. This is mainly attributed to cement maturation as the initial setting is based on methacrylate crosslinking after exposure to light, but the slower acid-base reaction hardens and strengthens the cement after ageing due to the formation of insoluble polysalts matrix (Tolidis et al., 1998). These findings are in agreement with Pilo et al., 2012 which reported a significantly enhanced bond strength of RMGIC from 48 h to 3 months and remained stable for the six months period. Furthermore, there is a possibility of chemical linkage between the inorganic portion of the substrate and the organic matrix of the RMGIC (Trajtenberg and Powers, 2004) induced by 10-MDP in the SU, as it contains two main functional chemical groups; the phosphate group which might bond to the silica of the composite filler and the organofunctional groups which co-polymerize to the methacrylate of the RMGICs. Moreover, the presence of phosphate groups in both pRMGIC and SU within a moisturised environment increase the reactivity of those groups to chelate within the cement matrix enhancing the cement's strength in one hand and promote its bonding to the substrate in another hand. Thus a robust, durable repair strength was achieved by pRMGIC as compared to all tested groups with 75% cohesive failures.

When comparing the SBS of the RMGICs and GIC bonded to resin composite, the values were consistently higher in RMGICs. This is supported by the literature (Mitra & Kedrowski, 1994; Jianguo et al., 1996; Hashem et al., 2014) and mainly correlated to the higher cohesive strength of the RMGICs, measured by flexure strength, added to the presence of HEMA in RMGICs which would promote chemical adhesion with resin composites (Chadwick and Woolford, 1993). In contrast, the lack of chemical union between GIC and resin composite indicates that the measured SBS of GIC may reflect the degree of micromechanical attachment only (Chadwick and Woolford, 1993). Although, the minimum bond strength for an effective resin composite repair is not reported but the repair strength of F9 is considered to be unsatisfactory as it exhibited very low early mean value (1 MPa) which further deteriorated after storage to 0.5 MPa with half of the samples being lost before testing associated with high percentage of adhesive failure. This coincides with the fact that early exposure of the cement to moisture during cement setting leads to a deterioration in the physical properties of the cement in which the erosion and plasticising effects overcome the additional crosslinking and build-up of silica gel phase (Kilpatrick et al., 1994; Cattani-Lorente et al., 1994). Accordingly, the comparatively low SBS and the spontaneous debonding of specimens before testing rendering the F9 unsuitable for repairing this substrate.

5.4.3 SBS to RMGIC substrate

Surface roughening of aged Fuji II LC substrates using a 600-grit Si-C paper was carried out to simulate the roughening effect of a diamond bur, followed by acid treatment to enhance the mechanical keying of a fresh material. The mechanical roughening was previously reported to enhance the repair strength and modify the fracture pattern to totally cohesive with no significant effect of acid etching under SEM (Yap et al., 1998; Maneenut et al., 2010; and Camilleri et al., 2013).

Initially, pRMGIC, F2LC and FS exhibited similar SBS values (5 MPa) when they applied without adhesive. The constituent resin component offers the potential for chemical adhesion to the underlying RMGIC, added to the micromechanical entanglement with the roughened surfaces (Rusz et al., 1992). This leads to significantly increased SBS values for all materials after storage (40% in FS, 65% in F2LC and 120% in pRMGIC). The double increase in the bond strength of pRMGIC (11.7 MPa) with 100% cohesive failure within the substrate might be due to the structural changes in the cement post maturation. The polarity of the phosphate group attracts more water to the matrix, which encourages more ionic interactions within the cement over time. Furthermore, the slower acid-base reaction of pRMGIC as manifested by prolonged working time, as shown in chapter three, within a moisturised environment encourages the reactive phosphate groups and the polyacrylic acid anions to chemically integrate with the exposed glass particles of the substrate which leads to a robust and durable bond strength that was limited by the cohesive strength of the substrate. These interactions were supported by McLean, (1992), and Yap et al. (2000) who suggested that the exposed glass particles in the ‘old’ GIC material could react with the polyacrylic acid in the ‘new’ material and establish a chemical bond.

Nicholson et al. (1992) hypothesised that bond strength of F2LC would be affected adversely by water storage due to water uptake by the hydrogel matrix, leading to increased plasticity and a reduction in strength over time associated with a predominant of adhesive failures. In fact, water absorption would affect the properties of the RMGIC at the early stages as the absorption occurs quickly and this also could be seen in the present study. However, water is also essential for the progression of the acid-base reaction which strengthens the cement over time with more ionic interactions, and polysalts formation leads to higher delayed strength for both RMGICs.

There is no doubt that the application of Scotchbond Universal prior to resin composite improved the bond strength to RMGIC substrates. The adhesive flows into surface irregularities promoting the micromechanical attachment to the underlying roughened cement. Additionally, the reactivity of 10-MDP monomer to chemically bond with the exposed glass particles produces further enhancement in bond strength after storage associated with 75% cohesive failure (Fig 5-9). pRMGIC also showed enhanced immediate and delayed bond strength values when applied with an adhesive with a higher percentage of cohesive failure, presumably due to their common resin components. The double increase in SBS values of pRMGIC (8, 17 MPa, respectively) in comparison to the control (4, 8 MPa) ($p<0.001$) is referred to the presence of pendant phosphate groups in SU (10- MDP) and pRMGIC (EGMP). Since under moist circumstances, these functional groups induce a zone of ionic interaction within the cement matrix and with the repaired substrate which produces higher bond strength over time. However, the delayed failures in pRMGIC are entirely cohesive irrespective of using bonding agent or not. In F2LC, in term of bond strength values, the presence of an adhesive seems not to be beneficial, while it is of interest to note the marked difference observed in the delayed failure mode of the cement which displayed a shift from mixed and adhesive failures without bonding agent to 100% cohesive when bonding was applied, which indicates a possibility that the bond strength constrained by the cohesive strength of the repaired cement (Sneed and Looper, 1985). This agrees with Welch et al. (2015) who revealed the importance of using a bonding agent on the micromechanical bond strength in RMGIC repair.

In agreement with previous work (Holtan et al., 1990), the bond strength of GIC is significantly lower than RMGIC. This is mainly attributed to the slow setting of GIC, sensitivity to clinical conditions and the low cohesive strength of the cement in which failure occurs cohesively within the cement before debonding (Sneed and Looper, 1985). The further reduction in strength of F9 after storage with a higher percentage of adhesive failure (62.5%) is due to the erosion and plasticising effects of water which overcomes the additional crosslinking and build-up of silica gel phase (Cattani-Lorente et al., 1994). The poor dependability of F9 when placed alone and its incompatibility to be used with adhesive rendering the material unsuitable for repairing failed RMGIC.

5.4.4 SBS to GIC substrate

Surface roughening followed by phosphoric acid application was accomplished to promote an effective seal to aged GIC substrates. However, the precise role of etching GIC surfaces in relation to retention and seal is unclear (Woolford, 1993), as it appears crucial in one group of studies (McLean et al.,1985; Suliman et al.,1989) while others indicate that it is not only unnecessary but also undesirable (Papagiannoulis et al., 1990; Taggart and Pearson, 1991). Generally, phosphoric acid etching for 20 s can modify the aged cement's surface for adhesion to the resin-based materials as previously suggested by Mount, (1989), without alteration in the chemical and or physical properties of the cement as shown by Camilleri et al. (2013).

The enhanced strength of pRMGIC, as shown in chapter 3, might contribute to the improvement in its bonding efficacy when applied directly over GIC surfaces without bonding agent in comparison to the control in both time periods (p=0.024, p<0.001, respectively). This coincides with other reports (Mitra and Kedrowski, 1994; Jianguo et al., 1996, Mitra et al., 2009) that suggested the higher cohesive strength of GIC produces a better adhesion strength. Furthermore, the increased hydrophilicity of pRMGIC due to the acid-functional groups enhances surface wetting and contributes in enhancing the adhesion strength (Mount, 1989). This moist environment triggers more ionic interactions by the reactive phosphate groups to the exposed alkaline glass particles in F9 substrate which presumed to produce a chemical union with the substrate that leads to further development in strength after storage (p=0.001) confirmed by entirely cohesive failure within the substrate. The adhesion of resin-based materials to GIC relies on the micromechanical attachment which can be further modified by using adhesive systems. This is because of the lower viscosity and lower contact angle with a greater adaptation of the adhesive over an etched cement surface which increase the wettability of the cement surface to resin, thus producing a better seal (Mount 1989; Hinoura et al., 1989; Wooford and Grieve, 1993) with reliable bond strength which is limited by the low cohesive strength of the GIC (Sneed and Looper, 1985; Hinoura et al., 1989). In line with the literature, the bond strength of pRMGIC, F2LC, and FS to GIC were higher than the tensile strength of the substrate itself. However, no superior SBS values were shown as compared to their values without adhesive, but all failures are cohesive within the GIC substrate. This masks the true bond strength values for these materials. This trend is due to the low mechanical strength of GIC, the non-uniform stress distribution concentrating on the cement and the microporosities within the cement which itself acting as potential stress points leading to its premature failure before the interface itself (Placido et al., 2007, Zhang et al ., 2011).

Mild etching for a short time may cause loss of matrix around the crystalline structures exposing the glass fillers and hence encouraging successful adhesion through micro-mechanical retention. In addition to the potential chemical interaction between the polyacrylic acid of the freshly mixed GIC and the exposed glass particles of the aged GIC (Pearson et al., 1989) which produced a reliable SBS at the early term with high percentages of cohesive and mixed failures. The changes within GIC over time are a complex phenomenon. In some instances, an improvement in cements' mechanical strength is occurred due to maturation which produces an enhanced interfacial bond strength (Jianguo et al., 1996), as noted in the present study (5 MPa). However, there is also a possibility of simultaneous weakening of the cement occurring via the plasticising effects of water (Pearson et al., 1989). Furthermore, failure of the self-adhesive bonding agent to bond old to new GICs was previously reported by Zhang et al., (2011). The incompatibility in chemistry and setting reactions of the CGIC and bonding agent make it acts as a separation layer which interferes with the chemical and mechanical interlocking with the substrate.

5.5 Limitation and future work

Shear bond strength was used to assess the bonding efficacy of the phosphorylated RMGIC to different restorative substrates to be used for repairing failed TRCs. This test is widely used for the assessment of the interfacial bond strength between similar/dissimilar restorative materials due to the simplicity and popularity allowing the comparison of the results to the previously reported values. The large bonded surface area reduces the risk of specimen damage due to the brittleness of the GICs (Van Meerbeek et al., 2010; De Munck et al., 2010). The major criticism for this method is that it does not represent the real adhesion strength due to many reasons such as the variations in the chemical and physical properties of the different repaired substrate-reparative materials. This testing mode may also induce non-uniform stress distribution that might concentrate near the interface, which leads to premature failure in the weaker material before the adhesive debonding. This is clearly identified in groups that were repaired the GIC substrates as most failures occurred within the substrate, which masks the actual bond strength of the reparative materials. Moreover, the larger bonded surface areas make the true interfacial testing, whether in shear or tension, to be as a cleavage test as soon as the first crack begins to propagate from defect, void or other source of stress concentration. Accordingly, the conventional bond strength testing method is no longer very useful in detecting the true adhesive bond strength, however a comparison of different groups under the same testing conditions provides a good indication of the differences. However, these variations between dissimilar interfaces cannot be avoided intraorally. There are some recent approaches are proposed to be alternative effective tools in evaluating the bond strength which modifies the conventional SBS settings. These are the mold-enclosed shear bond strength (ME-SBS) (Van Meerbeek et al., 2010; Cheetham et al., 2014) and a lever-induced mold-enclosed shear bond strength (*de novo* LIME-SBS) (Jin et al., 2016). However, more studies are required to evaluate the efficacy and validity of these techniques.

The other limitation is related to the hydrophilicity of the new cement, which promotes adhesion when combined with the adhesive bonding agent that contains 10-MDP after three months’ ageing. This moisturised environment increases the reactivity of these functional groups to chelate within the cement matrix enhancing the cement’s strength in one hand and promote its chemical bonding to the substrates on the other hand. However, the higher hydrophilicity of this cement may or may not also influence the hydrolysis of this bonding, which needs to be investigated in future by testing the bond strength after long-term storage, thermocycling, and the use of cyclic loading which mimic the oral environment.

5.6 Conclusions

Within the limitation of the present study it was concluded that:

1. The stated hypotheses were rejected when pRMGIC treated conditioned amalgam unless being used with adhesive.
2. pRMGIC was effective for resin composite repair up to three months, whether used with or without adhesive, hence the proposed hypotheses were rejected for the three months interval only.
3. Although the stated hypotheses were rejected at the 3 months study whether adhesive is applied or not, a preference of SU adhesive with the pRMGIC was noted.
4. In GIC repair, the first and third hypotheses were rejected only when pRMGIC was placed without adhesive. Although the SBS values did not exhibit an improvement when pRMGIC used with an adhesive, all failures are entirely cohesive within the GIC substrate.

Chapter six

6 General discussion, conclusions and suggestions for future work

6.1 Summary

The concept of resealing a failing restoration falls within the minimally invasive dentistry philosophy. Marginal defects, discolouration, degradation/wear/gap formation and CARS are clinical features of a failing tooth- restoration complex. Repair provides a less complex restorative therapy, as well as conserving tooth structure. It reduces the potential harmful effects on the dentine-pulp complex and consequently increases the longevity of the tooth-restoration complex.

The review of relevant background literature identified that repairing failed restorations with adhesive resin restorations is a treatment of choice, which conserves remaining tooth structure and provides the most reliable results (Fernández et al., 2015; Estay et al., 2018). Although laboratory studies showed favourable physical properties of the adhesive resin-based restorations, the longevity still considered as a challenge. This is attributed to hydrolytic degradation of the resinous components which deteriorate the properties over time (Ferracane and Marker, 1992; Ferracane, 2006). Additionally, the inherent volumetric reduction post polymerisation that develops stresses which may compromise the synergism at the restoration-tooth interface. (Davidson et al., 1984; Atai and Watts, 2006). Clinically, wide range of failures have been documented in the literature (Davidson et al., 1984; Jokstad et al. 2016), indicating the limitations of the currently utilised materials. Thus, the overall aim of this research was to develop, characterise, and assess newer reparative materials for effective prolonged repair of failed TRCs.

The initial experiments highlighted the differences in the physical properties of manually-mixed commercial GIC/RMGIC from the mechanical-mixed versions where clinically induced variability is excluded. This experiment helped in understating the properties of different GIC systems and standardising the cement handling and dispensing, taking in to account cement manipulation that was performed by a single operator using fixed amount of powder and liquid to reduce variation and standardise the cement preparation and the resultant mechanical properties for all formulations that were carried out in the study.

In agreement with the literature, mechanical-mixing reduced the porosity with more thorough wetting of the powder particles, which yield cements with enhanced physical properties in comparison to the equivalent manually-mixed versions (Nomoto et al., 2004; Dowling and Fleming, 2008; Dowling and Fleming, 2009; Molina et al., 2013).

This study also explored the strengthening effect of including nano-sized reactive glass particles that enhance the immediate and short-term mechanical properties of conventional GICs, as shown in Equia® Forte Fil. However, the presence of <6% hydroxyapatite/fluorapatite (HAp/FAp) nanoparticles and liquid silica within the GIC matrix (glass carbomer cement) disrupted the cement forming process and resulted in inferior mechanical properties (CS, CM, MH and BFS). This finding is well supported by previous studies (Yamamoto, 1984; Nicholson et al., 1993; Arita et al., 2011). However, ageing produced a dramatic rise in the mechanical strength of glass carbomer partly due to cement maturation (Mesquita et al., 1999), and also due to the dissolution-precipitation of HAp within the GC matrix which reinforced the cement matrix over time. The smaller-sized glass particles compared to that of the conventional GICs are thought to aid its dissolution and ultimate conversion to FAp and HAp (Van Duinen et al., 2004). SEM confirms the dispersion of mineral deposits on the surface of the aged GC cement (Figures 2-4-B-1, B-2), associated with abundant quantities of Ca and P in EDX (Figure 2-5, b). Zainuddin et al. (2012) supported the enhanced strength of GC over time and the formation of highly cross linked polyacid salt matrix.

In accordance to the literature (Mitra, 1991; Nicholson et al., 1992; Mitra and Kedrowsk, 1994; Uno et al., 1996), there were contradictory results on the effect of ageing on the mechanical properties of the investigated GIC/RMGICs. Some mechanical properties were significantly improved, some remained unchanged, while others were deteriorated post storage. When a polyalkenoic acid is mixed with the calcium aluminofluorosilicate glass in presence of water, the protons released from the acid cause hydrolysis of the glass to release Ca^{2+}, Al^{3+}, F^{-} and

PO_4^{3-} ions to form polyacrylates and a siliceous layer around the glass particles inhibiting its further degradation. Since the maturation of the cement occurs over time, the mechanical properties of GIC's tend to improve after storage and this trend was also clearly observed in the RMGICs. Strengthening is attributed to the additional crosslinking and build-up of a silica gel phase, whereas weakening may result from erosion and plasticising effect of water (Cattani-Lorente et al., 1993). The enhanced compressive strength in F2LC (C & H) post-ageing might be attributed to the lower early water sensitivity and the continuation of acid-base complexation which resulted in the formation of poly-HEMA and polyacrylic metal salts matrix surrounding the glass particles (Wilson, 1990; Tosaki, 1994; Momoi et al., 1994; Uno et al., 1996; Gladys et al., 1997).

There is a general agreement in the relevant literature regarding the better long-term physical integrity of the RMGIC in comparison to conventional GICs, characterised by higher flexural/tensile strengths and toughness (Douglas, 1994; Mitra and Kedrowski, 1994; Li et al., 1995; McCabe, 1998). The cross-linked poly-HEMA and polyacrylate salts form a homogeneous matrix that reduces the flexural modulus coupled with higher flexural strength, which increases the material resilience in stress-bearing sites. This fact is true for the BFS of the F2LC (C&H), which showed more than two-fold higher values than the CGICs (C&H), but for other mechanical properties (CS and modulus, MH), F2LC (C&H) showed lower values than the correspondent CGIC (C&H). This might be related to the use of recent improved GIC products (HVGICs) that are characterised by smaller glass particle size and an increased powder: liquid ratio, compared to the previously launched normal viscosity GICs. Greater cross-linking in the high-viscosity GIC's matrix enhance the mechanical properties including compressive strength, flexural strengths, surface hardness and wear resistance as previously reported (Pereira et al., 2002; Sidhu, 2011; Zanata et al., 2011).

The inferior properties of the hand-mixed F2LC when compared to all tested GICs encourage selecting this commercial product to engineer a synthetic dental biomaterial that widens its dental applications whilst improving the clinical performance, which is the rationale for developing this novel class of material with potential function as a reparative material. The concept integrates fundamental principles of the inherent ability of resin-modified glass-ionomers (Nicholson and

Czarnecka, 2016) to adhere to tooth tissue (Nicholson, 2016) and manipulate the incorporation of a polymerisable phosphate based monomer namely ethylene glycol methacrylate phosphate in immobilising it in the cement thereby promoting the interaction of the ligating phosphate groups with the ions within the glass matrix and the mineral component of tooth tissue. In addition, these polar phosphate groups tethered to the polymer backbone may enable higher affinity and bonding efficacy to relevant substrates. The EGMP-HEMA allows for the polymerisation, which is hypothesised to not only create a network of covalently linked phosphate groups but additionally improve adhesion to dental resin composites, RMGIC/GIC's and amalgams by virtue of the polar phosphate groups.

Ethylene glycol methacrylate phosphate (EGMP) is a proton-conducting electrolyte and the complexation behaviour of the carbonyl and phosphoryl ligating groups has been reported to enable remineralisation in hydrogels (Kemal et al., 2011) due to the charge in the gel and also improves the bonding efficacy and durability of self-etching adhesives (Inoue et al., 2005). Combining the properties of EGMP monomer as an adhesion promoter within RMGICs is a unique and interesting concept especially as the pendant phosphate groups are expected to interact with metallic cations during setting of the cement to form complexes that might alter the physical/biological properties of the cement itself and influence its adhesion to other restorative and dental substrates.

EGMP monomer is miscible and compatible with the co-monomer HEMA (Suzuki et al., 2006) and the liquid phase of Fuji II LC, which was confirmed by the lack of any evidence of phase separation. At lower concentrations of EGMP, no discernible changes were observed in the working time of the cements, however a significant increase ($p<0.05$) resulted with the higher concentrations of EGMP (30-40% wt.) in the formulation. It is interesting to note that at lower concentrations of EGMP the working time remained unaffected most likely due to the phosphoric acid groups being neutralised and integrated with the calcium ions. At higher concentration, there may a competing reaction of the phosphate and carboxyl groups to interact with the calcium ions, which prolongs the working time significantly, nevertheless it still meets the clinical requirements for water-based cements (ISO, 9917-2, 2010). Importantly, the inclusion of EGMP did not intervene with the setting time of the resultant cements. Since the acid-base reaction occurs due to the presence of the acidic polymer solution and alkaline glass powder irrespective of the presence of EGMP, which undergoes photo-polymerisation, the setting reaction remains unaffected.

pRMGIC manifested a pronounced enhancement in the compressive strength post-ageing when compared to the control cement (p<0.05). On short-term ageing, only highly contained EGMP (30% and 40%) showed a significantly improved CS when compared to the control (p<0.05), however after 6 months' storage, all modified formulations showed this trend. The compressive modulus of the experimental cements were significantly higher (p<0.05) than the control cements at most time points and increased proportionally with higher EGMP content within cement matrix.

Ageing has a significant influence on the mechanical properties of all modified formulations when compared to their early values. The enhanced properties became more evident after storage and maturation of the cement. The EGMPs in the composition of the cement act as spacer molecules in the polyacid milieu, assisting the movement of the carboxylic acid groups tagged to the rigid polymer backbone providing a greater degree of freedom for the carboxylate ions. This allows higher conversion of the carboxylic acids to metal carboxylate complexes (salt-bridge formation) during the setting reaction and reduces the number of unreacted carboxylic acid groups due to steric hindrance, which in turn improves the strength of the resultant cement (Nuttelman et al., 2006). Furthermore, the EGMP monomer with methacrylate residues can be polymerised readily via a free-radical initiation producing a covalently linked matrix of random homopolymers or even copolymerized with HEMA-producing copolymers of EGMP-HEMA (Kemal et al., 2011; Nuttelman et al., 2006). Both polymers can reinforce the matrix yielding cements with improved properties.

The presence of the ligating phosphate groups in the matrix produces a synergistic effect via the formation of a double-network structure. The reduction of pores result in a denser matrix with improved microhardness and the data shown in Table 3-5 confirmed that the EGMP 30 and EGMP 40 cements exhibited statistically significantly higher values than the control cement (p<0.001) and this trend was apparent even with low fractions of EGMP when the cements were aged over 4 weeks.

The more prominent effect of EGMP was on the biaxial flexural strength (BFS) which showed a double increase in value ($p<0.001$) in comparison to the control cement at all time intervals. The interatomic or intermolecular forces within the material have a significant effect on the BFS and the presence of the strong hydrophilic domains within the cement matrix is likely to inhibit the separation of the planes of atoms within matrix (Yamazaki et al., 2005), increasing the polar-polar interaction (Xie et al, 2004; Yamazaki et al., 2005). This is confirmed by the fact that an increase in the concentration of EGMP increases BFS values with minimal effect on ageing. The physicochemical interactions may also affect the strength of the cement matrix since there is a possibility of formation of H-bonds due to the presence of hydroxyl, phosphate and carbonyl groups within the matrix, reinforcing the cement. The stronger bonds between the organic and inorganic network of the set cement, lead to superior mechanical properties of final set cement.

The enhanced overall strength are consistent with the SEM findings that show a denser microstructure of the experimental cements (EGMP20, 30) in comparison to the control. The microstructural assessment of the fractured EGMP30 surface using SEM in Figure 3-9, EGMP30-1 and 2, shows dispersion of particles with varying size and shape with clear evidence of mineral deposits within matrix pores accompanied by morphological surface variations distinctly different from the particles of the cement. The IR spectra (Fig 3-3b) confirmed these findings, as the modified formulations showed a strong, absorption band at 966 cm^{-1} assigned to $v1$ stretching vibrations of the phosphate PO_4 in the apatite (Rey et al., 1990; Gadaleta et al., 1996). These changes confirm the inductive ability of these negatively charged functional groups for apatite precipitation within the body environment, similar to pervious findings of Stancu et al. (2004). Furthermore, EDX analysis for both experimental and control cements showed a distribution of the F, Si, Al, P and Ca within their matrices, however it was difficult to obtain a meaningful Ca/P ratio, but an abundance of P was observed within the modified formulations, which increased proportionally with the increasing content of the acidic functional monomer within the cement matrix.

