

A COMPARATIVE EVALUATION OF TENSILE STRENGTH AND SHEAR BOND STRENGTH OF TWO NEW EXPERIMENTAL SHOCK-ABSORBING DENTAL CEMENTS

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ABSTRACT

Statement of problem: Dental cements exhibit varying degrees of brittleness, which may concentrate the masticatory forces on the cement, veneer materials or underlying abutment either natural tooth or implant.

Materials and method: A total of 28 upper maxillary molar resin dies were constructed, on which metal copings –with loops- were fabricated. Samples were divided into four groups according to cement used for coping cementation; group A; cemented using GIC cement (Ketac-Cem Apilcap, 3M ESPE, USA), group B; cemented using experimental medical grade silicon adhesive (EMGS) (Dow Corning Corporation, USA), group C; cemented using experimental medical grade rubber adhesive (EMGR) (Dow Corning Corporation, USA) and group D; cemented using Resin cement (Rely X U200, 3M ESPE, USA). Samples were submitted to a tensile test using a universal testing machine (Instron 8874; Instron Corp. Canton, Mass) under a constant crosshead speed of 1.00 mm/min. Another 28 upper maxillary molar were embedded in transparent acrylic resin (Vertex-Dental B.V., The Netherlands). The blocks were trimmed to expose dentine surface on which a 3x3 mm Cr-Co cubes with a thickness of 2 mm were cemented on them. Samples were also divided according to cement used. A mono-bevelled chisel shaped metallic rod attached to the upper movable compartment of testing machine traveling at cross-head speed of 0.5 mm/min was used to apply load on the side metal cubes attached to lower compartment, for shear bond testing. All data was collected and statistically analyzed.

Results: The highest mean value was observed in group D (611.5 MPa) and the lowest mean value was observed for group C (12.86 MPa). Statistical analysis revealed no significant difference between groups (P= 1.03). Statistical analysis among groups showed significant difference between groups A and B (P= 0.003), between groups A and C (P= 0.02) and between groups B and C (P= 0.0001). But no significant differences were found between groups A and D (P= 8.11), groups B and D (P= 4.36) or between groups C and D (P= 3.96). Regarding shear bond strength the highest mean value was observed in group D (13.23 MPa) and the lowest mean value was observed for group C (0.2 MPa). Statistical analysis revealed no significant difference between groups (P= 4.93). Statistical analysis among groups showed significant difference between groups A and C (P= 0.0002), between groups A and D (P= 0.01), between groups B and C (P= 0.01) and between groups C and D (P= 0.0002). But no significant differences were found between groups A and B (P= 1.16), or between groups B and D (P= 9.22).

Conclusions: None of the two experimental medical grade adhesives showed comparable results to either GIC or resin cements.

KEYWORDS: Stress-Breaking, Shock-Absorbing, Glass ionomer, Resin, cement, Rubber adhesive, Silicon adhesive, Tensile strength and Shear bond strength

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INTRODUCTION

An ideal dental cement would have: easy manipulation, low film thickness, long working time with rapid set, low solubility, high compressive and tensile strengths, adhesion to tooth/restoration, anticariogenicity, biocompatibility, and translucency or radiopacity. Physical properties should be taken into consideration along with handling characteristics, technique sensitivity, and results from long term clinical trials ⁽¹⁾.

Glass Ionomer cement (GIC) is a water based cement, and therefore displays an inherent sensitivity to moisture. The release of clinically meaningful amounts of fluoride requires a chemically active (degrading) cement. Some claimed that this is the reason for the swelling of glass ionomer cements after placement, leading to the possible fracture of all-ceramic restorations. However, the advantages of its bacteriostatic effect and adequate compressive and tensile strength make it an acceptable cement, although GIC takes time to develop maximum strength.