The study of the interaction of these class of cements with water is vital, since the progression of cement setting and maturation is dependent on moisture that drives the acid-base reaction, gained from the oral environment. The RMGICs (pRMGIC and F2LC), in agreement with previous findings (Kanchanavasita et al., 1997), exhibited a Fickian behaviour at the early stages of water diffusion. The trend in water uptake showed a concomitant increase with increasing the EGMP content (Table 3-7), which is attributed to the water affinity of the polar molecule which parallels to earlier reports on the proportional correlation between the equilibrium water content of a HEMA-co-EGMP containing hydrogel and the content of EGMP copolymer (Kemal et al., 2011). In the EGMP-RMGICs, the polarity of the acidic functional phosphate group is responsible for the higher water affinity, which contributes to the ongoing acid-base reaction and the formation of stable ionic interactions with time that leads to the formation of a denser matrix with lower porosity, as evidenced by SEM images. This dense matrix imposes a certain resistance to water intrusion, decreasing the rate of water diffusion and significantly reducing the solubility as shown in Table 3-7. They are both correlated with the proportion of the EGMP monomer within the matrix; the greater weight percentage of the monomer, the lower the diffusion coefficient and solubility. Furthermore, this tightly bonded polyalkenoate matrix restricts the early fluoride elution during the first 48 h which is proportional to the concentration of the EGMP within the cement matrix, since the release of F- ion is mainly through diffusion from micro-porosities at this stage (Thanjal et al., 2010). However the migration of the ions with time show that the cements ultimately show similar behaviour when there is no replenishment of the fluoride ions (Figure 3-6).

RMGIC has a self-etching effect on dentine that augments the hybridisation with dentine, which is reflected in the micro-tensile bond strengths achieved. Since the pRMGIC's contain the additional functional co-monomer EGMP, the lower pH and the chelating ability of the phosphate groups with the residual hydroxyapatite (Nicholson, 2016) in dentine, led to a robust and durable bond with dentine over time, thereby accounting for the higher micro-tensile bond strengths reported in this study. These interactions are further confirmed by a shift in the mode of failure from predominantly adhesive in the control group to mixed and cohesive patterns for the EGMP30 and 40, (Figure 3-8). An improved adaptation of the cement to the dentine is most likely caused by the moisture present that is expected to be reused by the reaction between the acidic functional groups and the ion-releasing basic filler particles. The interaction of EGMP0 versus the EGMP30 with dentine was determined morphologically using scanning electron micrograph (Figure 5-11, 12, respectively). Partial and complete closure of the dentine tubules can be observed on the dentine surface as compared to the control group. Additionally, dispersion of irregularly shaped particles could be recognised over dentine surface following the interfacial adhesive failure of the dentine/EGMP30, as shown in Figure 3-12 (C, and C-1). The mineral deposition over the surface after two weeks immersion in SBF at 37°C indicates the possibility of pRMGIC to induce mineralisation on the adjacent tooth interface. However, as alluded earlier, it is difficult to analyse the deposits by EDX-SEM, and further studies are required for confirmation.

EGMP30 was selected for the final cement formulation pRMGIC as it showed an enhanced stable mechanical strength and bonding strength to dentine as compared to the control ($p<0.05$). Although, some values were comparable to the EGMP20 and EGMP40 cements, but this percentage allows the availability of pendent phosphate groups within the matrix without compromising the acid-base reaction of the cement.

One of the proposed aims of this study was achieved, the acidic functional groups in pRMGIC confer the self-etching nature of the cement which induce greater prism exposure leading to more micro-mechanical interlocking to sound and demineralised enamel tissues. This leads to increased early and delayed SBS values to sound enamel (24, 26 MPa, respectively) as compared to the control cement that shows comparable values to what are previously reported in the literature (≃18.7MPa) (Glasspoole et al., 2002). The pendant phosphate units in the resin matrix of the cement provide additional functional groups that can interact with tooth enamel thus producing stable bond strength after ageing ($p=0.233$). Chemical integration to enamel was confirmed by higher percentages of cohesive failures. Under SEM, Figure 4-6 (A-3) pRMGIC appeared to be well integrated to the exposed enamel rods in mixed failure specimens.

The morphological and structural alterations in the hypomineralised enamel on prolonged acid exposure reduce the surface hardness with irregular apatitic structure (Xie et al., 2008; Chan et al., 2010) and porosities at the resin-enamel interface (William et al., 2006) added to the irregular etch patterns following the application of self-etch resin cement which might compromise the cement infiltration into the substrate (William et al., (2006). These factors lead to reduction in the early and delayed SBS of pRMGIC to the demineralised enamel as compared to sound enamel ($p<0.001$). Nevertheless, this modification produces cement with higher early and prolonged bond strength in comparison to all tested commercial materials ($p<0.05$). The inductive ability of the phosphate groups as previously investigated by Stancu et al. (2004) encourage the formation of CaP complexes when stored in SBF at 37°C by chelation with the residual calcium ions. These complexes are precipitated in the form of minerals which could potentially penetrate the demineralised lesion and enhance the remineralisation, thus reinforcing the substrate resistance against shear loads over time, however, the increase in strength after storage was not statistically significant (p =0.177). High bond strengths were associated with higher percentages of mixed and cohesive failures, Fig (4-8, A-3) in which pRMGIC was identified firmly integrated to the white enamel lesion that are not recognised in other groups. However, extensive qualitative and quantitative assessments of the interfacial area are needed to confirm these observations.

The same trend was recognised in bonding to sound dentine. Functionalisation of the RMGIC produced higher initial and delayed SBS values (11.7, 18.8 MPa, respectively) than the control (8.2, 10.7 MPa, respectively) ($p<0.001$) and the reported values in the literature (≃7 MPa) (Nicholson, 2016). The hydrophilicity of the cement offered by polar phosphate groups and HEMA with a naturally hydrated substrate facilitate the ionisation of the acid groups that promoted the self-etching characteristics of the RMGIC and thus polymer infiltration into the substrate, which produce higher immediate bond strength (Lima et al., 2008; Van Meerbeek et al., 1998). Subsequently, water is reused by acid-base reaction between the functional groups and the ion-releasing basic filler particles during maturation which expected to further enhanced the bond strength of pRMGIC after ageing ($p<0.001$). The chemical bonding ability of the cement and the chelating effect of the phosphate group impart in more chemical interactions with the residual hydroxyapatite (Fu et al., 2005; Yoshida et al., 2004; Inoue et al., 2005) which yield a shift in the mode of failure from predominantly adhesive to mostly mixed and cohesive post-ageing, (Fig 4-9). Mineral formation in the dentinal tubule entrances are deposited gradually until the complete sealing of the lumen of the tubule as shown by scanning electron micrograph figure 4-10 (A-3). In contrast, failure in F2LC and F9 occurs above the hybrid layer where cements' remnant entirely covers the debonded interface, Figure 4-10 (B-3, and C-3, respectively).

Theoretically, the reduced mineral phase with lower hardness (Ceballos et al., 2003; Banerjee et al., 2010) and poor quality of the hybrid layer (Nakajima et al., (2005) can jeopardise the micromechanical interlocking of the adhesive polymers to carious affected dentine in comparison to sound dentine (Nakajima et al.,1999; Sattabanasuk et al.,2005; Erhardt et al., 2008). Even though, the self-etching effect as previously explained, may induce further mineral loss from the lesion. However, a thicker hybrid layer due to the availability of the ions might affect the micromechanical interlocking. But the potential of forming complexes between functional groups and hydroxyapatite is expected to reinforce the defective substrate and produce a stable bond over time. This is also translated by the lack of adhesive failure in both terms, Fig (4-11).

Although at the early term SBS values of pRMGIC to CAD was comparable to the control cement (p=1.000), pRMGIC maintained its strength over time (p=0.294), while F2LC lost half of its strength after storage (*p*=0.001). The extra moisture in this substrate (Ito et al., 2005) might induce hydrolysis of the resin and collagen fibrils reducing the bonding strength after storage for both F2LC and FS groups. In contrast, the promoted chemical interaction with surface minerals via the virtue of phosphate groups that maintain the strength against hydrolytic degradation of the resinous component in the cement. A similar behaviour was seen in the conventional GIC (F9) that also maintain stable strength after storage when bonded to carious lesion. SEM observation reveals that failure of pRMGIC occurred within the hybrid layer, and the CAD surface is well infiltrated by the cement with some resin tags remain occluding the tubules, (Figure 4-12, A-3). The durable seal of pRMGIC to the natural carious lesion might be efficient to hinder the caries progression and help in repairing defective tissue and thus reinforce the tooth-restoration complex which may need to be further investigated.

Although the phosphate groups have affinity to enhance physical bonding to amalgam via their hydroxyl groups (Wada, 1986; Yoshida et al., 2005) with a possibility of chemical union between the carboxyl group anions of polyacrylic acid and the metallic cations in dental amalgam (tin and silver) (Negm et al., 1982), pRMGIC could not prove an enhanced adhesion strength to this substrate. However, using adhesive contained 10-MDP prior to pRMGIC could significantly improve the initial and long-term bonding strength that is comparable to that of adhesive resin composite restorations after storage. The low viscosity of the bonding agent and polarity of the phosphate groups in pRMGIC and adhesive enhance surface wetting enabling better infiltration with improved physical interlocking into the micro-retentive amalgam surfaces. Furthermore, the compatibility in the resinous components of pRMGIC and adhesive that might covalently copolymerised through the long carbon chain in 10-MDP, and bonds chemically via phosphate moieties to the positively charged metallic ions at the alloy surface (Souza et al., 1994) producing durable SBS over time associated with a shift in the mode of failure from entirely adhesive to partly mixed failures (50%).

Interestingly, pRMGIC revealed a significantly improved durable bond strength to conditioned resin composite substrates with and without adhesive intervention. The enhanced wettability due to cements' initial hydrophilicity facilitates cement infiltration into the irregularities created by surface roughening, producing an initial strength that is significantly higher than its early value with adhesive, added to the possibility of chemical coupling with the aged resin matrix or the exposed filler particles (Kallio et al., 2001). After three months' storage, the values are higher than all tested materials indicating that the photoreactive monomer may create a network of covalently linked phosphate groups that enhanced the bulk properties of the resultant cement and additionally can bond to the cross-linked polymer matrix and filler particles of composite producing durable bond higher than the cohesive strength of the cement or the substrate characterised by high percentages of cohesive failures.

The same trend was seen in repairing conditioned RMGIC surfaces. The structural reinforcement post maturation enhanced the cohesive strength of the pRMGIC and increased the resistance to debonding forces. pRMGIC augmented the long-term bond strength in comparison to all tested materials, which was further enhanced when applied in conjunction with an adhesive. The moisturised environment due to polarity of the functional group encourages more interactions within the cement over time which enhances the mechanical strength and encourages the reactive phosphate groups and the polyacrylic acid anions to chemically integrate with the exposed glass particles of the substrate which leads to a robust and durable bond strength that was limited by the cohesive strength of the substrate. These interactions were supported by McLean, (1992), and Yap et al. (2000) who suggested that the exposed glass particles in the ‘old’ GIC material could react with the polyacrylic acid in the ‘new’ material and establish a chemical bond. The adhesive agent flows into the surface irregularities promoting the micromechanical attachment to the underlying roughened cement producing higher adhesion strength. Additionally, under a moist environment, functional groups in pRMGIC and SU induce a zone of ionic interaction within the cement matrix that reinforce the cement strength as well as promotes its bonding to the substrate. However, the failure observed at 3 months in pRMGIC are entirely cohesive irrespective of a bonding agent being used or not.

The enhanced mechanical strength of the pRMGIC established a strong early and delayed bond to conditioned GIC substrate that are comparable to that of resin composite without applying an adhesive. The hydrophilicity of the pRMGIC enhance the wettability that improve mechanical interlocking to this substrate (Mount, 1989). This moist environment triggers more ionic interactions by the reactive phosphate groups to the exposed alkaline glass particles in F9 substrate which presumed to produce a chemical union with the substrate that leads to further development in strength after storage (p=0.001) confirmed by entirely cohesive failure within the substrate. However, applying an adhesive prior to pRMGIC placement did not influence the bond strength values but the failures are totally cohesive (100%) within the GIC substrate. This indicates that these values did not reflect the true interfacial bond strength between the adhesive restorations and substrate rather the cohesive strength of the substrate itself. This trend may be due to the low mechanical strength of the substrate, the non-uniform stress distribution concentrating on the cement and the microporosities within the cement which itself acting as potential stress points leading to its premature failure before the interface itself (Placido et al., 2007, Zhang et al ., 2011).

In conclusion, this thesis has demonstrated that functionalisation of the RMGIC by intercalation of a 30% EGMP monomer can yield a photo-polymerisable cement with enhanced mechanical strength and promoted physical and chemical adhesion strength to different tooth/restorative interfaces by virtue of polar phosphate groups. pRMGIC can adhere and potentially seal diseased enamel and dentine tissues paving the way for further assessment for the structural changes of the dental tissues at the interfacial areas and the remineralisation potential of the new cement. The enhanced and prolonged quality of the repair produced by pRMGIC presented by high interfacial bond strength between joined surfaces indicate that this novel designated approach can enhance the integrity of tooth-restoration complex and thus prolonging the longevity of repair for failing restorations.

6.2 Conclusions

1. Encapsulated GICs/RMGICs exhibited superior physical properties compared to their hand-mixed equivalents tested at two time intervals. Encapsulated RMGIC’s showed satisfactory mechanical properties in comparison to the conventional GICs, whilst hand-mixed RMGIC exhibited inferior compressive strength, modulus and microhardness values. The inclusion of reactive glass particles enhanced the properties in Equia® Forte Fil, whilst Glass Carbomer cements (< 6% hydroxyapatite) showed inferior mechanical strength coupled with high fluoride release and a potential remineralising capability.

2. The incorporation of a phosphate-based monomer to the commercial RMGIC showed promising results for the phosphorylated glass-ionomer systems. It improved the immediate and long-term compressive strength and modulus, microhardness, with a two-fold increase in the biaxial flexural strength.

3. The polarity of the acidic functional phosphate group is responsible for the higher water affinity, which contributes to the ongoing acid-base reaction and the formation of stable ionic interactions with time that leads to the formation of a homogenous matrix, as evidenced by SEM images. This dense matrix imposes a certain resistance to water intrusion, decreasing the rate of water diffusion and significantly reducing the solubility and the early fluoride release.

4. The optimal proportion for EGMP is 30% by weight that showed an enhanced higher strength up to 6 months with higher adhesion strength to sound dentine. SEM observation showed minerals precipitation within the EGMP-contained cements after two weeks storage in SBF at 37°C coupled with the results of the FTIR analysis showed promise for a repair material and/or workable cements for several dental applications.

5. The potential of augmented chemical integrations of the innovative cement via the phosphate group with the remaining tooth structure enhance the bonding strength to sound and demineralised enamel showing chemical

integration to the exposed enamel rods under SEM. The same trend was seen in the adhesion strength to dentine and CAD showing obliterated dentinal tubules by resinous tags under SEM. Failures are mostly mixed and cohesive in most substrates.

6. The hydrophilic environment due to polarity of the functional groups encourage better interactions within the pRMGIC over time which enhance the mechanical strength and encourages the reactive phosphate groups to chemically integrate with the underlying restorative substrate producing an enhanced prolonged bond strength to conditioned resin composite, RMGIC and GIC substrates with high percentages of mixed and cohesive failures. However, the adhesion of pRMGIC to dental amalgam was established only when used in conjunction with SU adhesive.

7. Finally, the phophorylated resin-modified glass ionomer cement has shown to be promising reparative/restorative grade material regarding the improved mechanical and adhesion properties as a newly developed reparative system for the treatment of compromised TRCs would potentially improve the success of the restoration and the survival of TRCs.

6.3 Suggestions for future work

1. The superior physical properties of the mechanical-mixing than manual-mixing in the study, added to the enhanced physical properties of the EGMP-contained cements lay the foundation to develop and investigate the properties using an automated method via an encapsulated system where the operator and dispensing variability are excluded.
2. The enhanced mechanical, bonding and biological properties of the EGMP-contained cement that developed in the current work can be accomplished by modulating the self-reinforcing and functionality of the matrix through the inclusion of reactive particles that are able to form apatitic phases like bioactive glass or β-Tricalcium phosphate (β-TCP). Thereby, the integrated structure might possess enhanced strength and functionality via the dissolution-precipitation concept that might be suitable for repairing defective tooth tissues.
3. The enhanced interfacial adhesive strength of the EGMP-contained resin-modified glass ionomer cement necessitates an extensive qualitative and quantitative assessment of the physical and chemical changes in tooth tissues using nano-indentation test, Raman spectroscopy & second-harmonic imaging microscopy (SHG), to clarify the therapeutic effect and the remineralising potential of this novel cement.
4. An effective reparative material for the treatment of failing TRCs following the MI philosophy was established in the current study via the development of a novel cement with enhanced physical properties, robust and stable adhesive strength to different tooth/restorative interfaces with a mineralising potential. This has paved the way for further studies to evaluate the performance and feasibility of this system clinically in a randomised clinical trial, evaluating the clinical significance of using this material on long-term survival of failed TRCs.

References

ABOUSH, Y.E. & JENKINS, C.B. 1989. The bonding of glass-ionomer cements to dental amalgam. *British Dental Journal,* 166, 255-7.

ABOUSH, Y.E., ELDERTON, R.J. 1991. Bonding of a light curing glass-ionomer to dental amalgam. *Dental Materials,* 7,130-2.

AGHA, A., PARKER, S. & PATEL, M.P., 2016. Development of experimental resin modified glass ionomer cements (RMGICs) with reduced water uptake and dimensional change. *Dental materials*, 32, 713-22.

AKINMADE, A. & NICHOLSON, J.W.1995. Poisson's ratio of glass polyalkenoate (glass-ionomer) cements determined by an ultrasonic pulse method. *Journal of Materials Science: Materials in Medicine*, 6, 483-5.

AL-ASSAF, K., CHAKMAKCHI, M., PALAGHIAS, G., KARANIKA-KOUMA, A. & ELIADES, G., 2007. Interfacial characteristics of adhesive luting resins and composites with dentine. *Dental Materials*, 23, 829-39.

ALIREZAEI, M., BAGHERIAN, A. & SHIRAZI, A.S., 2018. Glass ionomer cements as fissure sealing materials: yes or no?: A systematic review and meta-analysis. *The Journal of the American Dental Association*. 149, 640-9.

ALLANDER, L., BIRKHED, D. & BRATTHALL, D., 1989. Quality evaluation of anterior restorations in private practice. *Swedish Dental Journal*, 13, 141-50.

ANDERSON, O.H. & DAHL, J.E. 1994. Aluminium release from glass ionomer cements during early water exposure in vitro. *Biomaterials,* 15, 882-8.

ANDRADE, A.P., SHIMAOKA, A.M. & DE CARVALHO, R.C., 2017. Composite resin repairs: What is the most effective protocol? *Brazilian Dental Science*, 20, 99-109.

ANDRZEJEWSKA, E., ANDRZEJEWSKI, M., SOCHA, E. & ZYCH-TOMKOWIAK, D., 2003. Effect of polyacid aqueous solutions on photocuring of polymerizable components of resin-modified glass ionomer cements. *Dental Materials,* 19,501-9.

ANSTICE, H.M. & NICHOLSON, J.W., 1992. Studies on the structure of light-cured glass-ionomer cements. *Journal of Materials Science: Materials in Medicine*, 3, 447-451.

ANUSAVICE, K.J., 1996. Restorative resins. Composite restorative materials. *Phillips' science of dental materials*, 298-9.

ANTONUCCI, J.M. 1987. Formulation and evaluation of resin modified glass ionomer cements. Transactions of the 13th Annual Meeting. *Social Biomaterials*, New York, 225.

ANTONUCCI, J.M. & STANSBURY, J.W. 1989. Polymer modified glass ionomer cements. *Journal of Dental Research,* 68, 251, [Abstract 555].

ALTMAN, D. G. *Practical Statistics for Medical Research.* 455-460.London: Chapman & Hall, 1991.

ANUSAVICE, K. J. 2003. *Phillips' science of dental materials,* Philadelphia, Pa.; London, W.B. Saunders.

ANUSAVICE, K.J., KAKAR, K. & FERREE, N., 2007. Which mechanical and physical testing methods are relevant for predicting the clinical performance of ceramic-based dental prostheses?. *Clinical Oral Implants Research*, 18, 218-231.

ARENDS, J., DIJKMAN, G.E. & DIJKMAN, A.G. 1995. Review of fluoride release and secondary caries reduction by fluoridating composites. *Advances in Dental Research,* 9, 367-76.

ARITA, K., YAMAMOTO, A., SHINONAGA, Y., HARADA, K., ABE, Y., NAKAGAWA, K., SUGIYAMA, S. 2011. Hydroxyapatite particle characteristics influence the enhancement of the mechanical and chemical properties of conventional restorative glass ionomer cement. *Dental Materials Journal,* 30, 672-83.

ARMSTRONG, S.R., BOYER, D.B. & KELLER, J.C., 1998. Microtensile bond strength testing and failure analysis of two dentin adhesives. *Dental Materials*, 14, 44-50.

ARMSTRONG, S., GERALDELI, S., MAIA, R., RAPOSO, L.H.A., SOARES, C.J. & YAMAGAWA, J., 2010. Adhesion to tooth structure: a critical review of “micro” bond strength test methods. *Dental Materials*, 26, 50-62.

ARSLANOGLU, Z., ALTAN, H., SAHIN, O., TEKIN, M. & ADIGÜZEL, M. 2015. Evaluation of surface properties of four tooth-coloured restorative materials. *Acta Physica Polonica A, 128*, B310-B3.

ATAI, M. & WATTS, D.C., 2006. A new kinetic model for the photopolymerization shrinkage-strain of dental composites and resin-monomers. *Dental materials*, 22, 785-91.

ATMEH, A.R., CHONG, E.Z., RICHARD, G., FESTY, F. & WATSON, T.F., 2012. Dentin-cement interfacial interaction: calcium silicates and polyalkenoates. *Journal of Dental Research,* 91, 454-59.

ATTIN, T., BUCHALLA, W., KIELBASSA, A.M. & HELLWIG, E., 1995. Curing shrinkage and volumetric changes of resin-modified glass ionomer restorative materials. *Dental materials*, 11, 359-62.

BADER, J.D., MARTIN, J.A. & SHUGARS, D.A., 1995. Preliminary estimates of the incidence and consequences of tooth fracture. *The Journal of the American Dental Association*, 126, 1650-4.

BAIG, M.S., DOWLING, A.H., CAO, X. & FLEMING, G.J., 2015. A discriminatory mechanical testing performance indicator protocol for hand-mixed glass-ionomer restoratives. *Dental Materials*, *31*, 273-83.

BALKAYA, H., DEMIRBUGA, S., ÇAKIR, N.N., KARADAS, M. & ZORBA, Y.O., 2018. Micro-shear bond strength of universal adhesives used for amalgam repair with or without Alloy Primer. *Journal of Conservative Dentistry,* 21, 274-9.

BAN, S. & ANUSAVICE, K.J., 1990. Influence of test method on failure stress of brittle dental materials. *Journal of Dental Research*, 69, 1791-9.

BANERJEE, A. 1999. *Applications of scanning microscopy in the assessment of dentine caries and methods for its removal*. University of London.

BANERJEE, A., SHERRIFF, M., KIDD, E. & WATSON, T. 1999. Cariology: A confocal microscopic study relating the autofluorescence of carious dentine to its microhardness. *British Dental Journal*, 187, 206-10.

BANERJEE, A., COOK, R., KELLOW, S., SHAH, K, FESTY, F., SHERRIFF, M. & WATSON, T. 2010. A confocal micro - endoscopic investigation of the relationship between the microhardness of carious dentine and its autofluorescence. *European Journal of Oral Sciences,* 118**,** 75-9.

BANERJEE, A. & WATSON, T.F., 2015. *Pickard's guide to minimally invasive operative dentistry*. OUP Oxford.

BANERJEE, A., 2013. 'MI'opia or 20/20 vision?. *British Dental Journal*, 214,101.

BANERJEE, A., 2017. 'Minimum intervention'–MI inspiring future oral healthcare?. *British Dental Journal,* 223, 133-5.

BARAÚNA MAGNO, M., RODRIGUES NASCIMENTO, G.C., DA ROCHA, P., SOUZA, Y., RIBEIROC, G., D'PAULA, B., CORDEIRO LORETTO, S. & COPLE MAIA, L., 2016. Silorane-based Composite Resin Restorations Are Not Better than Conventional Composites-A Meta-Analysis of Clinical Studies. *Journal of Adhesive Dentistry*, 18, 375, 86.

BARACCO, B., PERDIGÂO, J., CABRERA, E., GIRÀLDEZ, I. & CEBALLOS, L. 2012. Clinical evaluation of a low-shrinkage composite in posterior restorations: one-year results. *Operative Dentistry,* 37,117-29.

BARRY, T. I., CLINTON, D. J. & WILSON, A. D. 1979. Structure of a Glass-Ionomer Cement and Its Relationship to the Setting Process. *Journal of Dental Research,* 58, 1072-9.

BEECH, D.R., 1973. Improvement in the adhesion of polyacrylate cements to human dentine. *British Dental Journal*, 135, 442-5.

BEHER, M., ROSENTRITT, M., LOHER, H. & HANDEL, G. 2006. Effect of variation from recommended powder/liquid ratio on some properties of resin-modified cements. *Acta Odontologica Scandinavica,* 64, 214-20.

BEHR, M., ROSENTRITT, M., REGNET, T., LANG, R. & HANDEL, G., 2004. Marginal adaptation in dentin of a self-adhesive universal resin cement compared with well-tried systems. *Dental materials*, 20,191-197.

BELLENGER, V., VERDU, J. & MOREL, E., 1989. Structure-properties relationships for densely cross-linked epoxide-amine systems based on epoxide or amine mixtures. *Journal of Materials Science*, 24, 63-8.

BERIAT, N.C. & NALBANT, D. 2009. Water absorption and HEMA release of resin-modified glass ionomers. *European Journal of Dentistry*, 3, 267-72.

BERRY, E.A. & POWERS, J.M., 1994. Bond strength of glass ionomers to coronal and radicular dentin. *Operative Dentistry*, 19, 122-6.

BERTACCI, A., BARONI, C., MARCHIONNI, S., GANDOLFI, M. G., MONTANARI, M. & PRATI, C. 2008. Clinical management and SEM evaluation of MIH affected teeth. *Journal of Operative Dentistry,* 6,185-92.

BERZINS, D.W., ABEY, S., COSTACHE, M.C., WILKIE, C.A. & ROBERTS, H.W., 2010. Resin-modified glass-ionomer setting reaction competition. *Journal of dental research*, 89, 82-6.

BILLINGTON, R., WILIAMS, J. & PEARSON, G., 1989. Glass ionomers: Practice variation in powder/liquid ratio. *Journal of Dental Research*, 68, 945.

BLATZ, M.B., CHICHE, G., HOLST, S. & SADAN, A., 2007. Influence of surface treatment and simulated aging on bond strengths of luting agents to zirconia. *Quintessence International*, 38, 745-53.

BLUM, I.R., SCHRIEVER, A., HEIDEMANN, D., MJÖR, I.A. & WILSON, N.H.F., 2003. The repair of direct composite restorations: an international survey of the teaching of operative techniques and materials. *European Journal of Dental Education*, 7, 41-8.

BLUM, I.R., HAFIANA, K., CURTIS, A., BARBOUR, M.E., ATTIN, T., LYNCH, C.D. & JAGGER, D.C., 2012. The effect of surface conditioning on the bond strength of resin composite to amalgam. *Journal of Dentistry*, 40, 15-21.

BLUM, I.R., LYNCH, C.D. & WILSON, N.H., 2014. Factors influencing repair of dental restorations with resin composite. *Clinical, Cosmetic and Investigational Dentistry*, *6*, 81-7.

BONSTEIN, T., GARLAPO, D., JOHN JR, D. & BUSH, P.J., 2005. Evaluation of varied repair protocols applied to aged composite resin. *Journal of Adhesive Dentistry*, 7, 41-9.

BORGES, B.C.D., DE SOUZA BORGES, J., DE ARAUJO, L.S.N., MACHADO, C.T., DOS SANTOS, A.J.S. & DE ASSUNÇAO PINHEIRO, I.V., 2011. Update on nonsurgical, ultraconservative approaches to treat effectively non-cavitated caries lesions in permanent teeth. *European Journal of Dentistry*, 5, 229-36.

BOYER, D.B., CHALKLEY, Y. & CHAN, K.C., 1982. Correlation between strength of bonding to enamel and mechanical properties of dental composites. *Journal of Biomedical Materials Research*, 16, 775-83.

BOSTON, D.W. 2003. Initial in vitro evaluation of DIAGNOdent for detecting secondary carious lesions associated with resin composite restorations. *Quintessence International*, 34,109-16.

BOURKE, A.M., WALLS, A.W. & MCCABE, J.F., 1992. Light-activated glass polyalkenoate (ionomer) cements: the setting reaction. *Journal of dentistry*, 20, 115-120.

BRAEM, M.J., LAMBRECHTS, P., GLADYS, S. & VANHERLE, G., 1995. In vitro fatigue behaviour of restorative composites and glass ionomers. *Dental Materials*, 11, 137-41.

BREDER, K., ANDERSSON, T. & SCHÖLIN, K., 1990. Fracture Strength of α-and β-SiAION Measured by Biaxial and Four-Point Bending. *Journal of the American Ceramic Society*, 73, 2128-30.

BRESCHI, L., MAZZONI, A., RUGGERI, A., CADENARO, M., DI LENARDA, R. & DORIGO, E.D.S., 2008. Dental adhesion review: aging and stability of the bonded interface. *Dental Materials*, 24, 90-101.

BRUNTON, P.A., GHAZALI, A., TARIF, Z.H., LOCH, C., LYNCH, C., WILSON, N. & BLUM, I.R., 2017. Repair vs replacement of direct composite restorations: a survey of teaching and operative techniques in Oceania. *Journal of Dentistry*, 59, 62-7.

BROSH, T., PILO, R., BICHACHO, N. & BLUTSTEIN, R., 1997. Effect of combinations of surface treatments and bonding agents on the bond strength of repaired composites. *The Journal of Prosthetic Dentistry*, 77, 122-6.

BONSTEIN T, GARLAPO D, JOHN JR D & BUSH PJ. 2005. Evaluation of varied repair protocols applied to aged composite resin. *Journal of Adhesive Dentistry*, 7, 41-9.

BURGESS, J., NORLING, B. & SUMMITT, J., 1994. Resin ionomer restorative materials: the new generation. *Journal of Esthetic and Restorative Dentistry*, 6, 207-15.

BURKE, F.M., & LYNCH, E. 1994. Glass polyalkenoate bond strength to dentine after chemomechanical caries removal. *Journal of Dentistry*, 22, 283–91

BURKE, F.J., CHEUNG, S.W., MÖHR, I.A. & WILSON, N.H., 1999. Restoration longevity and analysis of reasons for the placement and replacement of restorations provided by vocational dental practitioners and their trainers in the United Kingdom. *Quintessence International*, 30, 234-42.

BURKE, F.J., WILSON, N.H., CHEUNG, S.W. & MJOR, I.A. 2001. Influence of patient factors on age of restorations at failure and reasons for their placement and replacement. *Journal of Dentistry*, 29, 317-24.

BURROW, M.F., INOKOSHI, S. & TAGAMI, J. 1999. Water sorption of several bonding systems. *American Journal of Dentistry,* 12, 295-8.

CEFALY, D.F.G., MELLO, L.L.C.P.D., WANG, L., LAURIS, J.R.P. & D'ALPINO, P.H.P., 2009. Effect of light curing unit on resin-modified glass-ionomer cements: a microhardness assessment. *Journal of Applied Oral Science*, 17, 150-4.

CALHEIROS, F.C., SADEK, F.T., BRAGA, R.R. & CARDOSO, P.E., 2004. Polymerization contraction stress of low-shrinkage composites and its correlation with microleakage in class V restorations. *Journal of Dentistry,* 32, 407-12.

CALVO, A.F.B., ALVES, F.B.T., LENZI, T.L., TEDESCO, T.K., REIS, A., LOGUERCIO, A.D. & RAGGIO, D.P. 2014. Glass ionomer cements bond stability in caries-affected primary dentin. *International Journal of Adhesion and Adhesives,* 48,183-87.

CAMILLERI, J. 2013. Investigation of Biodentine as dentine replacement material. *Journal of Denti*stry, 41, 600-10.

CASAGRANDE, L., LASKE, M., BRONKHORST, E.M., HUYSMANS, M.C.D. & OPDAM, N.J. 2017. Repair may increase survival of direct posterior restorations- A practice based study. *Journal of Dentistry*, 64, 30-6.

CATTANI-LORENTE, M.A., GODIN, C. & MEYER, J.M. 1993. Early strength of glass ionomer cements. *Dental Materials,* 9, 57-62.

CATTANI-LORENTE, M.A., GODIN, C. & MEYER, J.M., 1994. Mechanical behavior of glass ionomer cements affected by long-term storage in water. *Dental Materials*, 10, 37-44.

CAUSTON, B. E. 1981. The physico-mechanical consequences of exposing glass ionomer cements to water during setting. *Biomaterials,* 2**,** 112-5.

CASAGRANDE, L., LASKE, M., BRONKHORST, E.M., HUYSMANS, M.C.D. & OPDAM, N.J., 2017. Repair may increase survival of direct posterior restorations– A practice based study. *Journal of Dentistry*, 64, 30-6.

CAVALCANTI, A.N., DE LIMA, A.F., PERIS, A.R., MITSUI, F.H.O. & MARCHI, G.M., 2007. Effect of surface treatments and bonding agents on the bond strength of repaired composites. *Journal of Esthetic and Restorative Dentistry*, 19, 90-8.

CEBALLOS, L., CAMEJO, D.G., VICTORIA, FUENTES M., OSORIO, R., TOLEDANO, M., CARVALHO, R.M. & PASHLEY, D.H. 2003. Microtensile bond strength of total-etch and self-etching adhesives to caries-affected dentine. *Journal of Dentistry,* 31, 469-77.

CEFALY, D.F., FRANCO, E.B., LIA MONDELLI, R.F., FRANCISCONI, P.A. & NAVARRO, M.F. 2003. Diametral tensile strength and water sorption of glass-ionomer cements used in atraumatic restorative treatment. *Journal of Applied Oral Science,* 11, 96-101.

CEFALY, D.F., BARATA, T.J., BRESCIANI, E., FAGUNDES, T.C., LAURIS, J.R. & NAVARRO, M.F. 2007. Clinical evaluation of multiple-surface ART restorations: 12-month follow-up. *Journal of Dentistry for Children*, 74, 203-8.

ÇEHRELI, S.B., ARHUN, N. & CELIK, C., 2010. Amalgam Repair: Quantitative evaluation of amalgam-resin and resin-tooth interfaces with different surface treatments. *Operative Dentistry*, 35, 337-44.

CHADWICK, R.G. & WOOLFORD, M.J., 1993. A comparison of the shear bond strengths to a resin composite of two conventional and two resin-modified glass polyalkenoate (ionomer) cements. *Journal of Dentistry*, 21,111-16.

CHAN, Y.L., NGAN, A.H. & KING, N.M. 2010. Degraded prism sheaths in the transition region of hypomineralized teeth. *Journal of Dentistry*, 38,237-44.

CHIRILA, T.V., HILL, D.J., WHITTAKER, A.K. & KEMP, A., 2007. Effect of phosphate functional groups on the calcification capacity of acrylic hydrogels. *Acta Biomaterialia*, 3, 95-102.

CHEETHAM, J.J., PALAMARA, J.E., TYAS, M.J. & BURROW, M.F., 2014a. A comparison of resin-modified glass-ionomer and resin composite polymerisation shrinkage stress in a wet environment. *Journal of the Mechanical Behavior of Biomedical Materials*, 29, 33-41.

CHEETHAM, J.J., PALAMARA, J.E., TYAS, M.J. & BURROW, M.F., 2014b. A comparison of the shear bond strength and failure mode to metals of unsupported and supported luting cement specimens. *Journal of Adhesive Dentistry*, *16*, 251-60.

CHUNG, S.M., YAP, A.U.J., CHANDRA, S.P. & LIM, C.T., 2004. Flexural strength of dental composite restoratives: Comparison of biaxial and three-point bending test. *Journal of Biomedical Materials Research Part B: Applied Biomaterials*, 71, 278-83.

COOK, W.D., 1983. Degradative analysis of glass ionomer polyelectrolyte cements. *Journal of biomedical materials research*, 17, 1015-27.

COSTA, S.M., MARTINS, C.C., BONFIM, M.D.L.C., ZINA, L.G., PAIVA, S.M., PORDEUS, I.A. & ABREU, M.H., 2012. A systematic review of socioeconomic indicators and dental caries in adults. *International Journal of Environmental Research and Public Health*, 9, 3540-74.

COUTINHO, E., YOSHIDA, Y., INOUE, S., FUKUDA, R., SNAUWAERT, J., NAKAYAMA, Y., DE MUNCK, J., LAMBRECHTS, P., SUZUKI, K. & VAN MEERBEEK, B. 2007. Gel phase formation at resin-modified glass-ionomer/tooth interfaces*. Journal of Dental Research,* 86, 656-61.

CRAIG, R.G., POWERS, J. M. & WATAHA, J.C. 2004. *Dental materials: properties and manipulation,* Mosby, St. Louis, 8th Ed.

CRANK, J., 1979. *The mathematics of diffusion*. Oxford university press.

CRISP, S. & WILSON, A. D. 1974. Reactions in glass ionomer cements: III. The precipitation reaction. *Journal of Dental Research,* 53**,** 1420-4.

CRISP, S., FERNER, A.J., LEWIS, B.G. & WILSON, A.D. 1975. Properties of improved glass-ionomer cement formulations. *Journal of Dentistry,* 3:125-30

CRISP, S. & WILSON A.D.1976. Reactions in glass ionomer cements. V. Effect of incorporating tartaric acid in the cement liquid, *Journal of Dental Research,* 55, 1023-31.

CRISP, S., LEWIS, B. G. & WILSON, A. D.1976. Characterization of glass-ionomer cements 1. Long term hardness and compressive strength. *Journal of Dentistry,* 4, 162-6.

CROLL, T.P. & NICHOLSON, J.W., 2002. Glass ionomer cements in pediatric dentistry: review of the literature. *Pediatric Dentistry*, 24, 423-9.

CULBERTSON, B.M. & KAO, E.C., Ohio State University, 1994. *Water soluble polymers containing amino acid residues for dental restoratives*. U.S. Patent 5,369,142.

CULBERTSON, B.M. 2006. New polymeric materials for use in glass-ionomer cements. *Journal of Dentistry*, 34,556-65.

CURZON, M.E.J. & LOSEE, F.L., 1977. Strontium content of enamel and dental caries. *Caries Research*, 11, 321-6.

CZARNECKA, B., LIMANOWSKA-SHAW, H. & NICHOLSON, J.W. 2002. Buffering and ion-release by a glass-ionomer cement under near-neutral and acidic conditions. *Biomaterials,* 23, 2783-8.

CZARNECKA, B. & NICHOLSON, J.W., 2006. Ion release by resin-modified glass-ionomer cements into water and lactic acid solutions. *Journal of Dentistry*, 34, 539-43.

CZARNECKA, B., DEREGOWSKA-NOSOWICZ, P., LIMANOWSKA-SHAW, H. & NICHOLSON, J.W. 2007a. Shear bond strengths of glass-ionomer cements to sound and to prepared carious dentine. *Journal of Materials Science: Materials in Medicine,* 18, 845-9.

CZARNECKA, B., LIMANOWSKA-SHAW, H., HATTON, R. & NICHOLSON, J.W., 2007b. Ion release by endodontic grade glass-ionomer cement. *Journal of Materials Science: Materials in Medicine*, 18, 649-652.

DA COSTA, T.R.F., SERRANO, A.M., ATMAN, A.P.F., LOGUERCIO, A.D. & REIS, A., 2012. Durability of composite repair using different surface treatments. *Journal of Dentistry*, 40, 513-21.

DALL'OCA, S., PAPACCHINI, F., GORACCI, C., CURY, Á.H., SUH, B.I., TAY, F.R., POLIMENI, A. & FERRARI, M., 2007. Effect of oxygen inhibition on composite repair strength over time. *Journal of biomedical materials research Part B: Applied Biomaterials*, 81,493-8.

D’ALPINO, P.H., BECHTOLD, J., DOS SANTOS, P.J., ALONSO, R.C.B., DI HIPÓLITO, V., SILIKAS, N. & RODRIGUES, F.P., 2011. Methacrylate-and silorane-based composite restorations: hardness, depth of cure and interfacial gap formation as a function of the energy dose. *Dental Materials*, 27, 11, 1162-9.

DARLING, M. & HILL, R. 1994. Novel polyalkenoate (glass ionomer) dental cements based on zinc silicate glasses. *Biomaterials,* 15, 299-306.

DARONCH, M., RUEGGEBERG, F.A., DE GOES, M.F. & GIUDICI, R., 2006. Polymerization kinetics of pre-heated composite. *Journal of Dental Research*, 85, 38-43.

DARVELL, B. W. 2000. Mechanical testing. In: Materials science for Dentistry. 6th ed. Hong Kong: University of Hong Kong, 1-34

DARVELL, B. 2012. Development of strength in dental silver amalgam. *Dental Materials,* 28, 207-17.

DAVIDSON, C.L., DE GEE, A. & FEILZER, A., 1984. The competition between the composite-dentin bond strength and the polymerization contraction stress. *Journal of Dental Research*, 63, 1396-9.

DAVIDSON, C.L., VAN ZEGHBROECK, L. & FEILZER, A.J., 1991. Destructive stresses in adhesive luting cements. *Journal of Dental Research,* 70, 880-2.

DAVIDSON, C.L. & MJÖR, I.A., 1999. Advances in glass-ionomer cements. Quintessence Publishing Co, Inc.

DEB, S. & NICHOLSON, J.W. 1999. The effect of strontium oxide in glass-ionomer cements. *Journal of Materials Science: Materials in Medicine,* 10, 471-4.

DELAVIZ, Y., FINER, Y. & SANTERRE, J.P., 2014. Biodegradation of resin composites and adhesives by oral bacteria and saliva: a rationale for new material designs that consider the clinical environment and treatment challenges. *Dental Materials*, 30, 16-32.

DEMARCO, F.F., CORRÊA, M.B., CENCI, M.S., MORAES, R.R. & OPDAM, N.J., 2012. Longevity of posterior composite restorations: not only a matter of materials. *Dental Materials*, 28, 87-101.

DE MOOR, R. J. & VERBEECK, R. M. 1998. Effect of acetic acid on the fluoride release profiles of restorative glass ionomer cements. *Dental materials,* 14**,** 261-8.

DE MORAES, M.D.R., DE MELO, M.A.S., BEZERRA, D.D.S., COSTA, L.S., SABOÍA, V.D.P.A. AND RODRIGUES, L.K.A., 2016. Clinical study of the caries-preventive effect of resin-modified glass ionomer restorations: aging versus the influence of fluoride dentifrice. *Journal of Investigative and Clinical Dentistry*, 7,180-6.

DE MUNCK, J., VARGAS, M., VAN LANDUYT, K., HIKITA, K., LAMBRECHTS, P. & VAN MEERBEEK, B., 2004. Bonding of an auto-adhesive luting material to enamel and dentin. *Dental Materials*, 20, 963-71.

DE MUNCK, J., VAN LANDUYT, K., PEUMANS, M., POITEVIN, A., LAMBRECHTS, P., BRAEM, M. & VAN MEERBEEK, B. 2005. A critical review of the durability of adhesion to tooth tissue: Methods and results. *Journal of Dental Research,* 84,118-32.

DE MUNCK, J., MINE, A., POITEVIN, A., VAN ENDE, A. & VAN MEERBEEK, B., 2010. Testing bond strength: A review of the literature. *Dental Materials*, 26, 139-140.

DE MUNCK, J., MINE, A., POITEVIN, A., VAN ENDE, A., CARDOSO, M.V., VAN LANDUYT, K.L., PEUMANS, M. & VAN MEERBEEK, B., 2012. Meta-analytical review of parameters involved in dentin bonding. *Journal of Dental Research*, 91, 351-7.

DERAND, T., BIRKHED, D. & EDWARDSSON, S., 1991. Secondary caries related to various marginal gaps around amalgam restorations in vitro. *Swedish Dental Journal*, 15, 133-8.

DE WITTE, A.M., DE MAEYER, E.A., VERBEECK, R.M. & MARTENS, L.C., 2000. Fluoride release profiles of mature restorative glass ionomer cements after fluoride application. *Biomaterials*, 21, 475-82.

DIAZ-ARNOLD, A.M., VARGAS, M.A. & HASELTON, D.R., 1999. Current status of luting agents for fixed prosthodontics. *The Journal of prosthetic dentistry*, 81, 135-41.

DAVIS, H.E., TROXELL, G.E. & WISKOCIL, C.T., 1964. The testing and inspection of engineering materials. 3rd edn. New York, McGraw–Hill.

DICKENS, S.H., FLAIM, G.M. & TAKAGI, S. 2003. Mechanical properties and biochemical activity of remineralizing resin-based Ca-PO_4 cements. *Dental Materials*, 19, 558-66.

DICKENS, S.H., KELLY, S.R., FLAIM, G.M. & GIUSEPPETTI, A.A. 2004. Dentin adhesion and microleakage of a resin-based calcium phosphate pulp capping and basing cement. *European Journal of Oral Sciences*, 112:452-7.

DICKENS, S.H. & CHO, B.H., 2005. Interpretation of bond failure through conversion and residual solvent measurements and Weibull analyses of flexural and microtensile bond strengths of bonding agents. *Dental Materials*, 21, 354-64.

DICKENS, S.H. & FLAIM, G.M. 2008. Effect of a bonding agent on in vitro biochemical activities of remineralizing resin-based calcium phosphate cements. *Dental Materials,* 24, 1273-80.

DIEM, V.T.K., TYAS, M.J., NGO, H.C., PHUONG, L.H. & KHANH, N.D., 2014. The effect of a nano-filled resin coating on the 3-year clinical performance of a conventional high-viscosity glass-ionomer cement. *Clinical Oral Investigations*, 18, 753-9.

DOBLOUG, A. & GRYTTEN, J., 2015. Dentist-specific effects on the longevity of dental restorations. *Community Dentistry and Oral Epidemiology*, 43, 68-74.

DOWLING, A.H. & FLEMING, G.J.P. 2008. Is encapsulation of posterior glass-ionomer restoratives the solution to clinically induced variability introduced on mixing? *Dental materials,* 24, 957-66.

DOWLING, A.H. & FLEMING, G.J. 2009. Are encapsulated anterior glass-ionomer restoratives better than their hand-mixed equivalents? *Journal of Dentistry,* 37, 133-40.

DOUGLAS, W.H., 1994. Strength of the new systems, Glass Ionomers. In *The next generation, Proceedings of the 2nd International Symposium on Glass Ionomers* (pp. 209-216).

DOS SANTOS, J.G., FONSECA, R.G., ADABO, G.L. & DOS SANTOS CRUZ, C.A., 2006. Shear bond strength of metal-ceramic repair systems. *The Journal of Prosthetic Dentistry,* 96,165-73.

DUNAND, D.C. & DERBY, B., 1993. Creep and thermal cycling. *Fundamentals of Metal Matrix Composites*, 191-214.

EISENBURGER, M., ADDY, M., HUGHES, J.A. & SHELLIS, R.P. 2001. Effect of time on the remineralisation of enamel by synthetic saliva after citric acid erosion. *Caries Research,* 35, 211-5.

EL HEJAZI, A.A. & WATTS, D.C., 1999. Creep and visco-elastic recovery of cured and secondary-cured composites and resin-modified glass-ionomers. *Dental Materials*, 15, 138-43.

ELIADES, G., PALAGHIAS, G. 1993. In vitro characterization of visible light cured glass ionomer liners. *Dental Materials*, 9, 198-203.

ELIASSON, S.T., TIBBALLS, J. & DAHL, J.E., 2014. Effect of different surface treatments and adhesives on repair bond strength of resin composites after one and 12 months of storage using an improved microtensile test method. *Operative Dentistry*, 39, E206-E16.

ELLAKURIA, J., TRIANA, R., MINGUEZ, N., SOLER, I., IBASETA, G., MAZA, J. & GARCIA-GODOY, F., 2003. Effect of one-year water storage on the surface microhardness of resin-modified versus conventional glass-ionomer cements. *Dental Materials*, 19, 286-90.

EL MALLAKH, B.F. & SARKAR, N.K., 1990. Fluoride release from glass-ionomer cements in de-ionized water and artificial saliva. *Dental Materials*, 6, 118-122.

ELTAHLAH, D., LYNCH, C.D., CHADWICK, B.L., BLUM, I.R. & WILSON, N.H., 2018. An update on the reasons for placement and replacement of direct restorations. *Journal of Dentistry*, 72, 1-7.

ERHARDT, M.C., TOLEDANO, M., OSORIO, R. & PIMENTA, L.A. 2008. Histomorphologic characterization and bond strength evaluation of caries-affected dentin/resin interfaces: Effects of long-term water exposure. *Dental Materials,* 24,786-98.

ERHARDT, M.C., PISANI-PROENCA, J., OSORIO, E., AGUILERA, F.S., TOLEDANO, M. & OSORIO, R., 2011. Influence of laboratory degradation methods and bonding application parameters on microTBS of self-etch adhesives to dentin. *American Journal of Dentistry*, 24, 103-8.