Composite resin cements are relatively insoluble and, compared to other dental cements, have the highest mechanical physical properties. ⁽²⁾ They demonstrated adhesion to, tooth structure, etched porcelain and sand-blasted metal. ⁽³⁻⁶⁾ When used with etched ceramic restorations, resin cements make the restoration more resistant to fracture. ⁽⁷⁾ They can be either self-cured, dual-cured or light-cured. Light-cure composite cements are recommended for cementation of veneers. ⁽⁸⁻¹⁰⁾ There has been concern that self-cure and dual-cure composites are chemically incompatible with light-cure ones. ⁽¹¹⁻¹⁵⁾

It was said that, self-adhesive resin cements could be used for all applications except with translucent ceramic restorations, where a possible color change in the cement could affect the color of the restoration. Even though these cements are self-adhesive using the mechanisms

of the self-etch, some cements require a separate primer on the tooth surface before using the cement. ^(4,16,17)

However, all of dental cements exhibit varying degrees of brittleness. This feature may concentrate the masticatory forces on both the cement and veneer materials. Leading to possible cracking and subsequent failure of the cementing medium. This may lead to chipping or failure of the veneer material. Larsson et al ⁽¹⁸⁾ noticed that significantly more porcelain veneer fractures are reported for implant supported zirconia fixed dental prostheses when compared with tooth-supported restorations. One explanation for this could be the role played by the periodontal ligament (PDL), which allows for shock absorption, sensory function, and tooth movement.

Implant dentistry has become an increasingly effective method for correcting edentulism. Implant treatments exhibit an overall excellent clinical success rate in the long term. ⁽¹⁹⁻²²⁾ Several authors consider occlusal load a crucial factor affecting the dental implant healing phase and the long-term survival and success of dental implants. ⁽²³⁻²⁹⁾ In teeth, a semi-elastic connection between the tooth and bone exists (PDL), whereas in implants, a direct and relatively rigid connection between the bone and implant is achieved if healing without complications has taken place. ^(30,31) Therefore, a direct transmission of forces on the peri-implant bone without any shock-absorbing element is consequent to implant loading. ⁽³¹⁾ It can usually be achieved by the adaptation capacity of peri-implant bone architecture toward changing load conditions. ⁽³²⁻³³⁾ According to Frost, ⁽³²⁻³³⁾ within the range of a physiologic loading, bone undergoes its physiologic turnover. In mild overloading, below bone's microdamage threshold, modeling drifts can begin adding to and/ or reshaping bone. But in the case of a pathologic overload, bone fractures and bone resorption may occur. ⁽³²⁻³³⁾ For these reasons,

it appears to be important to control the forces transmitted on the bone-implant interface.

Implant supported restorations materials should have resiliency in order not to transmit excessive occlusal forces to underlying bone with subsequent die-back of bone and increased risk for implant failure.⁽³⁴⁾ Their occlusal table should be also 100 μ below occlusal table of adjacent natural teeth. So teeth would “sink” first into PDL before occlusal table of implant supported prosthesis comes into contact with opposing dentition,⁽³⁵⁾ in order to protect bone around implant from excessive forces. Occlusal forces can be minimized on implant supported prosthesis by either the material of the prosthesis itself with introduction of veneering composite: composite fused to metal CFM or composite fused to zirconia CFZ. Recently hybrid ceramic material was also introduced; Vita Enamic (VITA Zahnfabrik H, Germany), Lava Ultimate (3M ESPE, USA), Paradigm (Dentsply, Weybridge, UK). It has a low modulus of elasticity and acts as a stress breaker against occlusal forces.⁽³⁶⁾ It is composed of composite matrix filled with different amounts and types of ceramic fillers.

All researches done before targeted to test the shock absorbing capacity of restoration material. The shock absorbing capacity of cement itself has never been addressed nor tested.

Can the cement act as a stress breaker, absorbing and neutralizing part of the occlusal stresses, and sparing both itself and veneer material? The hypothesis of the research that the new experimental adhesives may exhibit lower tensile and shear strengths than the tested cements.