ERMIS, R.B., & AYDIN, U. 2004. Examiner agreement in the replacement decision of Class I amalgam restorations. *Journal of Contemporary Dental Practice,* 5, 81-92.

ESTAY, J., MARTÍN, J., VIERA, V., VALDIVIESO, J., BERSEZIO, C., VILDOSOLA, P., MJOR, I.A., ANDRADE, M.F., MORAES, R.R., MONCADA, G. & GORDAN, V.V., 2018. 12 Years of Repair of Amalgam and Composite Resins: A Clinical Study. *Operative Dentistry*, 43, 12-21.

FALSAFI, A., MITRA, S.B., OXMAN, J.D., TON, T.T. & BUI, H.T. 2014. Mechanisms of setting reactions and interfacial behavior of a nano-filled resin-modified glass ionomer. *Dental Materials,* 30, 632-43.

FARAH, R.A., SWAIN, M.V., DRUMMOND, B.K., COOK, R. & ATIEH, M. 2010. Mineral density of hypomineralized enamel. *Journal of Dentistry*, 38, 50-8.

FEATHERSTONE, J.D.B., SHIELDS, C.P., KHADEMAZAD, B. & OLDERSHAW, M.D., 1983. Acid reactivity of carbonated apatites with strontium and fluoride substitutions. *Journal of Dental Research*, 62, 1049-53.

FEILZER, A.J., DE GEE, A.J. & DAVIDSON, C.L., 1987. Setting stress in composite resin in relation to configuration of the restoration. *Journal of Dental Research*, 66, 1636-9.

FEILZER, A.J., DE GEE, A.J. & DAVIDSON, C.L., 1988. Curing contraction of composites and glass-ionomer cements. *Journal of Prosthetic Dentistry*, 59, 297-300.

FEILZER, A.J., KAKABOURA, A.I., DE GEE, A.J. & DAVIDSON, C.L., 1995. The influence of water sorption on the development of setting shrinkage stress in traditional and resin-modified glass ionomer cements. *Dental Materials*, 11, 186-190.

FERRACANE, J.L. & MARKER, V.A., 1992. Solvent degradation and reduced fracture toughness in aged composites. *Journal of dental research*, 71, 13-9.

FERRACANE, J.L., 2006. Hygroscopic and hydrolytic effects in dental polymer networks. *Dental Materials*, 22, 211-22.

FERRACANE, J.L., STANSBURY, J.W. & BURKE, F.J.T., 2011. Self-adhesive resin cements–chemistry, properties and clinical considerations. *Journal of Oral Rehabilitation*, 38, 295-314.

FERNÁNDEZ, E., MARTÍN, J., VILDÓSOLA, P., JUNIOR, O.O., GORDAN, V., MJOR, I., BERSEZIO, C., ESTAY, J., DE ANDRADE, M.F. & MONCADA, G., 2015. Can repair increase the longevity of composite resins? Results of a 10-year clinical trial. *Journal of Dentistry*, 43, 279-86.

FERRARI, M. & DAVIDSON, C.L., 1997. Interdiffusion of a traditional glass ionomer cement into conditioned dentin. *American Journal of Dentistry*, 10, 295-7.

FERRACANE, J.L. & BERGE, H.X., 1995. Fracture toughness of experimental dental composites aged in ethanol. *Journal of dental research*, 74, 1418-23.

FERRACANE, J.L., MITCHEM, J.C., CONDON, J.R. & TODD, R., 1997. Wear and marginal breakdown of composites with various degrees of cure. *Journal of Dental Research*, 76, 1508-1516.

FERRACANE, J.L., 2006. Hygroscopic and hydrolytic effects in dental polymer networks. *Dental Materials*, 22, 211-22.

FERRARI, M. & DAVIDSON, C.L., 1996. Sealing performance of Scotchbond Multi-Purpose-Z100 in Class II restorations. *American Journal of Dentistry*, 9, 145-149.

FLEMING, G.J., KENNY, S.M. & BARRALET, J.E., 2006. The optimisation of the initial viscosity of an encapsulated glass-ionomer restorative following different mechanical mixing regimes. *Journal of dentistry*, 34,155-63.

FLEMING, G.J.P., KHAN, S., AFZAL, O., PALIN, W.M. & BURKE, F.J.T., 2007. Investigation of polymerisation shrinkage strain, associated cuspal movement and microleakage of MOD cavities restored incrementally with resin-based composite using an LED light curing unit. *Journal of Dentistry*, 35, 97-103.

FORSS, H. & SEPPÄ, L., 1995. Studies on the effect of fluoride released by glass ionomers in the oral cavity. *Advances in Dental Research*, 9, 389-93.

FORSTEN, L., 1995. Resin-modified glass ionomer cements: fluoride release and uptake. *Acta Odontologica Scandinavica*, 53, 222-5.

FRANKENBERGER, R., LOHBAUER, U., SCHAIBLE, R.B., NIKOLAENKO, S.A. & NAUMANN, M., 2008. Luting of ceramic inlays in vitro: marginal quality of self-etch and etch-and-rinse adhesives versus self-etch cements. *Dental Materials,* 24,185-91.

FRENCKEN, J.E., VAN'T HOF, M.A., TAIFOUR, D. & AL-ZAHER, I., 2007. Effectiveness of ART and traditional amalgam approach in restoring single-surface cavities in posterior teeth of permanent dentitions in school children after 6.3 years. *Community Dentistry and Oral Epidemiology*, 35, 207-14.

FRIEDL, K.H., POWERS, J.M. & HILLER, K.A., 1995. Influence of different factors on bond strength of hybrid ionomers. *Operative Dentistry*, 20, 74-80.

FRUITS, T.J., DUNCANSON JR, M.G. & COURY, T.L., 1998. Interfacial bond strengths of amalgam bonded to amalgam and resin composite bonded to amalgam. *Quintessence International,* 29, 327-34.

FU, B., SUN, X., QIAN, W., SHEN, Y., CHEN, R. & HANNIG, M., 2005. Evidence of chemical bonding to hydroxyapatite by phosphoric acid esters. *Biomaterials*, 26, 5104-10.

FUKAZAWA, M., MATSUYA, S. & YAMANE, M. 1987. Mechanism for erosion of glass-ionomer cements in an acidic buffer solution. *Journal of Dental Research*, 66, 1770-4.

FUKUDA, R., YOSHIDA, Y., NAKAYAMA, Y., OKAZAKI, M., INOUE, S., SANO, H., SUZUKI, K., SHINTANI, H. & VAN MEERBEEK, B., 2003. Bonding efficacy of polyalkenoic acids to hydroxyapatite, enamel and dentin. *Biomaterials*, 24, 1861-7.

FUSAYAMA, T., OKUSE, K. & HOSODA, H. 1966. Relationship between hardness, discoloration, and microbial invasion in carious dentin. *Journal of Dental Research,* 45**,** 1033-46.

FUSAYAMA, T., 1997. The process and results of revolution in dental caries treatment. *International Dental Journal*, 47, 157-66.

GADALETA, S.J., PASCHALIS, E.P., BETTS, F., MENDELSOHN, R. & BOSKEY, A.L., 1996. Fourier transform infrared spectroscopy of the solution-mediated conversion of amorphous calcium phosphate to hydroxyapatite: new correlations between X-ray diffraction and infrared data. *Calcified Tissue International*, 58, 9-16.

GERDOLLE, D.A., MORTIER, E. & DROZ, D., 2008. Microleakage and polymerization shrinkage of various polymer restorative materials. *Journal of Dentistry for Children,* 75, 125-33.

GERTH, H.U., DAMMASCHKE, T., ZÜCHNER, H. & SCHÄFER, E., 2006. Chemical analysis and bonding reaction of RelyX Unicem and Bifix composites-a comparative study. *Dental Materials*, 22, 934-41.

GEURTSEN, W., 1992. The cracked-tooth syndrome: clinical features and case reports. *International Journal of Periodontics & Restorative Dentistry*, 12, 394-405.

GEURTSEN, W., 2000. Biocompatibility of resin-modified filling materials. *Critical Reviews in Oral Biology & Medicine*, 11, 333-5.

GEURTSEN, W. & LEYHAUSEN, G. 2001**.** Chemical**-**biological interaction of the resin monomer triethyleneglycoldimetacrylate (TEGDMA). *Journal of Dental Research*, 80, 2046-50.

GLADYS, S., VAN MEERBEK, B., BRAEM, M., LAMBRECHTS, P. & VANHERLE, G.1997. Comparative physic-mechanical characterization of new hybrid restorative materials. *Journal of Dental Research,* 16, 883-94.

GLANTZ, P.O. 1977. Adhesion to teeth. *International dental journal,* 27,324-32.

GLASSPOOLE, E.A. & ERICKSON, R.L., 1993. In vitro investigation of the caries inhibition effects of fluoride releasing materials. *Journal of Dental Research*, *72*, Special issue: Abstract 1448.

GLASSPOOLE, E.A., ERICKSON, R.L. & DAVIDSON, C.L. 2002. Effect of surface treatments on the bond strength of glass ionomers to enamel. *Dental Materials,* 18, 454-62.

GLADYS, S., VAN MEERBEEK, B., LAMBRECHTS, P. & VANHERLE G. 1998. Marginal adaptation and retention of a glass-ionomer, resin-modified glass-ionomers and a polyacid-modified resin composite in cervical Class-V lesions. *Dental Materials,* 14, 294-306.

GOLDMAN, M., 1985. Fracture properties of composite and glass ionomer dental restorative materials. *Journal of Biomedical Materials Research*, 19,771-83.

GONZÁLEZ-LÓPEZ, S., VILCHEZ DÍAZ, M.A., DE HARO-GASQUET, F., CEBALLOS, L. & DE HARO-MUÑOZ, C. 2007. Cuspal flexure of teeth with composite restorations subjected to occlusal loading. *Journal of Adhesive Dentistry,* 9, 11-5.

GORACCI, C., SADEK, F.T., FABIANELLI, A., TAY, F.R. & FERRARI, M., 2005. Evaluation of the adhesion of fiber posts to intraradicular dentin. *Operative Dentistry-University of Washington*, 30, 627-35.

GORDAN, V.V., RILEY III, J.L., BLASER, P.K. AND MJÖR, I.A., 2006. 2-year clinical evaluation of alternative treatments to replacement of defective amalgam restorations. *Operative Dentistry*, 31, 418-25.

GORDAN, V.V., RILEY III, J.L., BLASER, P.K., MONDRAGON, E., GARVAN, C.W. & MJÖR, I.A., 2011. Alternative treatments to replacement of defective amalgam restorations: results of a seven-year clinical study. *The Journal of the American Dental Association*, 142, 842-9.

GORDAN, V.V., RILEY III, J.L., GERALDELI, S., RINDAL, D.B., QVIST, V., FELLOWS, J.L., KELLUM, H.P., GILBERT, G.H. & DENTAL PRACTICE-BASED RESEARCH NETWORK COLLABORATIVE GROUP, 2012. Repair or replacement of defective restorations by dentists in The Dental Practice-Based Research Network. *The Journal of the American Dental Association*, 143, 593-601.

GREEN, D., MACKENZIE, L. & BANERJEE, A., 2015. Minimally invasive long-term management of direct restorations: the ‘5 Rs’. *Dental Update*, 42, 413-26.

GREGORY, W.A., POUNDER, B. & BAKUS, E. 1990. Bond strengths of chemically dissimilar repaired composite resins. *Journal of Prosthetic Dentistry*, 64,664-8.

GUIDA, A., HILL, R. G., TOWLER, M. R. & ERAMO, S. 2002. Fluoride release from model glass ionomer cements. *Journal of Materials Science-Materials in Medicine,* 13**,** 645-9.

GUGGENBERGER, R., MAY, R. & STEFAN, K.P., 1998. New trends in glass-ionomer chemistry. *Biomaterials*, 19, 479-83.

GURGAN, S., KUTUK, Z.B., ERGIN, E., OZTAS, S.S. & CAKIR, F.Y., 2017. Clinical performance of a glass ionomer restorative system: a 6-year evaluation. *Clinical Oral Investigations*, 21, 2335-43.

HAFSHEJANI, T.M., ZAMANIAN, A., VENUGOPAL, J.R., REZVANI, Z., SEFAT, F., SAEB, M.R., VAHABI, H., ZARRINTAJ, P. & MOZAFARI, M., 2017. Antibacterial glass-ionomer cement restorative materials: A critical review on the current status of extended release formulations. *Journal of Controlled Release*, 262, 317-28.

HALLETT, K.B. & GARCIA-GODOY, F., 1993. Microleakage of resin-modified glass ionomer cement restorations: an in vitro study. *Dental Materials*, 9, 306-11.

HAMID, A., OKAMOTO, A., IWAKU, M. & HUME, W.R., 1998. Component release from light-activated glass ionomer and compomer cements. *Journal of oral rehabilitation*, *25*, 94-9.

HARASHIMA, I., HIRASAWA, T., TOMIOKA, K. & OKADA, J.I., 1988. Fractography of the bonding between light-cured resin and tooth substrates. *Dental materials journal*, 7, 151-159.

HARRIS, J.S., JACOBSEN, P.H. & O'DOHERTY, D.M., 1999. The effect of curing light intensity and test temperature on the dynamic mechanical properties of two polymer composites. *Journal of oral rehabilitation*, 26,635-9.

HASHEM, D.F., FOXTON, R., MANOHARAN, A., WATSON, T.F. & BANERJEE, A., 2014. The physical characteristics of resin composite–calcium silicate interface as part of a layered/laminate adhesive restoration. *Dental Materials*, 30, 343-9.

HASHIMOTO, M., OHNO, H., KAGA, M., ENDO, K., SANO, H. & OGUCHI, H. 2000. In vivo degradation of resin-dentin bonds in humans over 1 to 3 years. *Journal of Dental Researches*, 79, 1385-91.

HATIBOVIC-KOFMAN, S., KOCH, G. & EKSTRAND, J., 1997. Glass ionomer materials as a rechargeable fluoride-release system. *International Journal of Paediatric Dentistry*, 7, 65-73.

HATTAB, F.N. & AMIN, W.M. 2001. Fluoride release from glass ionomer restorative materials and the effects of surface coating. *Biomaterials,* 22, 1449-58

HICKEL, R. & VOSS, A., 1988. Long-term results with glass ionomer cements. *Deutsche Zahnarztliche Zeitschrift*, 43, 263-71.

HICKEL, R. & MANHART, J. 2001. Longevity of restorations in posterior teeth and reasons for failure. *Journal of Adhesive Dentistry*, 3, 45-64.

HICKEL, R., PESCHKE, A., TYAS, M., MJÖR, I., BAYNE, S., PETERS, M., HILLER, K.A., RANDALL, R., VANHERLE, G. & HEINTZE, S.D., 2010. FDI World Dental Federation: clinical criteria for the evaluation of direct and indirect restorations-update and clinical examples. *Clinical Oral Investigations*, 14, 349-66.

HIKITA, K., VAN MEERBEEK, B., DE MUNCK, J., IKEDA, T., VAN LANDUYT, K., MAIDA, T., LAMBRECHTS, P. & PEUMANS, M., 2007. Bonding effectiveness of adhesive luting agents to enamel and dentin. *Dental Materials*, 23, 71-80.

HINOURA, K., ONOSE, H., MOOREB, K. & PHILLIPSR, W. 1989. Effect of the bonding agent on the bond strength between glass-ionomer cement and composite resin. *Quintessence International*, 20, 31-5.

HINOURA, K., MIYAZAKI, M. & ONOSE, H., 1991. Dentin bond strength of light-cured glass-ionomer cements. *Journal of Dental Research*, 70, 1542-4.

HINOURA, K., ONOSE, H., MASUTANI, S., MATSUZAKI, T. & MOORE, B.K., 1993. Volumetric change of light cured glass ionomer in water. *Journal of Dental Research,* 72, 222.

HODGES, D.J., MANGUM, F.I. & WARD, M.T., 1995. Relationship between gap width and recurrent dental caries beneath occlusal margins of amalgam restorations. *Community Dentistry and Oral Epidemiology*, 23, 200-4.

HOLTAN, J.R., NYSTROM, G.P., OLIN, P.S., RUDNEY, J. & DOUGLAS, W.H., 1990. Bond strength of light-cured and two auto-cured glass-ionomerliners. *Journal of Dentistry*, 18, 271-5.

HONG, Y.W., KIM, J.H., LEE, B.H., LEE, Y.K., CHOI, B.J., LEE, J.H. & CHOI, H.J. 2008. The Effect of nano-sized β-tricalcium phosphate on remineralisation in glass ionomer dental luting cement. In *Key Engineering Materials*, 361, 861-4.

HÖRSTED-BINDSLEV, P. 1994. Fluoride release from alternative restorative materials. *Journal of Dentistry*, 22, 17-20.

HUA, H. & DUBÉ, M.A., 2001. Off-line monitoring of butyl acrylate, methyl methacrylate and vinyl acetate homo-and copolymerizations in toluene using ATR-FTIR spectroscopy. *Polymer*, *42*(14), pp.6009-6018.

IBARRA, G., JOHNSON, G.H., GEURTSEN, W. &VARGAS, M.A., 2007. Microleakage of porcelain veneer restorations bonded to enamel and dentin with a new self-adhesive resin-based dental cement. *Dental Materials*, 23, 218-25.

ILIE, N. & HICKEL, R., 2009. Investigations on mechanical behaviour of dental composites. *Clinical Oral Investigations*, 13, 427-38.

IMBERY, T.A., NAMBOODIRI, A., DUNCAN, A., AMOS, R., BEST, A.M. & MOON, P.C., 2013. Evaluating dentin surface treatments for resin-modified glass ionomer restorative materials. *Operative Dentistry*, 38, 429-38.

INOUE, S., VAN MEERBEEK, B., ABE, Y., YOSHIDA, Y., LAMBRECHTS, P., VANHERLE, G. & SANO, H., 2001. Effect of remaining dentin thickness and the use of conditioner on micro-tensile bond strength of a glass-ionomer adhesive. *Dental Materials*, 17, 445-55.

INOUE, S., KOSHIRO, K., YOSHIDA, Y., DE MUNCK, J., NAGAKANE, K., SUZUKI, K., SANO, H. & VAN MEERBEEK, B., 2005. Hydrolytic stability of self-etch adhesives bonded to dentin. *Journal of Dental Research*, 84, 1160-4.

IRIE, M., SUZUKI, K. & WATTS, D.C., 2002. Marginal gap formation of light-activated restorative materials: effects of immediate setting shrinkage and bond strength. *Dental Materials*, 18, 203-10.

ISO, 9917-1, 2007. Dentistry- water-based cements. Geneva: ISO

ISO, 9917-2, 2010. Dentistry - Water-based cements - Part 2: Resin-modified cements, Geneva: ISO.

ISO 4049. Dentistry-resin based dental fillings. International Organization for Standardization; 2009.

ISO/TS 11405. Dentistry-Testing the adhesion to tooth structure, 3rd edition, 2015

ITO, S., SAITO, T., TAY, F.R., CARVALHO, R.M., YOSHIYAMA, M. & PASHLEY, D.H. 2005. Water content and apparent stiffness of non-caries versus caries-affected human dentin. *Journal of Biomedical Material Research Part B Applied Biomaterials,* 72, 109-16.

ITOTA, T., NAKATSUKA, T., TANAKA, K., TASHIRO, Y., MCCABE, J.F. & YOSHIYAMA, M., 2010. Neutralizing effect by resin-based materials containing silane-coated glass fillers. *Dental Materials Journal*, 29, 362-68.

JAKUBIAK, J., ALLONAS, X., FOUASSIER, J.P., SIONKOWSKA, A., ANDRZEJEWSKA, E., LINDEN, L.Å. & RABEK, J.F., 2003. Camphorquinone–amines photoinitating systems for the initiation of free radical polymerization. *Polymer*, 44, 5219-5226.

JÄLEVIK, B., ODELIUS, H., DIETZ W. & NOREN, J. 2001. Secondary ion mass spectrometry and x-ray microanalysis of hypomineralized enamel in human permanent molars. *Archives of Oral Biology*, 46, 239-47.

JAMALUDDIN, A. & PEARSON, G.J., 1993. SEM analyses of repaired glass-ionomer cements. *Asian Journal of Aesthetic Dentistry*, 1, 19-23.

JIANGUO, L.I., YAJUAN, LIU, YUN LIU, RUNE SÖREMARK & FOLKE SUNDSTRÖM. 1996. Flexure strength of resin-modified glass-ionomer cements and their bond strength to dental composites. *Acta Odontologica Scandinavica*, 54, 55-8.

JIN, X.Z., HOMAEI, E., MATINLINNA, J.P. & TSOI, J.K.H., 2016. A new concept and finite-element study on dental bond strength tests. *Dental materials*, 32, e238-e250.

JOKSTAD, A., BAYNE, S., BLUNCK, U., TYAS, M. & WILSON, N., 2001. Quality of dental restorations FDI Commission Project 2-95. *International Dental Journal*, 51, 117-58.

JOKSTAD, A., 2016. Secondary caries and microleakage. *Dental Materials*, 32, 11-25.

JONES, F. H., HUTTON, B. M., HADLEY, P. C., ECCLES, A. J., STEELE, T. A., BILLINGTON, R. W. & PEARSON, G. J. 2003. Fluoride uptake by glass ionomer cements: a surface analysis approach. *Biomaterials,* 24**,** 107-19.

JUNIOR, S.A., FERRACANE, J.L. & DELLA BONA, Á. 2009. Influence of surface treatments on the bond strength of repaired resin composite restorative materials. *Dental Materials*, 25, 442-51.

KAKABOURA, A., ELIADES, G. & PALAGHIAS, G., 1996. An FTIR study on the setting mechanism of resin-modified glass ionomer restoratives. *Dental Materials*, 12, 173-8.

KALLIO, T.T., LASTUMA¨KI, T.M. & VALLITTU, P.K. 2001. Bonding of restorative and veneering composite resin to some polymeric composites. *Dental Materials*, 17, 80-6.

KANCHANAVASITA, W., ANSTICE, H.M. & PEARSON, G.J. 1997. Water sorption characteristics of resin-modified glass-ionomer cements. *Biomaterials*, 18, 343-9.

KANCHANAVASITA, W., ANSTICE, H.M. & PEARSON, G.J., 1998. Long-term surface micro-hardness of resin-modified glass ionomers. *Journal of Dentistry*, 26,707-12.

KANZOW, P., WIEGAND, A. & SCHWENDICKE, F., 2016. Cost-effectiveness of repairing versus replacing composite or amalgam restorations. *Journal of Dentistry*, 54, 41-7.

KANZOW, P., HOFFMANN, R., TSCHAMMLER, C., KRUPPA, J., RÖDIG, T. & WIEGAND, A., 2017. Attitudes, practice, and experience of German dentists regarding repair rest orations. *Clinical Oral Investigations*, 21, 1087-93.

KANZOW, P., WIEGAND, A., GOESTEMEYER, G. & SCHWENDICKE, F., 2018. Understanding the management and teaching of dental restoration repair: Systematic review and meta-analysis of surveys. *Journal of Dentistry*, 69, 1-21.

KAPLAN, A.E., WILLIAMS, J. BILLINGTON, R.W. & BRADEN, M. 2004. Effects of variation in particle size on biaxial flexural strength of two conventional glass-ionomer cements. *Journal of Oral Rehabilitation*, 31:373-8.

KATO, S., 1993. Fluoride release from light cured glass ionomer cement for restoration. *Journal of Dental Research,* 72 Spec issue: Abstract 945.

KEMAL, E., ADESANYA, K.O. & DEB, S., 2011. Phosphate based 2-hydroxyethyl methacrylate hydrogels for biomedical applications. *Journal of Materials Chemistry*, 21, 2237- 45.

KIDD, E.A. & O'HARA, J.W. 1990. The caries status of occlusal amalgam restorations with marginal defects. *Journal of Dental Research,* 69, 1275-7.

KIDD, E., JOYSTON-BECHAL, S. & BEIGHTON, D. 1993. Microbiological validation of assessments of caries activity during cavity preparation. *Caries Research,* 27**,** 402-08.

KIDD, E.A., JOYSTON-BECHAL, S. & BEIGHTON, D., 1995. Marginal ditching and staining as a predictor of secondary caries around amalgam restorations: a clinical and microbiological study. *Journal of Dental Research*, 74, 1206-11.

KILPATRICK, N.M., MCCABE, J.F. & MURRAY, J.J., 1994. Factors that influence the setting characteristics of encapsulated glass-ionomer cements. *Journal of Dentistry*, 22, 182-7.

KIM, Y.K., MAI, S., MAZZONI, A., LIU, Y., TEZVERGIL-MUTLUAY, A., TAKAHASHI, K., ZHANG, K., PASHLEY, D.H. & TAY, F.R., 2010a. Biomimetic remineralization as a progressive dehydration mechanism of collagen matrices–implications in the aging of resin-dentin bonds. *Acta Biomaterialia*, 6, 3729-39.