MATERIALS AND METHOD

Samples preparation for tensile strength test

An acrylic upper right maxillary first molar #16 (Kavo, Germany) was mounted in a 10x10x20 mm stone block. The molar was prepared to a 0.5 mm chamfer finish line using round-end taper diamond bur with guiding pin # 8881 P (Komet, Brasseler, Germany). 1.5 mm occlusal reduction was done using round-edge wheel stone (5909-314-040, Komet, Brasseler, Germany). The stone block containing prepared acrylic tooth was marked “M1”. Using industrial silicon (Body Double, Smooth-On, Inc., Pennsylvania, USA), a mold was fabricated for “M1”. 28 replicas for “M1” were fabricated using acrylic resin (Vertex-Dental B.V., The Netherlands) “T1”. A wax coping was fabricated on “M1”. A silicon index was fabricated to the wax coping, then filled with molten wax and mounted on each “T1” to construct 28 standardized wax copings. The margins were then reflowed and adjusted using marginal wax (Renfert GmbH, Hilzingen, Germany) to ensure proper marginal integrity. (Fig.1)



FIG. (1) Wax patterns over resin dies.

An occlusal loop was attached to each coping. Wax patterns were sprued (Fig. 2) and invested each 12-17 copings in one ring to be cast into Co-Cr copings (Remanium Star, Dentaureum, Germany). (Fig.3 a&b) Copings were finished using heatless stones (Koolies, Dedeco Inc., USA) and inspected for any deformity. They were air particle abraded using $100\ \mu\text{m}\ \text{Al}_2\text{O}_3$ at 80 psi and then were cleaned using ultrasonic water bath (BioSonic, UC 125, Coltène/Whaledent AG, Switzerland). Copings were labeled and divided into 4 groups according to cement used. (Table 1) Each cement was manipulated according to manufacturer's directions and applied to the fitting surface of metal coping. Each coping was cemented to its corresponding "T1" and held under

static load of 5 kg. for 10 minutes. Excess cement was removed using a sickle scaler (Hu-Friedy Mfg. Co., Chicago, USA). (Fig.4)

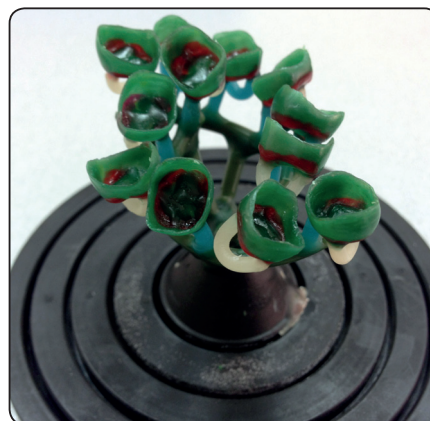


FIG. (2) Wax patterns attached to cone former.

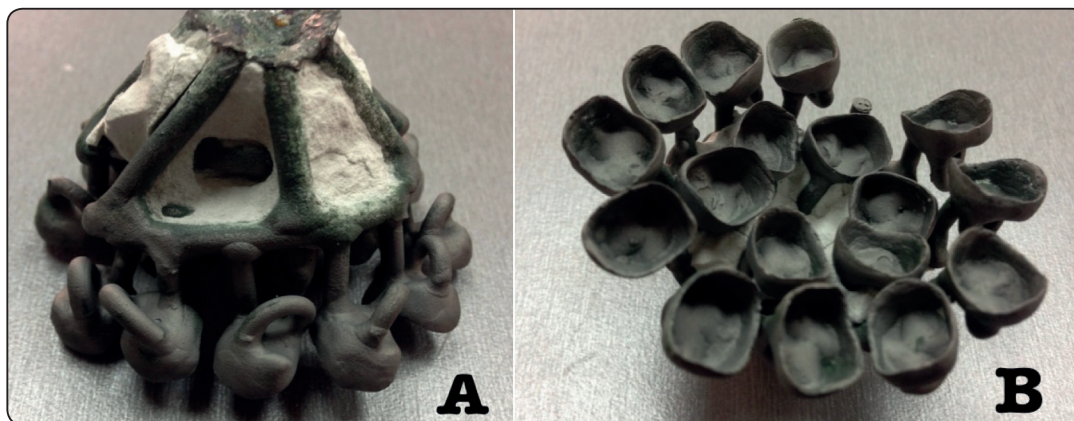


FIG. (3) Co-Cr copings.



FIG. (4) Cemented copings over resin dies.

TABLE (1) Sample grouping

Group	Type of cement	Type of test	No of samples
A	Glass Ionomer cement (Ketac-Cem Apilcap, 3M ESPE, USA)	Shear Bond test	7
		Tensile test	7
B	EMGS cement (Dow Corning Corporation, USA)	Shear Bond test	7
		Tensile test	7
C	EMGR cement (Dow Corning Corporation, USA)	Shear Bond test	7
		Tensile test	7
D	Resin cement (Rely X U200, 3M ESPE, USA)	Shear Bond test	7
		Tensile test	7
Total			56

Samples preparation for Shear bond strength test

Twenty-Eight caries and restoration-free human upper right maxillary first molar #16 of approximate sizes were collected from clinics of BAU. Teeth were cleaned and sterilized in an autoclave at 121°C, 15 Psi for 40 minutes. Teeth were then embedded completely in an auto-polymerized transparent acrylic resin (Vertex-Dental B.V., The Netherlands) blocks with a size of 10 x 10 x 20 mm. After complete setting, the blocks were mounted and trimmed on a trimmer (SQM 25 N, Zhermack SpA Via Bovazecchino, Italy) to expose 4x4 mm patch of dentine on their buccal side “S1”. The exposed dentine surface was cleaned, polished using polishing paste (Prophy paste,

Procter & Gamble, USA) on a rotary cup (Upgrade Disposable Prophy Angle’s and prophy cup, Sultan healthcare, Hackensack Avenue, USA). (Fig. 5A) A wax cube was constructed having dimensions of 3x3 mm and thickness of 2 mm. A silicon mold was constructed for standardized fabrication of 28 wax pattern cubes. The wax cubes were cast into Cr-Co cubes, sandblasted, ultrasonically cleaned and dried “C1”. (Fig.6) Each cement was manipulated according to manufacturer’s directions and applied to the fitting surface of metal cubes “C1”. Using a tweezers (Hu-Friedy Mfg. Co., Chicago, USA) each cube was cemented to its corresponding patch of dentine “S1” and held under static load of 5 kg. for 10 minutes. Excess cement was removed using a sickle scaler (Hu-Friedy Mfg. Co., Chicago, USA).

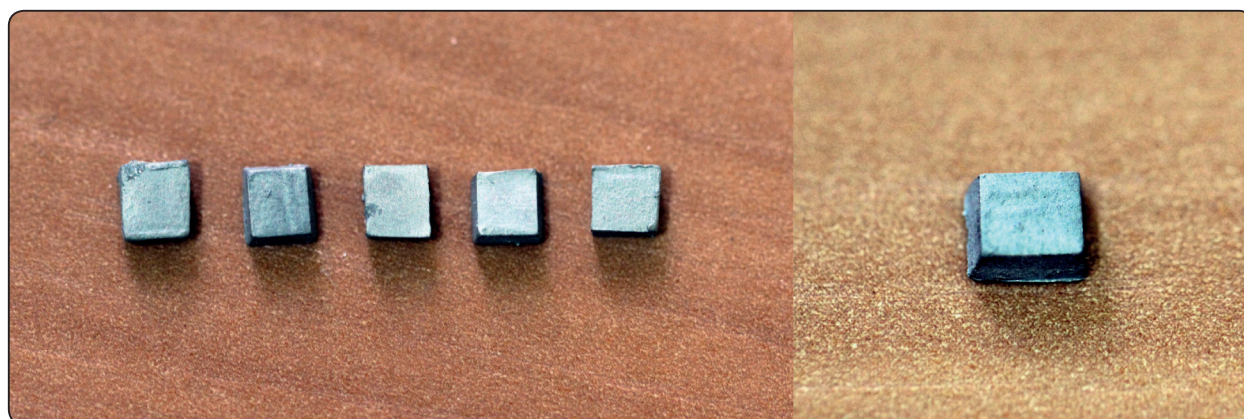


FIG. (5) Cr-Co cubes.