KIM, Y., YIU, C., KIM, J., GU, L., KIM, S., WELLER, R., PASHLEY, D. & TAY, F. 2010b. Failure of a glass ionomer to remineralize apatite-depleted dentin. *Journal of Dental Research*, 89, 230-35.

KIM, K.L., NAMGUNG, C. & CHO, B.H., 2013. The effect of clinical performance on the survival estimates of direct restorations. *Restorative Dentistry & Endodontics*, 38, 11-20.

KIMMES, N.S., BARKMEIER, W.W., ERICKSON, R.L. & LATTA, M.A. 2010. Adhesive Bond Strengths to Enamel and Dentin Using Recommended and Extended Treatment Times. *Operative Dentistry*, 35, 112-9.

KNIGHT, G.M., 1984. The use of adhesive materials in the conservative restoration of selected posterior teeth**.** *Australian dental journal,* 29, 324-31**.**

KNIGHT, G.M., 1992. The tunnel restoration–nine years of clinical experience using capsulated glass ionomer cements. Case report. *Australian Dental Journal*, 37,245-51.

KOIBUCHI, H., YASUDA, N. & NAKABAYASHI, N., 2001. Bonding to dentin with a self-etching primer: the effect of smear layers. *Dental Materials*, 17, 122-126.

KOKUBO, T. & TAKADAMA, H., 2006. How useful is SBF in predicting in vivo bone bioactivity? *Biomaterials*, 27, 2907-15.

KOPPERUD, S.E., TVEIT, A.B., GAARDEN, T., SANDVIK, L. & ESPELID, I., 2012. Longevity of posterior dental restorations and reasons for failure. *European journal of Oral Sciences*, 120, 539-48.

KOPPERUD, S.E., STAXRUD, F., ESPELID, I. & TVEIT, A.B., 2016. The post-amalgam era: Norwegian dentists' experiences with composite resins and repair of defective amalgam restorations. *International Journal of Environmental Research and Public Health*, 13, 441.

KRÄMER, N., KHAC, N-H., N., LÜCKER, S., STACHNISS V. & FRANKENBERGER, R. 2018. Bonding strategies for MIH-affected enamel and dentin. *Dental Materials,* 34, 331-40.

KUHN, E., REIS, A., CHIBINSKI, A.C.R. & WAMBIER, D.S., 2016. The influence of the lining material on the repair of the infected dentin in young permanent molars after restoration: A randomized clinical trial. *Journal of Conservative Dentistry,* 19, 516-21.

KUPER, N.K., VAN DE SANDE, F.H., OPDAM, N.J.M., BRONKHORST, E.M., DE SOET, J.J., CENCI, M.S. & HUYSMANS, M.C. 2015. Restoration materials and secondary caries using an in vitro biofilm model. *Journal of Dental Research*, 94, 62-8.

LABELLA, R., FRANKEL, N.T. & PEARSON, G.J., 1996, January. Long term flexural strength of aesthetic restoratives. *Journal of Dental Research*, 75,173.

LEELAWAT, C., SCHERER, W., CHANG, J., DAVID, S. & SCHULMAN, A., 1992. Addition of fresh amalgam to existing amalgam utilizing various adhesive liners: a SEM study. *Journal of Esthetic and Restorative Dentistry*, 4, 50-3.

LEE, H.S., BERG, J.H., GARCÍA-GODOY, F. & JANG, K.T., 2008. Long term evaluation of the remineralization of interproximal caries-like lesions adjacent to glass-ionomer restorations: A micro-CT study. *American Journal of Dentistry*, 21, 129-32.

LEWIS, S.M., COLEMAN, N.J., BOOTH, S.E. & NICHOLSON, J.W., 2013. Interaction of fluoride complexes derived from glass-ionomer cements with hydroxyapatite. Ceramics-Silikaty, 57,196-200.

LI, J., BEETZEN, M.V. & SUNDSTRÖM, F., 1995. Strength and setting behavior of resin-modified glass ionomer cements. *Acta Odontologica Scandinavica*, 53, 311-7.

LI, J., LIU, Y., LIU, Y., SÖREMARK, R. & SUNDSTRÖM, F., 1996. Flexure strength of resin-modified glass ionomer cements and their bond strength to dental composites. *Acta Odontologica Scandinavica*, 54, 55-8.

LIN, A., MCINTYRE, N.S. & DAVIDSON, R.D., 1992. Studies on the adhesion of glass-ionomer cements to dentin. *Journal of Dental Research*, 71, 1836-41.

LIMA, G.S., OGLIARI, F.A., DA SILVA, E.O., ELY, C., DEMARCO, F.F., CARRENO, N.L., PETZHOLD, C.L. & PIVA, E. 2008. Influence of water concentration in an experimental self-etching primer on the bond strength to dentin. *Journal of Adhesive Dentistry*, 10,167-72.

LIU, Y., TJÄDERHANE, L., BRESCHI, L., MAZZONI, A., LI, N., MAO, J., PASHLEY, D.H. & TAY, F.R., 2011. Limitations in bonding to dentin and experimental strategies to prevent bond degradation. *Journal of Dental Research*, 90, 953-68.

LLOYD, C. & MITCHELL, L. 1984. The fracture toughness of tooth coloured restorative materials. *Journal of Oral Rehabilitation,* 11, 257-72.

LOGUERCIO, A.D., REIS, A. & BALLESTER, R.Y., 2004. Polymerization shrinkage: effects of constraint and filling technique in composite restorations. *Dental Materials*, 20, 236-43.

LOHBAUER, U. 2010. Dental Glass Ionomer Cements as Permanent Filling Materials? Properties, Limitations and Future Trends. *Materials*, 3, 76-96.

LOOMANS, B.A.C., MINE, A., ROETERS, F.J.M., OPDAM, N.J.M., DE MUNCK, J., HUYSMANS, M.C.D.N.J.M. & VAN MEERBEEK, B., 2010. Hydrofluoric acid on dentin should be avoided. *Dental Materials*, 26, 643-9.

LOOMANS, B.A, CARDOSO, M.V., OPDAM, N.J., ROETERS, F.J., DE MUNCK, J., HUYSMANS, M.C. & VAN MEERBEEK, B., 2011a. Surface roughness of etched composite resin in light of composite repair. *Journal of Dentistry*, 39, 499-505.

LOOMANS, B.A.C., CARDOSO, M.V., ROETERS, F.J.M., OPDAM, N.J.M., DE MUNCK, J., HUYSMANS, M.C. & VAN MEERBEEK, B., 2011b. Is there one optimal repair technique for all composites?. *Dental Materials*, 27, 701-9.

LOVELL, L.G., NEWMAN, S.M. & BOWMAN, C.N., 1999. The effects of light intensity, temperature, and comonomer composition on the polymerization behavior of dimethacrylate dental resins. *Journal of Dental Research*, 78, 1469-76.

LUCAS, M.E., ARITA, K. & NISHINO, M. 2003.Toughness, bonding and fluoride-release properties of hydroxyapatite-added glass ionomer cement. *Biomaterials*, 24, 3787-94.

LUNG, C.Y.K. & MATINLINNA, J.P., 2012. Aspects of silane coupling agents and surface conditioning in dentistry: an overview. *Dental Materials*, 28, 467-77.

MACDONALD, S.A., SCHARDT, C.R., MASIELLO, D.J. & SIMMONS, J.H., 2000. Dispersion analysis of FTIR reflection measurements in silicate glasses. *Journal of non-crystalline solids*, 275, 72-82.

MACHADO, C., SANCHEZ, E., ALAPATI, S., SEGHI, R. & JOHNSTON, W. 2007. Shear bond strength of the amalgam-resin composite interface. *Operative Dentistry*, 32, 341-6.

MALACARNE, J., CARVALHO, R.M., MARIO, F., SVIZERO, N., PASHLEY, D.H., TAY, F.R., YIU, C.K. & DE OLIVEIRA CARRILHO, M.R., 2006. Water sorption/solubility of dental adhesive resins. *Dental Materials*, 22, 973-80.

MANEENUT, C., SAKOOLNAMARKA, R. & TYAS, M.J., 2010. The repair potential of resin-modified glass-ionomer cements. *Dental Materials*, 26,659-65.

MANEENUT, C., SAKOOLNAMARKA, R. & TYAS, M.J., 2011. The repair potential of resin composite materials. *Dental Materials*, 27, e20-e27.

MANHART, J., CHEN, H.Y., HAMM, G. & HICKEL, R., 2004. Review of the clinical survival of direct and indirect restorations in posterior teeth of the permanent dentition. *Operative Dentistry-University of Washington*, 29, 481-508.

MARQUEZAN, M., OSORIO, R., CIAMPONI, A.L. & TOLEDANO, M. 2010. Resistance to degradation of bonded restorations to simulated caries-affected primary dentin. *American Journal of Dentistry,* 23, 47-52.

MATSUMURA, H., TANAKA, T. & ATSUTA, M., 1997. Effect of acidic primers on bonding between stainless steel and auto-polymerizing methacrylic resins. *Journal of Dentistry*, 25, 285-90.

MATSUYA, S., MAEDA, T. & OHTA, M. 1996. IR and NMR analyses of hardening and maturation of glass-ionomer cement. *Journal of Dental Research,* 75**,** 1920-7.

MAYANAGI, G., IGARASHI, K., WASHIO, J., DOMON-TAWARAYA, H. & TAKAHASHI, N., 2014. Effect of fluoride-releasing restorative materials on bacteria-induced pH fall at the bacteria–material interface: An in vitro model study. *Journal of Dentistry,* 42, 15-20.

MCCABE, J.F. 1998. Resin-modified glass-ionomers. *Biomaterials*, 19, 521-7.

MCKENZIE, M.A., LINDEN, R.W.A. & NICHOLSON, J.W., 2003. The physical properties of conventional and resin-modified glass-ionomer dental cements stored in saliva, proprietary acidic beverages, saline and water. *Biomaterials*, 24, 4063-9.

MCKINNEY, J.E. & ANTONUCCI, J.M. 1986. Wear and microhardness of two experimental dental composites (abstract). *Journal of Dental Research*, 65, 848.

MCLEAN, J.M., 1985. The use of glass-ionomer cements in bonding composite resins to dentine. *British Dental Journal,* 158, 410-14.

MCLEAN, J.W., 1991. The clinical use of glass-ionomer cements-future and current developments. *Clinical Materials*, 7, 283-8.

MCLEAN, J.W. 1992. Clinical applications of glass-ionomer cement. *Operative Dentistry*, 17, 184.

MCLEAN, J.W.1994. Evolution of Glass-Ionomer Cements: A Personal View. *Journal of Esthetic and Restorative Dentistry,* 6, 195-206*.*

MCLEAN, J.W. 1994. Proposed nomenclature for glass-ionomer dental cements and related materials. *Quintessence International, 25*, 587-589.

MECHOLSKY, J.J. & BARKER, L.M., 1984. A chevron-notched specimen for fracture toughness measurements of ceramic-metal interfaces. In *Chevron-Notched Specimens: Testing and Stress Analysis*. ASTM International.

MESQUITA, M.F., DOMITTI, S.S., CONSANI, S. & DE GOES, M.F.1999. Effect of storage and acid etching on the tensile bond strength of composite resins to glass ionomer cement. *Brazilian Dental Journal,* 10, 5-9.

MITCHELL, C.A. & DOUGLAS, W.H., 1997. Comparison of the porosity of hand-mixed and capsulated glass-ionomer luting cements. *Biomaterials*, 18, 1127-31.

MITRA, S.B., 1991. Adhesion to dentin and physical properties of a light-cured glass-ionomer liner/base. *Journal of Dental Research*, 70, 72-4.

MITRA, S.B., 1992. Setting reaction of Vitrabond light cure glass ionomer liner/base. *Setting Mechanism of Dental Materials*. In: WATTS, D.C., editor. Setting Mechanisms of Dental Materials. Proceedings of a Symposium: June 30, Loch Lomond, Scotland, l-22.

MITRA, S.B. & KEDROWSKI, B.L., 1994. Long-term mechanical properties of glass ionomers. *Dental Materials*, 10, 78-82.

MITRA, S.B., LEE, C.Y., BUI, H.T., TANTBIROJN, D. & RUSIN, R.P., 2009. Long-term adhesion and mechanism of bonding of a paste-liquid resin-modified glass-ionomer. *Dental Materials*, 25, 459-66.

MITSUHASHI, A., HANAOKA, K., TERANAKA, T. 2003. Fracture toughness of resin-modified glass ionomer restorative materials: effect of powder/liquid ratio and powder particle size reduction on fracture toughness. *Dental Materials,* 19, 747-57.

MJÖR, I.A., 1996. Glass-ionomer cement restorations and secondary caries: A preliminary report. *Quintessence International*, 27, 171-4.

MJÖR, I.A., 1997. The reasons for replacement and the age of failed restorations in general dental practice. *Acta Odontologica Scandinavica*, 55, 58-63.

MJÖR, I.A., REEP, R.L., KUBILIS, P.S. & MONDRAGON, B.E., 1998. Change in size of replaced amalgam restorations: a methodological study. *Operative Dentistry*, 23, 272-7.

MJÖR, I., 1998. The location of clinically diagnosed secondary caries. *Quintessence International*, 29, 313-7.

MJÖR, I.A., MOORHEAD, J.E. & DAHL, J.E., 2000. Reasons for replacement of restorations in permanent teeth in general dental practice. *International Dental Journal*, 50, 361-6.

MJÖR, I.A. & GORDAN, VV. 2002. Failure, repair, refurbishing and longevity of restorations. *Operative Dentistry*, 27, 528-34.

MODENA, K., CASAS-APAYCO, L., ATTA, M., COSTA, C., HEBLING, J., SIPERT, C., NAVARRO, M. & SANTOS, C. 2009. Cytotoxicity and biocompatibility of direct and indirect pulp capping materials. *Journal of Applied Oral Science,* 17**,** 544-54.

MOLINA, G.F., CABRAL, R.J., MAZZOLA, I., LASCANO, L.B. & FRENCKEN, J.E. 2013. Mechanical performance of encapsulated restorative glass-ionomer cements for use with Atraumatic Restorative Treatment (ART). *Journal of Applied Oral Science*, 21, 243-49.

MOMOI, Y. & MCCABE, J.F., 1993. Fluoride release from light-activated glass ionomer restorative cements. *Dental Materials*, 9, 151-4.

MOMOI, Y. & MCCABE, J.F., 1994. Hygroscopic expansion of resin based composites during 6 months of water storage. *British Dental Journal*, 176, 91-6.

MOMOI, Y., HIROSAKI, K., KOHNO, A. & MCCABE, J.F., 1995. Flexural properties of resin-modified “hybrid” glass-ionomers in comparison with conventional acid-base glass-ionomers. *Dental Materials Journal*, 14, 109-19.

MONCADA, G., FERNÁNDEZ, E., MARTIN, J., ARANCIBIA, C., MJÖR, I. & GORDAN, V.V., 2008. Increasing the longevity of restorations by minimal intervention: a two-year clinical trial. *Operative Dentistry*, 33, 258-64.

MONCADA, G., MARTIN, J., FERNÁNDEZ, E., HEMPEL, M.C., MJÖR, I.A. & GORDAN, V.V., 2009. Sealing, refurbishment and repair of Class I and Class II defective restorations. *The Journal of the American Dental Association*, 140, 425-32.

MONCADA, G., VILDÓSOLA, P., FERNÁNDEZ, E., ESTAY, J., DE OLIVEIRA JÚNIOR, O.B., DE ANDRADE, M.F., MARTIN, J., MJÖR, I.A. & GORDAN, V.V., 2015. Longitudinal results of a 10-year clinical trial of repair of amalgam restorations. *Operative Dentistry*, 40, 34-43.

MOSHAVERINIA, A., ANSARI, S., MOSHAVERINIA, M., ROOHPOUR, N., DARR, J.A. & REHMAN, I. 2008. Effects of incorporation of hydroxyapatite and fluoroapatite nanobioceramics into conventional glass ionomer cements (GIC). *Acta Biomaterialia,* 4, 432-40.

MOSHAVERINIA, A., ROOHPOUR, N., ANSARI.S, MOSHAVERINIA, M., SCHRICKER, S., DARR.J. & REHMAN, I. 2009. Effects of N-vinylpyrrolidone (NVP) containing polyelectrolytes on surface properties of conventional glass-ionomer cements (GIC). *Dental Materials*, 25, 1240-7.

MOSHAVERINIA, A., ROOHPOUR, N., CHEE, W.W. & SCHRICKER, S.R. 2011. A review of powder modifications in conventional glass-ionomer dental cements. *Journal of Materials Chemistry*, *21,* 1319-28.

MOSHAVERINIA, A., ROOHPOUR, N., CHEEA, W.W. & SCHRICKER, S.R. 2012. A review of polyelectrolyte modifications in conventional glass-ionomer dental cements. *Journal of Material Chemistry,* 22, 2824-33.

MOUNT, G. J. & MAKINSON, O. F. 1982. Glass-ionomer restorative cements: clinical implications of the setting reaction. *Operative Dentistry,* 7**,** 134-41.

MOUNT, G.J., 1986. Longevity of glass ionomer cements. *Journal of Prosthetic Dentistry*, 55, 682-5.

MOUNT, G.J., 1989. Clinical requirements for a successful ‘sandwich’-dentine to glass-ionomer cement to composite resin. *Australian Dental Journal*, 34, 259-65.

MOUNT, G.J. &, HUME, W. R. 1998. *Preservation and restoration of tooth structure*. London: Mosby.

MOUNT, G.J. 2002. An atlas of glass-ionomer cements. A clinician’s guide. 3rd edn. London: Martin Dunitz Ltd.

MOUNT, G.J. 2005. Glass ionomer cements. In: Mount, G. J., Hume, W. R., editors. Preservation and restoration of tooth structure. 2nd Ed. Queensland: Knowledge Books and Software, 63-98.

MÜNCHOW, E.A., DA SILVA, A.F., DA SILVEIRA LIMA, G., WULFF, T., BARBOSA, M., OGLIARI, F.A. & PIVA, E., 2015. Polypropylene glycol phosphate methacrylate as an alternative acid-functional monomer on self-etching adhesives. *Journal of Dentistry*, 43, 94-102.

NAKABAYASHI, N., KOJIMA, K. & MASUHARA, E., 1982. The promotion of adhesion by the infiltration of monomers into tooth substrates. *Journal of Biomedical Materials Research*, 16,265-73.

NAKAJIMA, M., OGATA, M., OKUDA, M., TAGAMI, J., SANO, H. & PASHLEY, D.H. 1999. Bonding to caries-affected dentin using self-etching primers. *American Journal of Dentistry*, 12,309-14.

NAKAJIMA, M., KITASAKO, Y., OKUDA, M., FOXTON, R.M., & TAGAMI, J. 2005. Elemental distributions and microtensile bond strength of the adhesive interface to normal and caries-affected dentin. *Journal of Biomedical Materials Research Part B: Applied Biomaterials,* 72, 268-75.

NAKAYAMA, W.T., HALL, D.R., GRENOBLE, D.E. & KATZ, J.L. 1974. Elastic properties of dental resin restorative materials. *Journal of Dental Research*, 53, 1121-6.

NEGM, M.M., BEECH, D.R., GRANT, A.A. 1982. An evaluation of mechanical and adhesive properties of polycarboxylate and glass-ionomer cements. *Journal of Oral Rehabilitation*, 9,161-7.

NEZU, T. & WINNIK, F.M. 2000. Interaction of water-soluble collagen with poly (acrylic acid)*. Biomaterials*, 21, 415-9.

NGO, H., MOUNT, G.J. & PETERS, M.C.R.B., 1997. A study of glass ionomer cement and its interface with enamel and dentin using a low-temperature, high-resolution scanning electron microscopic technique. *Quintessence International,* 28**,** 63-9.

NGO, H.C., FRASER, M., MOUNT, G., MCINTYRE, J. & DO, L.G., 2002. Remineralisation of artificial carious dentine exposed to two glass-ionomers. *Journal of Dental Research*, 81 (special issue), A386.

NGO, H.C., MOUNT, G., MC INTYRE, J., TUISUVA, J. & VON DOUSSA, R.J., 2006. Chemical exchange between glass-ionomer restorations and residual carious dentine in permanent molars: an in vivo study. *Journal of Dentistry*, 34, 608-13.

NGO, H.C., MOUNT, G., MCINTYRE, J. & DO, L., 2011. An in vitro model for the study of chemical exchange between glass ionomer restorations and partially demineralized dentin using a minimally invasive restorative technique. *Journal of Dentistry*, *39*, S20-6.

NICHOLSON, J.W., BROOKMAN, P.J., LACY, O.M. & WILSON, A.D. 1988. Fourier transform infrared spectroscopic study of the role of tartaric acid in glass-ionomer dental cements. *Journal of Dental Research,* 67, 1451-4.

NICHOLSON, J.W., ANSTICE, H.M. & MCLEAN, J.W., 1992. A preliminary report on the effect of storage in water on the properties of commercial light-cured glass-ionomer cements. *British Dental Journal*, 173, 98-101.

NICHOLSON, J.W., HAWKINS, S.J. & SMITH, J.E. 1993. The incorporation of hydroxyapatite into glass-polyalkenoate (“glass-ionomer”) cements: a preliminary study. *Journal of Materials Science: Materials in Medicine,* 4, 418-21.

NICHOLSON, J., 1994. The development of modified glass-ionomer cements for dentistry. *Trends Polymer Science, 2*, 272-6.

NICHOLSON, J.W. & ANSTICE, H.M., 1994. The physical chemistry of light-curable glass-ionomers. *Journal of Materials Science: Materials in Medicine*, 5, 119-122.

NICHOLSON, J.W. 1998. Chemistry of glass-ionomer cements: a review. *Biomaterials* 19, 485-94.

NICHOLSON, J.W. & CZARNECKA, B., 2008. Kinetic studies of water uptake and loss in glass-ionomer cements. *Journal of Materials Science: Materials in Medicine*, 19, 1723-7.

NICHOLSON, J.W. & CZARNECKA, B., 2012. Maturation affects fluoride uptake by glass-ionomer dental cements. *Dental Materials*, 28, 1-5.

NICHOLSON, J. & CZARNECKA, B., 2016. *Materials for the Direct Restoration of Teeth*. Woodhead Publishing.

NICHOLSON, J.W., 2016. Adhesion of glass-ionomer cements to teeth: a review. *International Journal of Adhesion and Adhesives*, 69, 33-8.

NIU, L.N., ZHANG, W., PASHLEY, D.H., BRESCHI, L., MAO, J., CHEN, J.H. & TAY, F.R., 2014. Biomimetic remineralization of dentin. *Dental Materials*, 30, 77-96.

NOMOTO, R. & MCCABE, J.F. 2001. Effect of mixing methods on the compressive strength of glass ionomer cements. *Journal of Dentistry*, 29, 205-10.

NOMOTO, R., KOMORIYAMA, M., MCCABE, J.F. & HIRANO, S. 2004. Effect of mixing method on the porosity of encapsulated glass ionomer cement. *Dental Materials,* 20, 972-8.

NUTTELMAN, C.R., BENOIT, D.S., TRIPODI, M.C. & ANSETH, K.S., 2006. The effect of ethylene glycol methacrylate phosphate in PEG hydrogels on mineralization and viability of encapsulated hMSCs. *Biomaterials*, 27, 1377-86.

O'BRIEN, W.J. & RASMUSSEN, S.T., 1984. A critical appraisal of dental adhesion testing. In *Adhesive Joints*. Springer, Boston, MA, 289-305.

OGAWA, K., YAMASHITA, Y., ICHIJO, T. & FUSAYAMA, T. 1983. The Ultrastructure and Hardness of the Transparent of Human Carious Dentin. *Journal of Dental Research,* 62**,** 7-10.

OILO, G., 1987. Adhesion of dental materials to dentin debonding tests. *Dentine and dentin reactions in the oral cavity*, In: Thylstrup A, Leach SA, Qvist V, eds. Oxford: IRL Press Ltd., 219-224.

OLIVA, A., DELLA RAGIONE, F., SALERNO, A., RICCIO, V., TARTARO, G., COZZOLINO, A., D'AMATO, S., PONTONI, G. & ZAPPIA, V., 1996. Biocompatibility studies on glass ionomer cements by primary cultures of human osteoblasts. *Biomaterials*, 17, 1351-56.

OPDAM, N.J., BRONKHORST, E.M., ROETERS, J.M., & LOOMANS, B.A. 2007. A retrospective clinical study on longevity of posterior composite and amalgam restorations. *Dental Materials*, 23, 2-8.

OPDAM, N.J., BRONKHORST, E.M., LOOMANS, B.A. & HUYSMANS, M.C.D., 2012. Longevity of repaired restorations: a practice based study. *Journal of Dentistry*, 40, 829-35.