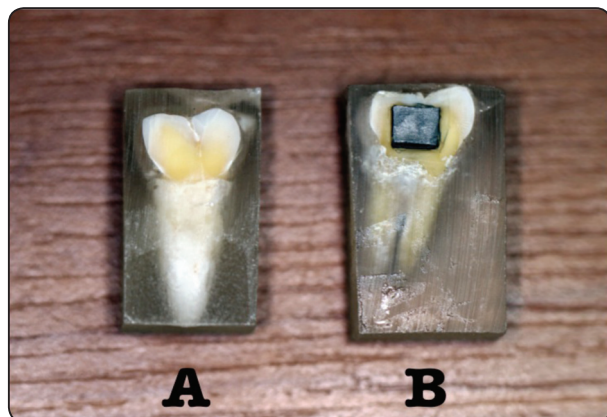


FIG. (6) A: Trimmed tooth showing exposed dentine in acrylic block. B: Cemented Cr-Co cube to exposed dentine surface.

All Tensile and shear bond strengths specimens were stored in distilled water for 72 hours, then thermocycled for 300 cycles with the sequence of 20 sec at 5°C and 20 sec at 55°C and 10 sec transport.

Tensile strength testing

Samples were submitted to a tensile test using a universal testing machine (Instron 8874; Instron Corp. Canton, Mass) under a constant crosshead speed of 1.00 mm/min. The fracture was confirmed by sudden drop in force measurements in the testing machine. (Fig.7)



FIG. (7) Sample during tensile testing.

Shear bond strength testing

A square interface shear test was designed to evaluate the bond strength. All samples were individually mounted on a computer-controlled materials testing machine (Model LRX-plus; Lloyd Instruments Ltd., Fareham, UK) with a loadcell of 5 kN and data were recorded using computer software (Nexygen-MT; Lloyd Instruments). Samples were secured to the lower fixed compartment of testing machine by tightening screws. Shearing test was done by compressive load applied at Cr-Co dentine interface using a mono-bevelled chisel shaped metallic rod attached to the upper movable compartment of testing machine traveling at cross-head speed of 0.5 mm/min. The load required to debonding was recorded in Newton. (Fig.8)

Bond strength were calculated using the following equation

The load at failure was divided by bonding area to express the bond strength in MPa : $\tau = P / L^2$ where; τ =shear bond strength (MPa, P =load at failure(N) and L =arm of the square(mm).

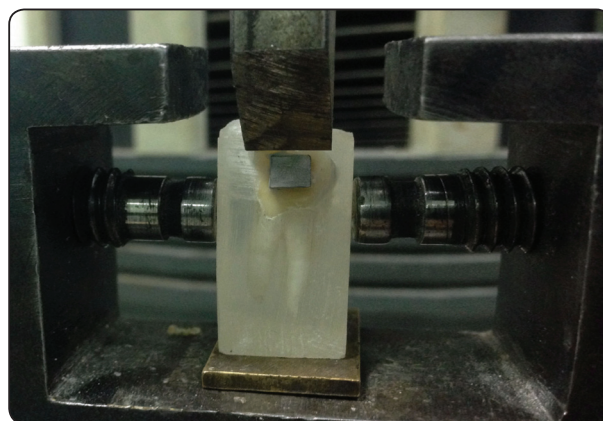


FIG. (8) Sample during shear bond testing.

RESULTS

The mean, median and standard deviation of tensile force values between groups, are summarized in. (Table 2 & Chart 1) The highest mean value was observed in group D (611.5 MPa) and the lowest mean value was observed for group C (12.86 MPa). Statistical analysis revealed no significant difference between groups ($P= 1.03$). Statistical analysis among groups showed significant difference between groups A and B ($P= 0.003$), between groups A and C ($P= 0.02$) and between groups B and C ($P= 0.0001$). But no significant differences were found between groups A and D ($P= 8.11$), groups B and D ($P= 4.36$) or between groups C and D ($P= 3.96$). (Table 3) Regarding shear bond strength, the mean, median and standard deviation values between groups, are summarized in Table 4 & Chart 2. The highest mean value was observed in group D (13.23 MPa) and the lowest mean value was observed for group C (0.2 MPa). Statistical analysis revealed no significant difference between groups ($P= 4.93$). Statistical analysis among groups showed significant difference between groups A and C ($P= 0.0002$), between groups A and D ($P= 0.01$), between groups B and C ($P= 0.01$) and between groups C and D ($P= 0.0002$). But no significant differences were found between groups A and B ($P= 1.16$), or between groups B and D ($P= 9.22$). (Table 5)

TABLE (2) Mean, Median and standard deviation of tensile force values of the tested groups.