OPDAM, N.J.M., VAN DE SANDE, F.H., BRONKHORST, E., CENCI, M.S., BOTTENBERG, P., PALLESEN, U., GAENGLER, P., LINDBERG, A., HUYSMANS, M.C. & VAN DIJKEN, J.W., 2014. Longevity of posterior composite restorations: a systematic review and meta-analysis. *Journal of Dental Research*, 93, 943-9.

OSORIO, R., OSORIO, E., AGUILERA, F.S., TAY, F.R., PINTO, A. & TOLEDANO, M. 2010. Influence of application parameters on bond strength of an ‘‘all in one’’ water-based self-etching primer/adhesive after 6 and 12 months of water aging. *Odontology,* 98,117-25.

OSORIO, E., FAGUNDES, T., NAVARRO, M.F., ZANOTTO, E.D., PEITL, O., OSORIO, R., TOLEDANO- OSORIO, M. & TOLEDANO. 2015. A novel bioactive agent improves adhesion of resin-modified glass ionomer to dentin. *Journal of Adhesive Science and Technology*, 29, 1543-52.

OUYANG, Z., SNECKENBERGER, S.K., KAO, E.C., CULBERTSON, B.M. & JAGODZINSKI, P.W., 1999. New method for monitoring the reaction of glass-ionomer cements: A spectroscopic study of the effects of polyacid structure on the decomposition of calcium aluminosilicate glasses. *Applied Spectroscopy*, 53, 297-301.

ÖZCAN, M., ALANDER, P., VALLITTU, P.K., HUYSMANS, M.C. & KALK, W., 2005. Effect of three surface conditioning methods to improve bond strength of particulate filler resin composites. *Journal of Materials Science: Materials in Medicine*, 16, 21-7.

ÖZCAN, M., SCHOONBEEK, G., GÖKÇE, B., CÖMLEKOGLU, E. & DÜNDAR, M., 2010. Bond strength comparison of amalgam repair protocols using resin composite in situations with and without dentin exposure. *Operative Dentistry*, 35, 655-62.

ÖZCAN, M., KOOLMAN, C., ALADAG, A. & DUNDAR, M. 2011. Effects of different surface conditioning methods on the bond strength of composite resin to amalgam. *Operative Dentistry*, 36,318-23.

ÖZCAN, M. & SALIHOĞLU-YENER E. 2011. A technical report on repair of amalgam-dentin complex. *Operative Dentistry, 36*, 563−66.

ÖZCAN, M., CORAZZA, P.H., MAROCHO, S.M., BARBOSA, S.H. & BOTTINO, M.A. 2013. Repair bond strength of microhybrid, nanohybrid and nanofilled resin composites: effect of substrate resin type, surface conditioning and ageing. *Clinical Oral Investigations,* 17, 1751-58.

ÖZER, L. & THYLSTRUP, A., 1995. What is known about caries in relation to restorations as a reason for replacement? A review. *Advances in Dental Research*, 9, 394-402.

ÖZER, F., UNLU, N., OZTURK, B. & D SENGUN, A., 2002. Amalgam repair: evaluation of bond strength and microleakage. *Operative Dentistry*, 27, 199-203.

PADDON, J. M. & WILSON, A. D. 1976. Stress relaxation studies on dental materials. 1. Dental cements. *Journal of Dentistry,* 4, 183-9.

PALIN, W.M., FLEMING, G.J.P., BURKE, F.J.T., MARQUIS, P.M. & RANDALL, R.C., 2005a. The influence of short and medium-term water immersion on the hydrolytic stability of novel low-shrink dental composites. *Dental Materials*, 21, 852-863.

PALIN, W.M., FLEMING, G.J., NATHWANI, H., BURKE, F.T. & RANDALL, R.C., 2005b. In vitro cuspal deflection and microleakage of maxillary premolars restored with novel low-shrink dental composites. *Dental Materials*, 21, 324-35.

PALMA-DIBB, R.G., DE CASTRO, C.G., RAMOS, R.P., CHIMELLO, D.T. &CHINELATTI, M.A., 2003. Bond strength of glass-ionomer cements to caries-affected dentin. *Journal of Adhesive Dentistry*, 5, 52-62.

PAMEIJER, C.H., GARCIA-GODOY, F., MORROW, B.R. & JEFFERIES, S.R. 2015. Flexural strength and flexural fatigue properties of resin-modified glass ionomers. *Journal of Clinical Dentistry*, 26, 23-7.

PAPAGIANNOULIS, L., KAKABOURA, A. & ELIADES, G., 2002. In vivo vs in vitro anticariogenic behavior of glass-ionomer and resin composite restorative materials. *Dental Materials*, 18, 561-9.

PAPACCHINI, F., MONTICELLI, F., RADOVIC, I., CHIEFFI, N., GORACCI, C., TAY, F.R., POLIMENI, A. & FERRARI, M., 2007a. The application of hydrogen peroxide in composite repair. *Journal of Biomedical Materials Research Part B: Applied Biomaterials,* 82, 298-304.

PAPACCHINI, F., MAGNI, E., RADOVIC, I., MAZZITELLI, C., MONTICELLI, F., GORACCI, C., POLIMENI, A. & FERRARI, M., 2007b. Effect of intermediate agents and pre-heating of repairing resin on composite-repair bonds. *Operative Dentistry*, 32, 363-71.

PAPACCHINI, F., DALL OCA, S., CHIEFFI, N., GORACCI, C., SADEK, F.T., SUH, B.I., TAY, F.R. & FERRARI, M., 2007c. Composite-to-composite microtensile bond strength in the repair of a microfilled hybrid resin: effect of surface treatment and oxygen inhibition. *Journal of Adhesive Dentistry*, 9, 25-31.

PAPAGIANNOULIS, L., ELIADES, G. & LEKKA, M., 1990. Etched glass-ionomerliners: surface properties and interfacial profile with composite resins. *Journal of Oral Rehabilitation*, 17, 25-36.

PARKER, S. & BRADEN, M., 1989. Water absorption of methacrylate soft lining materials. *Biomaterials*, *10*, 91-5.

PASCHOAL, M.A., GURGEL, C.V., RIOS, D., MAGALHÃES, A.C., BUZALAF, M.A., MACHADO, M.A. 2011. Fluoride Release Profile of a Nanofilled Resin-Modified Glass Ionomer Cement. *Brazilian Dental Journal,* 22, 275-79.

PASHLEY, D.H., SANO, H., CIUCCHI, B., YOSHIYAMA, M. & CARVALHO, R.M., 1995. Adhesion testing of dentin bonding agents: a review. *Dental Materials*, 11, 117-25.

PASHLEY, D.H. & CARVALHO, R.M. 1997. Dentine permeability and dentine adhesion. *Journal of Dentistry*, 5,355-72.

PASHLEY, D.H., CARVALHO, R.M., SANO, H., NAKAJIMA, M., YOSHIYAMA, M., SHONO, Y., FERNANDES, C.A. & TAY, F., 1999. The microtensile bond test: A review. *Journal of Adhesive Dentistry*, 1, 299-309.

PERDIGÃO, J. & SWIFT JR, E.J., 2015. Universal adhesives. *Journal of Esthetic and Restorative Dentistry*, 27,331-4.

PEARSON, G.J., BOWEN, G., JACOBSEN, P. & ATKINSON, A.S., 1989. The flexural strength of repaired glass-ionomer cements. *Dental Materials*, 5, 10-2.

PEARSON, G. J. & ATKINSON, A.S. 1991. Long-term flexural strength of glass ionomer cements. *Biomaterials*, 12,658-60.

PERDIGÃO, J., DUTRA-CORRÊA, M., SARACENI, C.H.C., CIARAMICOLI, M.T., KIYAN, V.H. & QUEIROZ, C.S., 2012. Randomized clinical trial of four adhesion strategies: 18-month results. *Operative Dentistry*, 37, 3-11.

PEREIRA, P.N., YAMADA, T., TEI, R. & TAGAMI, J., 1997. Bond strength and interface micromorphology of an improved resin-modified glass ionomer cement. *American Journal of Dentistry*, 10, 128-32.

PEREIRA, L.C.G., NUNES, M.C.P., DIBB, R.G.P., POWERS, J.M., ROULET, J.F. & DE LIMA NAVARRO, M.F., 2002. Mechanical properties and bond strength of glass-ionomer cements. *Journal of Adhesive Dentistry*, 4, 73-80.

PETERSEN, P.E., BOURGEOIS, D., OGAWA, H., ESTUPINAN-DAY, S. & NDIAYE, C. 2005. The global burden of oral diseases and risks to oral health. *Bull World Health Organ*, 83, 661-9.

PEUMANS, M., KANUMILLI, P., DE MUNCK, J., VAN LANDUYT, K., LAMBRECHTS, P. & MEERBEEK, B., 2005. Clinical effectiveness of contemporary adhesives: a systematic review of current clinical trials. *Dental Materials*, 21, 864-81.

PEUMANS, M., DE MUNCK, J., VAN LANDUYT, K.L., POITEVIN, A., LAMBRECHTS, P. & VAN MEERBEEK, B., 2010. Eight-year clinical evaluation of a 2-step self-etch adhesive with and without selective enamel etching. *Dental Materials*, 26, 1176-84.

PEUTZFELDT, A. & ASMUSSEN, E. 1992. Modulus of resilience as predictor for clinical wear of restorative resins, *Dental Materials,* 8,146-8.

PEUTZFELDT, A., GARCÍA-GODOY, F. & ASMUSSEN, E.1997. Surface hardness and wear of glass ionomers and compomers, *American Journal of Dentistry*, 10, 15-7.

PIANELLI, C., DEVAUX, J., BEBELMAN, S. & LELOUP, G. 1999. The micro-Raman spectroscopy, a useful tool to determine the degree of conversion of light-activated composite resins. *Journal of Biomedical Materials Research Part A,* 48, 675-81.

PILO, R., BROSH, T., SHAPINKO, E. & DODIUK, H. 1996. Long term durability of adhesive systems bonded to fresh amalgam. *Journal of Prosthetic Dentistry,* 76, 432–6.

PILO, R., NISSAN, J., SHAFIR, H., SHAPIRA, G., ALTER, E. & BROSH, T., 2012. The influence of long term water immersion on shear bond strength of amalgam repaired by resin composite and mediated by adhesives or resin modified glass ionomers. *Journal of Dentistry*, 40, 594-602.

PIWOWARCZYK, A., LAUER, H.C. & SORENSEN, J.A., 2005. Microleakage of various cementing agents for full cast crowns. *Dental Materials*, 21, 445-53.

PLACIDO, E., MEIRA, J.B., LIMA, R.G., MUENCH, A., DE SOUZA, R.M. & BALLESTER, R.Y., 2007. Shear versus micro-shear bond strength test: a finite element stress analysis. *Dental Materials*, 23, 1086-92

PLASMANS, P.J.J.M., CREUGERS, N.H.J. & D MULDER, J., 1998. Long-term survival of extensive amalgam restorations. *Journal of Dental Research*, 77, 453-60.

POPOFF, D.A.V., GONÇALVES, F.S., MAGALHÃES, C.S., MOREIRA, A.N., FERREIRA, R.C. & MJÖR, I.A., 2011. Repair of amalgam restorations with composite resin and bonded amalgam: A microleakage study. *Indian Journal of Dental Research*, 22, 799-803.

POWERS, J. M. & SAKAGUCHI, R. L. 2006. *Craig's restorative dental materials.* Elsevier Mosby, 12th ed. London.

POWIS, D.R., FOLLERAS, T., MERSON, S.A. & WILSON, A.D. 1982. Improved adhesion of a glass ionomer cement to dentin and enamel. *Journal of Dental Research,* 61, 1416-22.

PROSSER, H.J., POWIS, D.R., BRANT, P. & WILSON, A.D., 1984. Characterization of glass-ionomer cements 7. The physical properties of current materials. *Journal of Dentistry*, 12, 231-40.

PROSSER, H.J., POWIS, D.R. & WILSON, A.D. 1986. Glass-ionomer cements of improved flexural strength. *Journal of Dental Research,* 65,146-48.

PUCKETT, A.D., BENNETT, B. &FITCHIE, J. 1995. Effect of delayed photocuring of hybrid glass ionomer cements. *Journal of Dental Research*, 74, 38.

QVIST, V., LAURBERG, L., POULSEN, A. & TEGLERS, P.T. 2004. Eight-year study on conventional glass ionomer and amalgam restorations in primary teeth. *Acta Odontologica Scandinavica*, 62, 37-45.

RAI, N., NAIK, R., GUPTA, R., SHETTY, S. & SINGH, A., 2017. Evaluating the effect of different conditioning agents on the shear bond strength of resin-modified glass ionomers. *Contemporary clinical dentistry*, 8, 604-12.

RAMASETTY, P.A., BHAT, K.Y.M. & PRASANNA, M., 2014. Comparative evaluation of remineralization, fluoride release and physical properties of conventional GIC following incorporation of 1% and 2% zinc acetate: An in vitro study. *International Journal of Oral Health Sciences*, 4, 4-12.

RANDALL, R.C. & WILSON, N.H.F., 1999. Glass-ionomer restoratives: a systematic review of a secondary caries treatment effect. *Journal of Dental Rresearch*, 78, 628-37.

RAMOS, J.C. & PERDIGAO, J., 1997. Bond strengths and SEM morphology of dentin-amalgam adhesives. *American Journal of Dentistry*, 10, 152-8.

REIS, A.F., OLIVEIRA, M.T., GIANNINI, M., DE GOES, M.F. & RUEGGEBERG F.A. 2003. The effect of organic solvents on one bottle adhesives' bond strength to enamel and dentin. *Operative Dentistry,* 28, 700-6.

REY, C., SHIMIZU, M., COLLINS, B. & GLIMCHER, M.J., 1990. Resolution-enhanced Fourier transform infrared spectroscopy study of the environment of phosphate ions in the early deposits of a solid phase of calcium-phosphate in bone and enamel, and their evolution with age. I: Investigations in thev4 PO_4 domain domain. *Calcified Tissue International*, 46, 384-94.

RIGGS, P.D., PARKER, S., BRADEN, M. & KALACHANDRA, S., 2001. Development of novel elastomer/methacrylate monomer soft lining materials. *Journal of Materials Science: Materials in Medicine*, *12*, 359-64

ROBERTELLO, F.J., COFFEY, J.P., LYNDE, T.A. & KING, P., 1999. Fluoride release of glass ionomer-based luting cements in vitro. *The Journal of Prosthetic Dentistry*, 82, 172-6.

RODRIGUES, C.C., BOLINELLI, A., SEABRA, B.G.M., FRANCA, M.T.C. & NAVARRO, M.F.L., 1998. Six months evaluation of ART technique using Fuji IX and Fuji Plus. *Journal of Dental Research*, 77, 636.

ROFFEY, G.C. 1985. Photopolymerization of surface coatings, 2nd ed. Norwich: John Wiley and Sons, 179-201.

ROGGENKAMP, C.L., BERRY, F.A. & LU, H., 2010. In vitro bond strengths of amalgam added to existing amalgams. *Operative Dentistry*, 35, 314-23.

ROJO, L., VA'ZQUEZ, B., ROMA´N, J. & DEB, S. 2008. Eugenol functionalized poly (acrylic acid) derivatives in the formation of glass-ionomer cements. *Dental Materials,* 24, 1709-16.

ROSENTRITT, M., BEHR, M., LANG, R. & HANDEL, G., 2004. Influence of cement type on the marginal adaptation of all-ceramic MOD inlays. *Dental Materials*, 20, 463-9.

RUEGGEBERG, F.A., HASHINGER, D.T. & FAIRHURST, C.W., 1990. Calibration of FTIR conversion analysis of contemporary dental resin composites. *Dental Materials*, 6, 241-49.

RUENGRUNGSOM, C., PALAMARA, J.E. & BURROW, M.F., 2018. Comparison of ART and conventional techniques on clinical performance of glass-ionomer cement restorations in load bearing areas of permanent and primary dentitions: A systematic review. *Journal of Dentistry*. https://doi.org/10.1016/j.jdent.2018.07.008.

RUSZ, J.E., ANTONUCCI, J.M., EICHMILLER, F. & ANDERSON, M.H. 1992. Adhesive properties of modified glass ionomer cements. *Dental Materials*, 8, 31-6.

SAITO, T., TOYOOKA, H., ITO, S. & CRENSHAW, M.A., 2003. In vitro study of remineralization of dentin: effects of ions on mineral induction by decalcified dentin matrix. *Caries research*, 37, 445-9.

SALZ, U., MUCKE, A., ZIMMERMANN, J., TAY, F.R. & PASHLEY, D.H. 2006. pKa value and buffering capacity of acidic monomers commonly used in self-etching primers. *Journal of Adhesive Dentistry,* 8,143-50.

SAMUEL, S.P., LI, S., MUKHERJEE, I., GUO, Y., PATEL, A.C., BARAN, G. & WEI, Y., 2009. Mechanical properties of experimental dental composites containing a combination of mesoporous and nonporous spherical silica as fillers. *Dental materials*, 25, 296-301.

SANO, H., SHONO, T., SONODA, H., TAKATSU, T., CIUCCHI, B., CARVALHO, R. & PASHLEY, D.H., 1994. Relationship between surface area for adhesion and tensile bond strength-evaluation of a micro-tensile bond test. *Dental Materials*, 10, 236-40.

SANTERRE, J.P., SHAJII, L. & LEUNG, B.W. 2001. Relation of dental composite formulations to their degradation and the release of hydrolyzed polymeric-resin-derived products. *Critical Review of Oral and Biological Medicine*, 12,136-51.

SARACOGLU, A., ÖZCAN, M., KUMBULOGLU, O. & TURKUN, M., 2011. Adhesion of resin composite to hydrofluoric acid-exposed enamel and dentin in repair protocols. *Operative ddentistry*, 36, 545-53.

SASKALAUSKAITE, E., TAM, L.E. & MCCOMB, D., 2008. Flexural strength, elastic modulus, and pH profile of self-etch resin luting cements. *Journal of Prosthodontics*, 17, 262-8.

SATTABANASUK, V., SHIMADA, Y. & TAGAMI, J. 2005. Bonding of resin to artificially carious dentin. *Journal of Adhesive Dentistry*, 7, 183-192.

SCHMITT, W., PURRMANN, R., JOCHUM, P. & GASSER, O. 1983. Calcium depleted aluminium fluorosilicate glass powder for use in dental or bone cements. US Patent 4, 376, 835.

SCHOLTANUS, J.D. & HUYSMANS, M.C.D., 2007. Clinical failure of class-II restorations of a highly viscous glass-ionomer material over a 6-year period: a retrospective study. *Journal of Dentistry*, 35, 156-62.

SELWITZ, R.H., ISMAIL, A.I. & PITTS, N.B., 2007. Dental caries. *The Lancet*, 369, 51-9.

SENNOU, H.E., LEBUGLE, A.A. & GRÉGOIRE, G.L., 1999. X-ray photoelectron spectroscopy study of the dentin–glass ionomer cement interface. *Dental Materials*, 15, 229-37.

SEPPÄ, L., KORHONEN, A. & NUUTINEN, A., 1995. Inhibitory effect on S. mutans by fluoride-treated conventional and resin-reinforced glass ionomer cements. *European Journal of Oral Sciences*, 103, 182-185.

SEZINANDO, A., LUQUE-MARTINEZ, I., MUÑOZ, M.A., REIS, A., LOGUERCIO, A. D. & PERDIGÃO, J. 2015. Influence of a hydrophobic resin coating on the immediate and 6-month dentin bonding of three universal adhesives. *Dental Materials,* 31, 236-46.

SHAHDAD, S.A. & KENNEDY, J.G., 1998. Bond strength of repaired anterior composite resins: an it>/it> study1. *Journal of dentistry*, 26, 685-694.

SHAHID, S., BILLINGTON, R.W. & PEARSON, G.J., 2008. The role of glass composition in the behaviour of glass acetic acid and glass lactic acid cements. *Journal of Materials Science: Materials in Medicine*, 19, 541-5.

SHAHID, S., HASSAN, U., BILLINGTON, R.W., HILL, R.G. & ANDERSON, P., 2014. Glass ionomer cements: effect of strontium substitution on esthetics, radiopacity and fluoride release. *Dental Materials*, 30, 308-13.

SHAFFER, R.A., CHARLTON, D.G. & HERMESCH, C.B., 1998. Repairability of three resin-modified glass-ionomer restorative materials. *Operative Dentistry*, 23,168-72.

SHARIF, M.O., CATLEUGH, M., MERRY, A., TICKLE, M., DUNNE, S.M., BRUNTON, P. & AGGARWAL, V.R., 2010a. Replacement versus repair of defective restorations in adults: resin composite. *Australian Dental Journal*, 55, 351-52.

SHARIF, M.O., FEDOROWICZ, Z., TICKLE, M. & BRUNTON, P.A., 2010b. Repair or replacement of restorations: do we accept built in obsolescence or do we improve the evidence?. *British Dental Journal*, 209, 171-4.

SHEN, C.: Dental cements, in: ANUSAVICE, K.J. (ed): Phillip's Science of Dental Materials (ed 11). Philadelphia, PA, Saunders, 2003, 443-494

SHEN, C., SPEIGEL, J. AND MJÖR, I.A., 2006. Repair strength of dental amalgams. *Operative Dentistry*, 31, 122-6.

SHETTY, D.K., ROSENFIELD, A.R., DUCKWORTH, W.H. & HELD, P. 1983. A biaxial-flexure test for evaluating ceramic strengths. *Journal of the American Ceramic Society*, 66, 36-42.

SHONO, T. 1995. Pulpal response to light-cured restorative glass polyalkenoate cements, and ultrastructure of cement dentin interface. *Japanese Journal of Conservative Dentistry,* 38, 514-48.

SHONO, Y., OGAWA, T., TERASHITA, M., CARVALHO, R.M., PASHLEY, E.L. & PASHLEY, D.H., 1999. Regional measurement of resin-dentin bonding as an array. *Journal of Dental Research*, 78, 699-705.

SIDHU, S.K., 1993. A comparative analysis of techniques of restoring cervical lesions. *Quintessence International*, 24, 553-9.

SIDHU, S.K., 1994. Fluid permeability and other interfacial characteristics of light-cured glass-ionomer restorations. *Journal of Dental Research*, *73*, 183.

SIDHU, S.K. & WATSON, T.F. 1995. Resin-modified glass-ionomer materials. A status report for the American. *American Journal of Dentistry,* 8, 59-67.

SIDHU, S.K. & WATSON, T.E., 1998. Interfacial characteristics of resin-modified glass-ionomer materials: a study on fluid permeability using confocal fluorescence microscopy. *Journal of Dental Research*, 77, 1749-59.

SIDHU, S.K., PILECKI, P., CHENG, P.C. & WATSON, T.F., 2002. The morphology and stability of resin-modified glass-ionomer adhesive at the dentin/resin-based composite interface. *American Journal of Dentistry*, 15, 129-36.

SIDHU, S.K., 2010. Clinical evaluations of resin-modified glass-ionomer restorations. *Dental Materials*, 26, 7-12.

SIDHU, S.K. 2011. Glass-ionomer cement restorative materials: a sticky subject?, *Australian Dental Journal,* 56, 23-30.

SINDEL, J., FRANKENBERGER, R., KRÄMER, N. & PETSCHELT, A., 1999. Crack formation of all-ceramic crowns dependent on different core build-up and luting materials1. *Journal of Dentistry*, 27, 175-81.

SMALES, R.J. & HAWTHORNE, W.S., 2004. Long-term survival of repaired amalgams, recemented crowns and gold castings. *Operative Dentistry-University of Washington*, 29, 249-53.

SMALES, R.J., NGO, H.C., YIP, K.H. & YU, C., 2005. Clinical effects of glass-ionomer restorations on residual carious dentin in primary molars. *American Journal of Dentistry*, 18, 188-93.

SMITH, D.C. & COOPER, W.E., 1971. The determination of shear strength. A method using a micro-punch apparatus. *British dental journal*, 130,333.

SMITH, E.D.K. & MARTIN, F.E., 1992. Microleakage of glass ionomer/composite resin restorations: a laboratory study. The influence of glass ionomer cement. *Australian Dental Journal*, 37, 23-30.

SMITH, D. C. 1998. Development of glass-ionomer cement systems. *Biomaterials,* 19, 467-78.

SNEED, W.D. & LOOPER, S.W., 1985. Shear bond strength of a composite resin to an etched glass ionomer. *Dental Materials*, 1,127-8.

SOUZA, M., RETIEF, D.H. & RUSSELL, C.M. 1994. Laboratory evaluation of phosphate ester bonding agents. *American Journal of Dentistry*, 7, 67-73.

SÖDERHOLM, K.J., ZIGAN, M., RAGAN, M., FISCHLSCHWEIGER, W. & BERGMAN, M., 1984. Hydrolytic degradation of dental composites. *Journal of Dental Research*, 63, 1248-54.

SÖDERHOLM, K.J., ANTONSON, D.E. & FISCHLSCHWEIGER, W., 1989. Correlation between marginal discrepancies at the amalgam/tooth interface and recurrent caries. *Quality evaluation of dental restorations*: criteria for placement and replacement. ANUSAVICE KJ, editor. Chicago, IL: Quintessence Publishing Co., Inc., pp. 95-110.

SÖDERHOLM, K.J., 1991. Correlation of in vivo and in vitro performance of adhesive restorative materials: a report of the ASC MD156 Task Group on Test Methods for the Adhesion of Restorative Materials. *Dental Materials*, 7, 74-83.

SÖDERHOLM, K.J., TYAS, M.J. & JOKSTAD, A., 1998. Determinants of quality in operative dentistry. *Critical Reviews in Oral Biology & Medicine*, 9, 464-79.

SÖDERHOLM, K.J., GERALDELI, S. & SHEN, C., 2012. What do microtensile bond strength values of adhesives mean? *Journal of Adhesive Dentistry*, 14, 307-14.

SOBRINHO, L.C., DE GOES, M.F., CONSANI, S., SINHORETI, M.A.C. & KNOWLES, J.C., 2000. Correlation between light intensity and exposure time on the hardness of composite resin. *Journal of Materials Science: Materials in Medicine*, 11, 361-4.

SPENCER, P., WANG, Y., WALKER, M. P., SWAFFORD, J. R. 2001. Molecular structure of acid-etched dentin smear layers–in situ study. *Journal of Dental Research,* 80, 1802-7.

SPINELL, T., SCHEDLE, A. & WATTS, D.C., 2009. Polymerization shrinkage kinetics of dimethacrylate resin-cements. *Dental Materials*, 25, 1058-1066.

STANCU, I.C., FILMON, R., CINCU, C., MARCULESCU, B., ZAHARIA, C., TOURMEN, Y., BASLE, M.F. & CHAPPARD, D., 2004. Synthesis of methacryloyloxyethyl phosphate copolymers and in vitro calcification capacity. *Biomaterials*, 25, 205-13.

STANISLAWSKI L, DANIAU X, LAUTI´E A, GOLDBERG M. 1999. Factors responsible for pulp cytotoxicity induced by resin-modified glass ionomer cements. *Journal of Biomedical Material Research; Applied Biomaterial,* 48, 277-88.

STAXRUD, F. & DAHL, J.E., 2011. Role of bonding agents in the repair of composite resin restorations. *European Journal of Oral Sciences*, 119, 316-22.

STAXRUD, F. & DAHL, J.E., 2015. Silanising agents promote resin-composite repair. *International Dental Journal*, 65, 311-5.

SUDSANGIAM, S. & VAN NOORT, R., 1999. Do dentin bond strength tests serve a useful purpose? *Journal of Adhesive Dentistry*, 1, 57-67.

SUH, B.I., FENG, L., PASHLEY, D.H. & TAY, F.R., 2003. Factors contributing to the incompatibility between simplified-step adhesives and chemically-cured or dual-cured composites. Part III. Effect of acidic resin monomers. *Journal of Adhesive Dentistry*, 5, 267-82.

SULIMAN, A.A., SCHULEIN, T.M., BOYER, D.B. & KOHOUT, F.J., 1989. The effects of etching and rinsing times and salivary contamination on etched glass-ionomer cement bonded to resin composites. *Dental Materials*, 5, 171-5.

SUZUKI, S., WHITTAKER, M.R., GRØNDAHL, L., MONTEIRO, M.J. & WENTRUP-BYRNE, E., 2006. Synthesis of soluble phosphate polymers by RAFT and their in vitro mineralization. *Biomacromolecules*, 7, 3178-87.

SWIFT JR, E.J., PAWLUS, M.A., VARGAS, M.A. & FORTIN, D., 1995. Depth of cure of resin-modified glass ionomers. *Dental Materials*, 11, 196-9.

TAGGART S. E. & PEARSON G. E. 1991. The effect of etching on glass polyalkenoate cements. *Journal of Oral Rehabilitation*, 18, 31-42.

TAHA, N.A., PALAMARA, J.E. & MESSER, H.H., 2012. Assessment of laminate technique using glass-ionomerand resin composite for restoration of root filled teeth. *Journal of Dentistry*, 40, 617-23.

TAKAHASHI, K., EMILSON, C.G. & BIRKHED, D., 1993. Fluoride release in vitro from various glass ionomer cements and resin composites after exposure to NaF solutions. *Dental Materials*, 9, 350-4.

TAKAMIZAWA, T., BARKMEIER, W.W., SAI, K., TSUJIMOTO, A., IMAI, A., ERICKSON, R.L., LATTA, M.A., MIYAZAK, M. 2018. Influence of different smear layers on bond durability of self-etch adhesives. *Dental Materials,* 34, 246-59.

TAM, L.E., MCCOMB, D. & PULVER, F., 1991. Physical properties of proprietary light-cured lining materials. *Operative Dentistry*, 16, 210-7.

TAM, L.E. & PILLIAR, R.M., 1994. Fracture surface characterization of dentin-bonded interfacial fracture toughness specimens. *Journal of dental research*, 73, 607-19.

TANIGUCHI, G., NAKAJIMA, M., HOSAKA, K., IWAMOTO, N., IKEDA, M., FOXTON, R.M. & TAGAMI, J., 2009. Improving the effect of NaOCl pretreatment on bonding to caries-affected dentin using self-etch adhesives. *Journal of Dentistry*, 37, 769-75.

TANUMIHARJA, M., BURROW, M.F., TYAS, M.J. 2000. Microtensile bond strengths of glass ionomer (polyalkenoate) cements to dentine using four conditioners. *Journal of Dentistry*, 28, 361-6.

TAO, L. & PASHLEY, D.H., 1988. Shear bond strengths to dentin: effects of surface treatments, depth and position. *Dental Materials*, 4, 371-8.

TAY, F.R., KING, N.M., SUH, B.I. & PASHLEY, D.H., 2001a. Effect of delayed activation of light-cured resin composites on bonding of all-in-one adhesives. *Journal of Adhesive Dentistry*, 3, 207-25.

TAY, F.R., SMALES, R.J., NGO, H., WEI, S.H., PASHLEY, D.H. 2001b. Effect of different conditioning protocols on adhesion of a GIC to dentin*. Journal of Adhesive Dentistry*, 3, 153-67.

TAY, F.R., SIDHU, S.K., WATSON, T.F. & PASHLEY, D.H., 2004. Water-dependent interfacial transition zone in resin-modified glass-ionomer cement/dentin interfaces. *Journal of Dental Research*, 83, 644-9.

TEIXEIRA, E.C., BAYNE, S.C., THOMPSON, J.Y., RITTER, A.V. & SWIFT, E.J., 2005. The shear bond strength of self-etching bonding systems in combination with various composites used for repairing aged composites. *Journal of Adhesive Dentistry*, 7, 159-64.

TEN CATE, J.M. & DUIJSTERS, P.P. 1982. Alternating demineralization and remineralization of artificial enamel lesions. *Caries Research,* 16**,** 201-10.

TEN CATE, J.M., BUUS, M.J. & DAMEN, J.J.M., 1995. The effects of GIC restorations on enamel and dentin demineralization and remineralization. *Advances in Dental Research*, 9, 384-8.

TEN CATE, J.M., 1996. Physicochemical aspects of fluoride-enamel interactions. *Fluoride in dentistry*. In: FEJERSKOV, O., EKSTRAND, J., BURT, B.A., editors. Fluoride in dentistry. Copenhagen: Munksgaard, 252-72.

TEZVERGIL-MUTLUAY, A., LASSILA, L.V. & VALLITTU, P.K., 2007. Degree of conversion of dual-cure luting resins light-polymerized through various materials. *Acta Odontologica Scandinavica*, 65, 201-5.

THANJAL, N.K., BILLINGTON, R.W., SHAHID, S., LUO, J., HILL, R.G. & PEARSON, G.J., 2010. Kinetics of fluoride ion release from dental restorative glass ionomer cements: the influence of ultrasound, radiant heat and glass composition. *Journal of Materials Science: Materials in Medicine, 21,* 589-95.

THUY, T.T., NAKAGAKI, H., KATO, K., HUNG, P.A., INUKAI, J., TSUBOI, S., NAKAGAKI, H., HIROSE, M.N., IGARASHI, S. & ROBINSON, C., 2008. Effect of strontium in combination with fluoride on enamel remineralisation in vitro. *Archives of Oral Biology*, 53, 1017-22.

THYLSTRUP, A., BRUUN, C. & HOLMEN, L., 1994. In vivo caries models-mechanisms for caries initiation and arrestment. *Advances in Dental Research*, 8, 144-57.

TOLEDANO, M., AGUILERA, F.S., OSORIO, E., CABELLO, I., TOLEDANO-OSORIO, M. & OSORIO, R. 2016. Efficacy and micro-characterization of pathophysiological events on caries-affected dentin treated with glass-ionomer cements. *International Journal of Adhesion & Adhesives*, 69, 91-109.

TOLIDIS, K., NOBECOURT, A. & RANDALL, R.C. 1998. Effect of a resin modified liner on volumetric polymerization shrinkage of various composites. *Dental Materials,* 14, 417-23.

TOSAKI, S., 1994. Current and future trends for light cured systems. In *Glass ionomers. The next generation Proceedings of the 2nd international symposium on glass ionomers* (pp. 35-46).

TRAJTENBERG, C.P. & POWERS, J.M. 2004. Bond strengths of repaired laboratory composites using three surface treatments and three primers. *American Journal of Dentistry*, 17,123-6.

TSUNEKAWA, M., SETCOS, J.C., USAMI, Y., IWAKU, M. & MARSHALL, S.J., 1992. A new light-activated adhesive composite: shear bond strength and microleakage. *Dental Materials*, 8, 234-7.

TURKISTANI, A., NAKASHIMA, S., SHIMADA, Y., TAGAMI, J. & SADR, A., 2015. Microgaps and demineralization progress around composite restorations. *Journal of Dental Research*, 94, 1070-7.

TYAS M J. 1991a. Cariostatic effect of glass ionomer cement: a 5-year clinical study. *Australian Dental Journal,* 36, 236-9.

TYAS, M.J. 1991b. Dental materials science - the maintenance of standards*.* *Journal of oral rehabilitation,* 18, 105-10.

TYAS, M.J. 1992. Dental materials science the maintenance of standards. *Oral Rehabilitation,* 18, 105-10.

TYAS, M.J., ANUSAVICE, K.J., FRENCKEN, J.E. & MOUNT, G.J., 2000. Minimal intervention dentistry—a review* FDI Commission Project 1–97. *International Dental Journal*, 50, 1-12.

TYAS, M.J., 2003. Milestones in adhesion: glass-ionomer cements. *Journal of Adhesive Dentistry*, 5, 259-266.

TYAS, M.J. & BURROW, M.F., 2004. Adhesive restorative materials: a review. *Australian Dental Journal*, 49, 112-21.

ULUKAPI, H., BENDERLI, Y. & SOYMAN, M., 1996. Determination of fluoride release from light-cured glass-ionomers and a fluoridated composite resin from the viewpoint of curing time. *Journal of oral rehabilitation*, 23, 197-201.

UNEMORI, M., MATSUYA, Y., MATSUYA, S., AKASHI, A. & AKAMINE, A., 2003. Water absorption of poly (methyl methacrylate) containing 4-methacryloxyethyl trimellitic anhydride. *Biomaterials*, 24, 1381-7.

UNO, S., FINGER, W.J. & FRITZ, U., 1996. Long-term mechanical characteristics of resin-modified glass ionomer restorative materials. *Dental Materials*, 12, 64-9.

VAN DIJKEN, F.W. 1996. 3-year clinical evaluation of a compomer, a resin-modified glass ionomer and a resin composite in Class III cavities. *American journal of Dentistry,* 9,195-8.

VAN DIJKEN, J.W. & SJOSTROM, S. 1995. Gingival reactions around and plaque formation on resin composites and glass-ionomer cements. *Advances in Dental Research*, 9, 363-6.

VAN DIJKEN, J.W, KIERI, C. & CARLEN, M., 1999. Longevity of extensive class II open-sandwich restorations with a resin-modified glass-ionomer cement. *Journal of Dental Research*, 78, 1319-25.

VAN DIJKEN, J.W. & PALLESEN, U., 2017. Bulk-filled posterior resin restorations based on stress-decreasing resin technology: a randomized, controlled 6-year evaluation. *European Journal of Oral Sciences*, 125, 303-9.

VAN DUINEN, R.N., DAVIDSON, C.L., DE GEE, A.J. & FEILZER, A.J. 2004. In situ transformation of glass-ionomer into an enamel-like material. *American Journal of Dentistry*, 17, 223-7.

VAN ENDE, A., DE MUNCK, J., MINE, A., LAMBRECHTS, P. & VAN MEERBEEK, B., 2010. Does a low-shrinking composite induce less stress at the adhesive interface?. *Dental Materials*, 26, 215- 22.

VAN GEMERT-SCHRIKS, M.C.M., VAN AMERONGEN, W.E., TEN CATE, J.M. & AARTMAN, I.H.A., 2007. Three-year survival of single-and two-surface ART restorations in a high-caries child population. *Clinical Oral Investigations*, 11, 337-43.

VAN LANDUYT, K.L., SNAUWAERT, J., DE MUNCK, J., PEUMANS, M., YOSHIDA, Y., POITEVIN, A., COUTINHO, E., SUZUKI, K., LAMBRECHTS, P. & VAN MEERBEEK, B., 2007. Systematic review of the chemical composition of contemporary dental adhesives. *Biomaterials*, 28, 3757-85.

VAN LANDUYT, K.L., YOSHIDA, Y., HIRATA, I., SNAUWAERT, J., DE MUNCK, J., OKAZAKI, M., SUZUKI, K., LAMBRECHTS, P. & VAN MEERBEEK, B., 2008. Influence of the chemical structure of functional monomers on their adhesive performance. *Journal of Dental Research*, 87,757-61.

VAN MEERBEEK, B., YOUSHIDA, Y., LAMBRECHTS, P., VANHERLE, G., WAAKASA, K. & NAKAYAMA, Y. 1998. Mechanisms of bonding a resin-modified glass-ionomer adhesive to dentin. *Journal of Dental Research,* 77,911 [abstract 2236].

VAN MEERBEEK, B., DE MUNCK, J., YOSHIDA, Y., INOUE, S., VARGAS, M., VIJAY, P., VAN LANDUYT, K., LAMBRECHTS, P. & VANHERLE, G., 2003. Adhesion to enamel and dentin: current status and future challenges. *Operative Dentistry-University of Washington*, 28, 215-35.

VAN MEERBEEK, B., YOSHIDA, Y., INOUE, S., DE MUNCK J, VAN LANDUYT, K. & LAMBRECHTS, P. 2006. Glass-ionomer adhesion: the mechanisms at the interface. *Journal of Dentistry*, 34, 615-17.

VAN MEERBEEK, B., PEUMANS, M., POITEVIN, A., MINE, A., VAN ENDE, A., NEVES, A. & DE MUNCK, J. 2010. Relationship between bond-strength tests and clinical outcomes. *Dental Materials*, 26, 100-21.

VAN MEERBEEK, B., YOSHIHARA, K., YOSHIDA, Y., MINE, A.J., DE MUNCK, J. & VAN LANDUYT, K.L., 2011. State of the art of self-etch adhesives. *Dental Materials*, 27, 17-28.

VAN NOORT, R., NOROOZI, S., HOWARD, I.C. & CARDEW, G., 1989. A critique of bond strength measurements. *Journal of Dentistry*, 17, 61-7.

VAN NOORT, R., CARDEW, G.E., HOWARD, I.C. & NOROOZI, S., 1991. The effect of local interfacial geometry on the measurement of the tensile bond strength to dentin. *Journal of Dental Research*, 70, 889-93.

VAN NOORT, R. & DAVIS, L.G. 1993. A prospective study of the survival of chemically activated anterior resin composite restorations in general dental practice: 5-year results. *Journal of Dentistry,* 21: 209-15.

VAN NOORT, R., 2014. *Introduction to Dental Materials-E-Book*. 4th edition, Elsevier Health Sciences.

VERBEECK, R.M., DE MOOR, R.J., VAN EVEN, D.F. & MARTENS, L.C. 1993. The short-term fluoride release of a hand-mixed versus capsulated system of a restorative glass-ionomer cement. *Journal of Dental Research,* 72,577-81.

VERBEECK, R.M., DE MAEYER, E.A., MARKS, L.A., DE MOOR, R.J.G., DE WITTE, A.M. & TRIMPENEERS, L.M.1998. Fluoride release process of (resin-modified) glass-ionomer cements versus (polyacid-modified) composite resins. *Biomaterials*, 19, 509-19.

VERMEERSCH, G., LELOUP, G., DELMEE, M. & VREVEN, J., 2005. Antibacterial activity of glass–ionomer cements, compomers and resin composites: relationship between acidity and material setting phase. *Journal of Oral Rehabilitation*, 32, 368-74.

VERSLUIS, A., TANTBIROJN, D., LEE, M.S., TU, L.S. & DELONG, R., 2011. Can hygroscopic expansion compensate polymerization shrinkage? Part I. Deformation of restored teeth. *Dental Materials,* 27, 126-33.

VROCHARI, A.D., ELIADES, G., HELLWIG, E. & WRBAS, K.T., 2009. Curing efficiency of four self-etching, self-adhesive resin cements. *Dental materials*, 25, 1104-8.

WADA, T., 1986. Development of a new adhesive material and its properties. In *Proceedings of the International Symposium on Adhesive Prosthodontics, Amsterdam, Netherlands.* Academy of Dental Materials.

WAGNER, A., WENDLER, M., PETSCHELT, A., BELLI, R. & LOHBAUER, U. 2014. Bonding performance of universal adhesives in different etching modes. *Journal of Dentistry,* 42, 800-7.

WALTER, R., MIGUEZ, P.A. & PEREIRA, P.N., 2005. Microtensile bond strength of luting materials to coronal and root dentin. *Journal of Esthetic and Restorative Dentistry*, 17, 165-171.

WAN, A.C., YAP, A.U. & HASTINGS, G.W., 1999. Acid-base complex reactions in resin-modified and conventional glass ionomer cements. *Journal of Biomedical Materials Research*, 48, 700-4.

WARREN, J.A. & SÖDERHOLM, K.J., 1988. Bonding amalgam to glass-ionomer with PAA. *Dental Materials*, 4,191-6.

WATTS, D.C. 1986. The development of surface hardness in visible light cured posterior composites. *Journal of Dentistry*, 14,169-174.

WATSON, T.F., BILLINGTON, R.W. & WILLIAMS, J.A., 1991. The interfacial region of the tooth/glass ionomer restoration: a confocal optical microscope study. *American Journal of Dentistry*, 4, 303-10.

WATSON, T.F., BANERJEE, A. 1993. Effectiveness of glass-ionomer surface protection treatments: a scanning optical microscope study. *The European Journal of Prosthodontics and Restorative Dentistry*, 2, 85-90.

WATSON, T.F., 1994. Ionomers vs. composites at the tooth interface. In *Proceedings of the Second International Symposium on Glass Ionomer Cements, 1994*. International Symposia in Dentistry.

WATSON, T.F., PAGLIARI, D., SIDHU, S.K. & NAASAN, M.A. 1998. Confocal microscopic observation of structural changes in glass-ionomer cements and tooth interfaces. *Biomaterials*, 19, 581-8.

WATSON, T. 1999. Bonding of glass-ionomer cements to tooth structure. In: Davidson CL, Mjör IA, eds. Advances in glass-ionomer cements. Chicago: Quintessence, 121-36.

WATSON, T.F., ATMEH, A.R., SAJINI, S., COOK, R.J. & FESTY, F., 2014. Present and future of glass-ionomers and calcium-silicate cements as bioactive materials in dentistry: biophotonics-based interfacial analyses in health and disease. *Dental Materials*, 30, 50-61.

WASSON, E.A., NICHOLSON, J.W.1991. Studies on the setting chemistry of glass ionomer cements. *Clinical Materials*, 7:289-93.

WASSON, E.A. & NICHOLSON, J.W., 1993. New aspects of the setting of glass-ionomer cements. *Journal of Dental Research*, 72, 481-3.

WEI, Y.J., SILIKAS, N., ZHANG, Z.T. & WATTS, D.C. 2011. Hygroscopic dimensional changes of self-adhering and new resin–matrix composites during water sorption/desorption cycles. *Dental Materials*, 27, 259-66.

WELCH, D., SEESENGOOD, B. & HOPP, C., 2015. Surface treatments that demonstrate a significant positive effect on the shear bond strength of repaired resin-modified glass ionomer. *Operative Dentistry*, 40, 403-9.

WELSH, E.L. & HEMBREE, J.H., 1985. Microleakage at the gingival wall with four class V anterior restorative materials. *Journal of Prosthetic Dentistry*, 54, 370-2.

WHITE, D.J., CHEN, W.C. & NANCOLLAS, G.H. 1988. Kinetic and physical aspects of enamel remineralization- a constant composition study. *Caries Research,* 22**,** 11-9.

WILLIAMS, J.A. & BILLINGTON, R.W. 1991. Changes in compressive strength of glass ionomer restorative materials with respect to time periods of 24h to 4 months. *Journal of Oral Rehabilitation,* 18,163-8.

WILLIAMS, J.A., BILLINGTON, R.W. & PEARSON, G.J. 1999. The influence of sample dimensions on fluoride ion release from a glass ionomer restorative cement. *Biomaterials,* 20, 1327-37.

WILLIAMS, J. A., BILLINGTON, R. W. & PEARSON, G. J. 2002. The glass ionomer cement: the sources of soluble fluoride. *Biomaterials,* 23, 2191-200.

WILLIAM, V., BURROW, M.F., PALAMARA, J.E., MESSER, L.B. 2006. Microshear bond strength of resin composite to teeth affected by molar hypomineralization using 2 adhesive systems. *Pediatric Dentistry,* 28,233-41.

WILSON, A. D. & KENT, B. E. 1972. A new translucent cement for dentistry. The glass ionomer cement. *British Dental Journal,* 132**,** 133-5.

WILSON, A.D., 1976. Specification test for the solubility and disintegration of dental cements: a critical evaluation of its meaning. *Journal of Dental Research*, 55, 721-9.

WILSON, A.D., CRISP, S. & FERNER, A.J. 1976. Reactions in glass-ionomer cements IV. Effect of chelating comonomers on setting behaviour. *Journal of Dental Research,* 55, 489-95.

WILSON, A. 1978. The chemistry of dental cements. *Chemical Society Reviews*, 7, 265-96.

WILSON, A. D., PADDON, J. M. & CRISP, S. 1979. The hydration of dental cements. *Journal of Dental Research,* 58, 1065-71.

WILSON, A. D. & PROSSER, H. J. 1982. Biocompatibility of the glass ionomer cement. *The Journal of the Dental Association of South Africa,* 37, 872-9.

WILSON, A.D. & MCLEAN, J.W. 1988. Composition. In: Wilson AD, McLean JW, editors. *Glass-ionomer cement*. Chapter 2. Chicago, IL: Quintessence Publishing Co., Inc.

WILSON, A.D. 1989. Developments in glass-ionomer cements. *International Journal of Prosthodontics*, *2*, 438-46.

WILSON, A.D., 1990. Resin-modified glass-ionomer cements. *International Journal of Prosthodontics*, 3, 425-9.

WILSON, N.H.F. 1990. The evaluation of materials: relationship between laboratory investigations and clinical studies. *Operative Dentistry,* 15, 149-55.

WILSON, A.D., & NICHOLSON, J.W. 1993. *Polyalkenoate cements*. In: WILSON, A.D. & NICHOLSON, J.W., editors. *Acid–base cements: their biomedical and industrial applications*. Chapter 5. Cambridge University Press, p. 90–196.

WILSON, A.D. 1994. Resin-modified glass ionomer cements. *International Journal of Prosthodontics*, 19, 82.

WILSON, A.D. & NICHOLSON, J.W., 2005. *Acid-base cements: their biomedical and industrial applications* (Vol. 3). Cambridge University Press.

WILSON, N.H.F., LYNCH, C.D., BRUNTON, P.A., HICKEL, R., MEYER-LUECKEL, H., GURGAN, S., PALLESEN, U., SHEARER, A.C., TARLE, Z., COTTI, E. & VANHERLE, G., 2016. Criteria for the replacement of restorations: Academy of Operative Dentistry European Section. *Operative Dentistry*, 41, S48-S57.

WOLFART, M., WOLFART, S. & KERN, M., 2006. Retention forces and seating discrepancies of implant-retained castings after cementation. *International Journal of Oral & Maxillofacial Implants*, 21, 519-25.

WOOD, D. & HILL, R. 1991. Structure–property relationship in ionomer glasses. *Clinical Materials,* 7, 301-12.

WOOLFORD, M.J. & GRIEVE, A.R. 1993. The use of intermediary resins when bonding glass polyalkenoate (ionomer) cement to composite resin. *Journal of Oral Rehabilitation*, 20, 249-55.

WOOLFORD, M., 1993. Composite resin attached to glass polyalkenoate (ionomer) cement-the laminate technique. *Journal of dentistry*, 21, 31-8.

WU, C.H. & SMALES, R.J., 2001. Occlusal wear of a resin-modified glass-ionomercement following three surface treatments. *Clinical oral investigations*, 5, 26-30.

XIE, D., CULBERTSON, B.M. & JOHNSTON, W.M., 1998. Formulation of visible light-curable glass-ionomer cements containing N-vinylpyrrolidone. *Journal of Macromolecular Science-Pure and Applied Chemistry*, 35, 1631-50.

XIE, D., BRANTLEY, W.A., CULBERTSON, B.M. & WANG, G., 2000. Mechanical properties and microstructures of glass-ionomer cements. *Dental Materials*, 16, 129-38.

XIE, D., WU, W., PUCKETT, A., FARMER, B. & MAYS, J.W., 2004. Novel resin modified glass-ionomer cements with improved flexural strength and ease of handling. *European polymer journal*, 40, 343-51.

XIE, Z., KILPATRICK, N.M., SWAIN, M.V., MUNROE, P.R. & HOFFMAN, M. 2008. Transmission electron microscope characterisation of molar-incisor-hypomineralisation. *Journal of Material Science Material in Medicine,* 19, 3187-92

XU, X. & BURGESS, J.O. 2003. Compressive strength, fluoride release and recharge of fluoride-releasing materials. *Biomaterials*, 24, 2451-61.

YAMAMOTO, Y. 1984. The study on hydroxyapatite-polyacrylic acid composite cement (hydroxyapatite-glassionomer cement). *Journal of the Japanese Society for Dental Materials and Devices, 3*,787-96.

YAMAZAKI, T., BRANTLEY, W., CULBERTSON, B., SEGHI, R. & SCHRICKER, S., 2005. The measure of wear in N-vinyl pyrrolidinone (NVP) modifed glass-ionomer cements. *Polymers for Advanced Technologies*, 16,113-16.

YAMAZAKI, T., SCHRICKER, S.R., BRANTLEY, W.A., CULBERTSON, B.M. & JOHNSTON, W., 2006. Viscoelastic behavior and fracture toughness of six glass-ionomer cements. *The Journal of Prosthetic Dentistry*, 96, 266-72.

YAMAUCHI, J.K., BUNKYO-WARD, E.M., MATSUDO, N.N., KURASHIKI, K.S., TAKATSUKI, T.W.1981. Dental filling material. Patent.

YANG, B., LUDWIG, K., ADELUNG, R. & KERN, M., 2006. Micro-tensile bond strength of three luting resins to human regional dentin. *Dental Materials*, 22, 45-56.

YAP, A.U. & NEO, J.C. 1995. Non-carious cervical tooth loss. Part 2: Management. *Dental Update*, 22, 364.

YAP, A.U. 1996. Resin-modified glass-ionomer cements: a comparison of water sorption characteristics. *Biomaterials,* 17, 1987.

YAP, A. & LEE, C.M., 1997. Water sorption and solubility of resin-modified polyalkenoate cements. *Journal of Oral Rehabilitation*, 24, 310-314.

YAP, A.U., QUEK, C.E. & KAU C.H. 1998. Repair of new-generation tooth-coloured restoratives: methods of surface conditioning to achieve bonding. *Operative Dentistry,* 23,173-8.

YAP, A.U.J., LYE, K.W. & SAU, C.W., 2000. Effects of aging on repair of resin-modified glass–ionomer cements. *Journal of Oral Rehabilitation*, 27,422-7.

YAP, A.U., MUDAMBI, S., CHEW, C.L. & NEO, J.C. 2001. Mechanical properties of improved visible light-cured resin-modified glass ionomer cement. *Operative Dentistry*, 26, 295-301.

YENGOPAL, V. & MICKENAUTSCH, S., 2011. Caries-preventive effect of resin-modified glass-ionomer cement (RM-GIC) versus composite resin: a quantitative systematic review. *European Archives of Paediatric Dentistry*, 12, 5-14.

YESILYURT, C., KUSGOZ, A., BAYRAM, M. & ULKER, M., 2009. Initial Repair Bond Strength of a Nano-filled Hybrid Resin: Effect of Surface Treatments and Bonding Agents. *Journal of Aesthetic and Restorative Dentistry*, 21, 251-60.

YIU, C.K., TAY, F.R., KING, N.M., PASHLEY, D.H., SIDHU, S.K., NEO, J.C., TOLEDANO, M. & WONG, S. L. 2004a. Interaction of glass-ionomer cements with moist dentin. *Journal of Dental Research,* 83, 283-89.

YIU, C.K.Y., TAY, F.R., KING, N.M., PASHLEY, D.H., CARVALHO, R.M. & CARRILHO, M.R.O. 2004b. Interaction of resin-modified glass-ionomer cements with moist dentine. *Journal of Dentistry,* 32, 521-30.

YIP, H.K., SMALES, R.J., NGO, H.C., TAY, F.R. & CHU, F.C., 2001. Selection of restorative materials for the atraumatic restorative treatment (ART) approach: a review. *Special Care in Dentistry*, 21, 216-21.

YIP, H.K., TAY, F.R., NGO, H.C., SMALES R.J. & PASHLEY D.H. 2001. Bonding of contemporary glass ionomer cements to dentin*. Dental Materials*, 17, 456-70

YLI-URPO, H., LASSILA, L.V., NÄRHI, T. & VALLITTU, P.K. 2005a. Compressive strength and surface characterization of glass ionomer cements modified by particles of bioactive glass. *Dental Materials*, 21, 201-9.

YLI-URPO, H., NÄRHI, M. & NÄRHI, T. 2005b. Compound changes and tooth mineralization effects of glass ionomer cements containing bioactive glass (S53P4), an in vivo study. *Biomaterials*, 26, 5934-41.

YOSHIDA, K., KAMADA, K. & ATSUTA, M., 1999. Adhesive primers for bonding cobalt–chromium alloy to resin. *Journal of Oral Rehabilitation*, 26, 475-8.

YOSHIDA, Y., VAN MEERBEEK, B., NAKAYAMA, Y., SNAUWAERT, J., HELLEMANS, L., LAMBRECHTS, P., VANHERLE, G. & WAKASA, K., 2000. Evidence of chemical bonding at biomaterial-hard tissue interfaces. *Journal of Dental Research*, 79, 709-14.

YOSHIDA, Y., VAN MEERBEEK, B., NAKAYAMA, Y., YOSHIOKA, M., SNAUWAERT, J., ABE, Y., LAMBRECHTS, P., VANHERLE, G. & OKAZAKI, M., 2001. Adhesion to and decalcification of hydroxyapatite by carboxylic acids. *Journal of Dental Research*, 80, 1565-9.

YOSHIOKA, M., YOSHIDA, Y., INOUE, S., LAMBRECHTS, P., VANHERLE, G., NOMURA, Y., OKAZAKI, M., SHINTANI, H. & VAN MEERBEEK, B. 2002.

Adhesion/decalcification mechanisms of acid interactions with human hard tissues. *Journal of Biomedical Materials Research*, 59, 56-62.

YOSHIDA, Y., NAGAKANE, K., FUKUDA, R., NAKAYAMA, Y., OKAZAKI, M., SHINTANI, H., INOUE, S., TAGAWA, Y., SUZUKI, K., DE MUNCK, J. & VAN MEERBEEK, B., 2004. Comparative study on adhesive performance of functional monomers. *Journal of Dental Research*, 83, 454-8.

YOSHIDA, Y., YOSHIHARA, K., NAGAOKA, N., HAYAKAWA, S., TORII, Y., OGAWA, T., OSAKA, A. & MEERBEEK, B.V., 2012. Self-assembled nano-layering at the adhesive interface. *Journal of Dental Research*, 91, 376-81.

YOSHIKAWA, T., HIRASAWA, M, TOSAKI, S., HIROTA, K. 1994. Concentration of HEMA eluted from light-cured glass ionomers. *Journal of Dental Research*, 73,133.

YOSHIOKA, M., YOSHIDA, Y., INOUE, S., LAMBRECHTS, P., VANHERLE, G., NOMURA, Y., OKAZAKI, M., SHINTANI, H. & VAN MEERBEEK, B., 2002. Adhesion/decalcification mechanisms of acid interactions with human hard tissues. *Journal of Biomedical Materials Research: An Official Journal of the Society for Biomaterials and the Japanese Society for Biomaterials*, 59, 56-62.

YOSHIYAMA, M., URAYAMA, A., KIMOCHI, T., MATSUO, T. & PASHLEY, D.H. 2000. Comparison of conventional vs self-etching adhesive bonds to caries-affected dentin. *Operative Dentistry,* 25,163-9.

YOSHIYAMA, M., TAY, F.R., DOI, J., NISHITANI, Y., YAMADA, T., ITOU, K. CARVALHO, R.M., NAKAJIMA, M. & PASHLEY, D.H. 2002. Bonding of self-etch and total-etch adhesives to carious dentin. *Journal of Dental Research,* 81, 556-60.

YOSHIYAMA, M., TAY, F.R., TORII, Y., NISHITANI, Y., DOI, J., ITOU, K., CIUCCHI, B. & PASHLEY, D.H. 2003. Resin adhesion to carious dentin. *American Journal of Dentistry*, 16, 47-52.

YOUNG, A.M. 2002. FTIR investigation of polymerization and polyacid neutralization kinetics in resin-modified glass-ionomer dental cements. *Biomaterials,* 23, 3289-95.

YOUNG, A.M., RAFEEK, S.A. & HOWLETT, J.A. 2004. FTIR investigation of monomer polymerisation and polyacid neutralisation kinetics and mechanisms in various aesthetic dental restorative materials. *Biomaterials,* 25, 823-33.

ZACHRISSON, B.U., BÜYÜKYILMAZ, T. & ZACHRISSON, Y., 1995. Improving orthodontic bonding to silver amalgam. *The Angle Orthodontist*, 65, 35-42.

ZAINUDDIN, N., KARPUKHINA, N., LAW, R.V. & HILL, R.G. 2012. Characterisation of a remineralising Glass Carbomer® ionomer cement by MAS-NMR spectroscopy. *Dental Materials*, 28, 1051-8.

ZANATA, R.L., FAGUNDES, T.C., DE ALMENDRA FREITAS, M.C.C., LAURIS, J.R.P. & DE LIMA NAVARRO, M.F., 2011. Ten-year survival of ART restorations in permanent posterior teeth. *Clinical oral investigations*, 15, 265-71.

ZANKULI, M.A., DEVLIN, H. & SILIKAS, N., 2014. Water sorption and solubility of core build-up materials. *Dental Materials*, 30, 324-9.

ZHANG, Y., BURROW, M.F., PALAMARA, J.E.A. & THOMAS, C.D.L., 2011. Bonding to glass-ionomer cements using resin-based adhesives. *Operative Dentistry,* 36, 618-25.

ZHAO, J. & XIE, D., 2011. A novel hyperbranched poly (acrylic acid) for improved resin-modified glass-ionomer restoratives. *Dental Materials*, 27, 478-486.

ZOERGIEBEL, J. & ILIE, N., 2013. Evaluation of a conventional glass-ionomer cement with new zinc formulation: effect of coating, aging and storage agents. *Clinical Oral Investigations*, 17, 619-26.

7 Appendices

7.1 Ethical approval for human teeth collection

NHS
Health Research Authority

Prof Tim Watson
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19 January 2017

Dear Prof Watson

Letter of HRA Approval

Study title: **Improving Caries and Toothwear Management Techniques**
IRAS project ID: **157705**
Protocol number: **N/A**
REC reference: **16/SW/0220**
Sponsor **King's College London**

I am pleased to confirm that HRA Approval has been given for the above referenced study, on the basis described in the application form, protocol, supporting documentation and any clarifications noted in this letter.

Participation of NHS Organisations in England

The sponsor should now provide a copy of this letter to all participating NHS organisations in England.

Appendix B provides important information for sponsors and participating NHS organisations in England for arranging and confirming capacity and capability. **Please read *Appendix B* carefully**, in particular the following sections:

- *Participating NHS organisations in England* – this clarifies the types of participating organisations in the study and whether or not all organisations will be undertaking the same activities
- *Confirmation of capacity and capability* - this confirms whether or not each type of participating NHS organisation in England is expected to give formal confirmation of capacity and capability. Where formal confirmation is not expected, the section also provides details on the time limit given to participating organisations to opt out of the study, or request additional time, before their participation is assumed.
- *Allocation of responsibilities and rights are agreed and documented (4.1 of HRA assessment criteria)* - this provides detail on the form of agreement to be used in the study to confirm capacity and capability, where applicable.

Further information on funding, HR processes, and compliance with HRA criteria and standards is also provided.

7.2 List of oral presentations in conferences and a publication in international peer-reviewed journal

Lamis Al-Taee, Sanjukta Deb, Avijit Banerjee, “An in-vitro assessment of the physical properties of hand-mixed & encapsulated glass ionomer cements” (oral Presentation), 18th Annual Postgraduate Research Day at King’s College London Dental Institute (2016).

Lamis Al-Taee, Sanjukta Deb, Avijit Banerjee, “New Reparative Material for Repairing Failed Restorations” (oral Presentation), 19th Annual Postgraduate Research Day at King’s College London Dental Institute (2017).

Lamis Al-Taee, Avijit Banerjee, Sanjukta Deb “Novel Resin-modified Glass-Ionomer Cement for Repairing Defective Tooth-Restoration complexes” (oral presentation). British society of Dental Research (BSODR), Plymouth, UK, 6th- 8th September (2017).

Lamis Al-Taee, Avijit Banerjee, Sanjukta Deb “Adhesion Strength of a New Reparative Material to Sound versus Defective Enamel/Dentine” (oral presentation). International Association of Dental Research (IADR), London, UK, 25th – 28th July (2018).

Lamis Al-Taee, Avijit Banerjee, Sanjukta Deb. 2019. An Integrated Multifunctional Hybrid Cement (pRMGIC) for Dental Applications. *Dental Materials,* 35,636-49.

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An integrated multifunctional hybrid cement (pRMGIC) for dental applications

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ABSTRACT

Objective. Glass-ionomer and resin-modified glass-ionomer cements are versatile materials with the ability to form a direct bond with tooth tissues. The aim of this study was to formulate a novel class of dental bio-interactive restorative material (pRMGIC) based on resin-modified glass-ionomer cements via the inclusion of an organophosphorus monomer, ethylene glycol methacrylate phosphate, with a potential to improve the mechanical properties and also function as a reparative restorative material.

Methods. pRMGIC was formulated with modification of the resin phase by forming mixes of ethylene glycol methacrylate phosphate (EGMP; 0–40%wt) and 2-hydroxyethyl methacrylate monomer into the liquid phase of a RMGIC (Fuji II LC, GC Corp.). The physical properties of the cements were determined including setting characteristics, compressive strength and modulus (CS &CM), microhardness (MH) and biaxial flexural strength (BFS). Fluid uptake and fluoride release were assessed up to 60 days storage. Adhesion to sound dentine was measured using micro-tensile bond strength and surface integrity was analysed using SEM coupled with EDX. Statistical analysis was performed using ANOVA and Bonferroni post-hoc tests.

Results. The pRMGIC cements exhibited an increase in working time with increasing EGMP concentration however were within the limits of standard clinical requirements. Although the compressive strength of pRMGIC cements were comparable to control cements in the early stages of maturation, the higher EGMP-containing cements (EGMP30 and 40) exhibited significantly greater values ($p < 0.05$) after 4 weeks storage (141.0 ± 9 and 140.4 ± 8 MPa, respectively), in comparison to EGMP0 (128.8 ± 7 MPa). A dramatic two fold increase in biaxial flexural strength ($p < 0.001$) was observed for the pRMGIC's. Furthermore, the ability to decalcify tooth apatite resulted in enhanced interfacial adhesion due to chelation with calcium ions of tooth apatite. The inclusion of EGMP encouraged formation of reinforcing complexes within the RMGIC, thus improving physical properties, decreasing solubility and lower fluoride release. A dense microstructure was observed with increasing EGMP content.

Significance. A novel universal bio-interactive adhesive repair material will enable clinicians to offer more effective repair of the tooth-restoration complex, thus future treatments will benefit both patient and a severely constrained healthcare budget.

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7.3 The of preparation of the artificial saliva

This solution was prepared at ambient temperature, (Eisenburge et al., .2001), 2006).

	Ingredients	**Gram/1Liter**
1	$CaCl_2.2H_2O$	0.1029
2	$MgCl_2$	0.0190
3	KH_2PO_4	0.5440
4	HEPES (acid form)	5.2060
5	KCl	2.2365

7.4 The preparation of Simulated Body Fluid

This solution was prepared at ambient temperature, (Kokubo and Takadama, 2006).

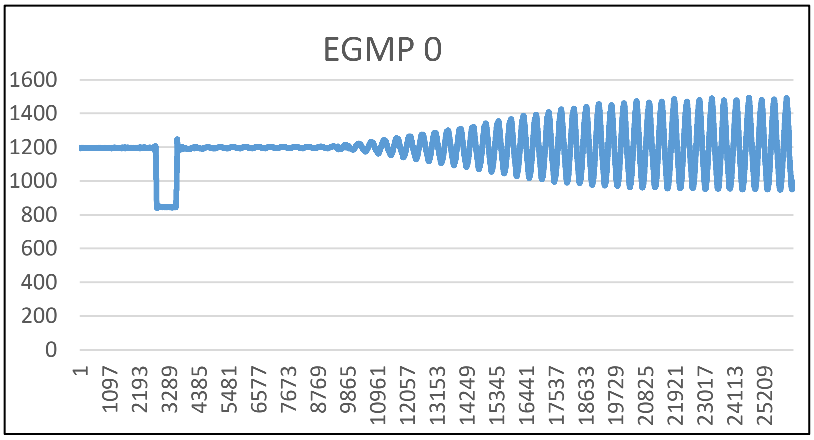
	Reagent	Amount (g)
1	NaCl	7.996
2	$NaHCO_3$	0.350
3	KCl	0.224
4	K_2 HPO_4. $3H_2O$	0.228
5	$MgCl_2.6H_2O$	0.305
6	$CaCl_2$	0.278
7	Na_2SO_4	0.071
8	NH_2C $(CH_2OH)_3$	6.057
9	1 M HCl	40 ml
10	De-ionised water	1000 ml

1. 700 ml of de-ionised water.
2. Add reagents 1,2,3,4,5,7,8 to the de-ionised water. They must be added one by one after each reagent has dissolved completely. When the eight reagent NH_2C (CH2OH) 3, has been added raise the ambient temperature to 37°C.
3. Add 15 ml of the 1 M HCL to the solution.
4. Add the sixth reagent, $CaCl_2$ to the solution.
5. Add the remaining 25 ml of HCL solution and 300 ml of de-ionised water during titration process to achieve pH of 7.4.

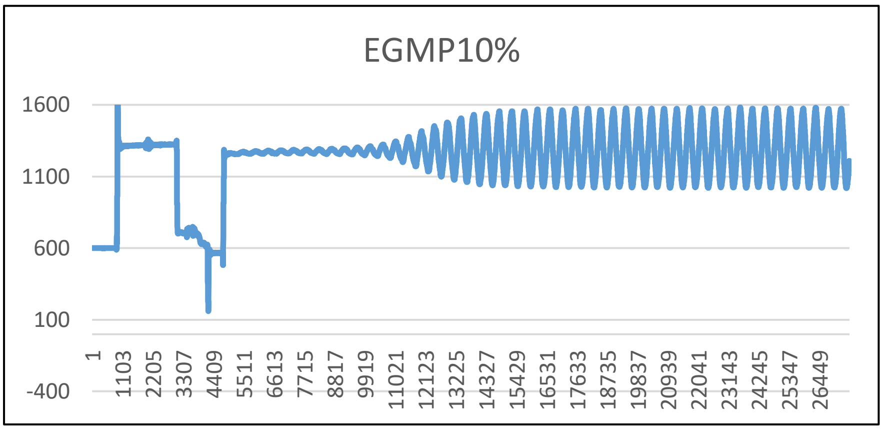
The correct storage temperature and preparation steps are vital for the phase purity and high-temperature stability of the produced HA powders as well as the kinetics of the precipitation process. The prepared SBF solution can be stored at 5°C for a month without degradation.

7.5 Working and setting time of EGMP-RMGICs, measured in seconds and minutes

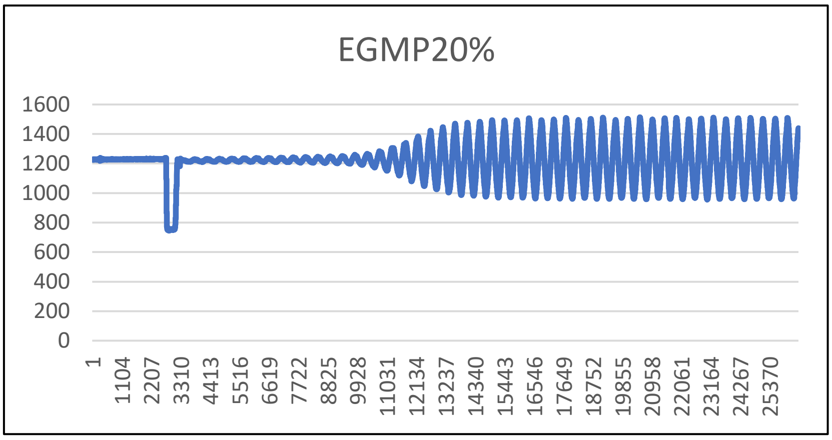
Control (Fuji II LC)				
	Working time		Setting time	
	(sec)	(min)	(sec)	(min)
Control 1	207.11	3.45	321.54	5.36
Control 1	216.1	3.60	326.49	5.44
Control 1	198.45	3.31	312.38	5.20
mean	207.22	3.45	320.14	5.33
SD	8.83	0.15	7.16	0.12



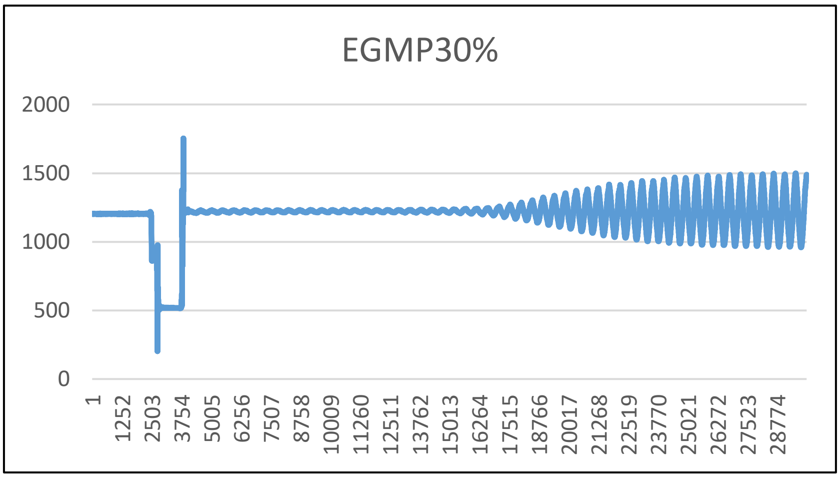
EGMP10				
	Working time		Setting time	
	(sec)	(min)	(sec)	(min)
EGMP10	208.12	3.47	323.10	5.39
EGMP10	210.11	3.50	338.41	5.64
EGMP10	200.34	3.34	310.16	5.17
mean	206.19	3.44	323.89	5.40
SD	5.16	0.1	14.14	0.2



EGMP20				
	Working time		Setting time	
	(sec)	(min)	(sec)	(min)
EGMP20	223.91	3.73	318.04	5.30
EGMP20	195.64	3.26	335.41	5.59
EGMP20	204.20	3.40	324.97	5.40
mean	207.92	3.46	326.14	5.43
SD	14.50	0.24	8.74	0.15



EGMP30				
	Working time		Setting time	
	(sec)	(min)	(sec)	(min)
EGMP30	258.02	4.30	331.62	5.3
EGMP30	246.88	4.1	341.23	5.7
EGMP30	257.96	4.30	350.1	5.83
mean	254.29	4.24	340.98	5.63
SD	6.41	0.1	9.24	0.21



EGMP40				
	Working time		Setting time	
	(sec)	(min)	(sec)	(min)
EGMP40	245.9	4.1	341.54	5.69
EGMP40	234.9	3.9	320.49	5.34
EGMP40	248.0	4.1	340.38	5.67
mean	242.9	4.0	334.1	5.6
SD	7.0	0.1	9.7	0.2