	Mean	Median	Standard Deviation
Group A	254.64	262.32	79.2
Group B	48.36	48.52	5.77
Group C	12.86	14.44	2.82
Group D	611.5	624.92	56.5

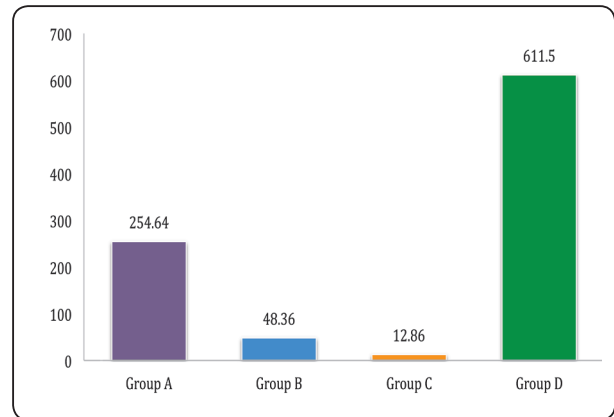


CHART 1: Mean tensile force values of tested groups

TABLE (3) One-way ANOVA tensile force values among tested groups.

	Group A	Group B	Group C	Group D
Group A		0.003	0.02	8.11
Group B	0.003		0.0001	4.36
Group C	0.02	0.0001		3.96
Group D	8.11	4.36	3.96	

TABLE (4) Mean, Median and standard deviation shear bond strength values of the tested groups.

	Mean	Median	Standard Deviation
Group A	8.39	8.68	1.57
Group B	0.78	0.8	0.25
Group C	0.2	0.18	0.04
Group D	13.23	14.47	2.51

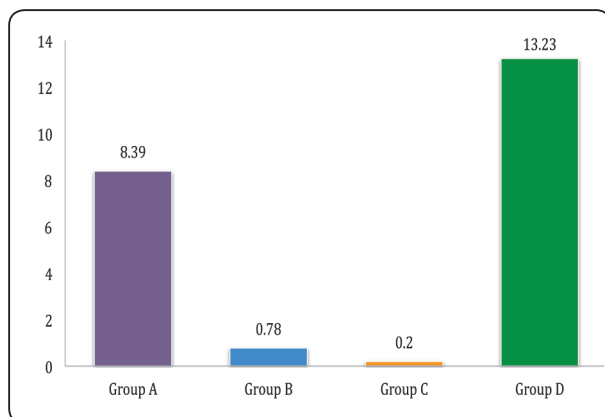


CHART 2: Mean shear bond strength values of tested groups.

TABLE (5) One-way ANOVA shear bond strength values among tested groups.

	Group A	Group B	Group C	Group D
Group A		1.16	0.0002	0.01
Group B	1.16		0.01	9.22
Group C	0.0002	0.01		0.0002
Group D	0.01	9.22	0.0002	

DISCUSSION

The purpose of this study was to evaluate the shear bond and tensile strengths of two types of medical grade adhesives. Each of them belong to a category never been used in dentistry before, but are already being used in the medical field for few years. They exhibit stress-breaking property and used in wet conditions under skin and in close contact with blood. The two adhesives could be a good choice for cementation of implant supported restorations and conditions needing stress breaking action like pier-abutment situation, to avoid overloading. True that, clinical evidence on the impact of overloading on peri-implant bone is not available. Only some case reports⁽³⁷⁻³⁹⁾, in vitro⁽⁴⁰⁻⁴⁵⁾ and animal studies^(26,29) are present. Some authors

claimed that the type of material used for the implant supported prosthesis could affect occlusal load.^(31,46-52) In particular, in the 1980s, some investigators recommended resilient occlusal materials such as acrylic resin to reduce the forces exerted on implants.^(31,53,54) However, contrasting results on this topic⁽⁵⁵⁻⁵⁸⁾ suggest the need for further investigation. The role of dental materials in occlusal stress transmission onto peri-implant bone seems to be especially relevant over the past few years because of the increasing use of esthetic but rigid materials, such as glass-ceramic and zirconia. These materials are reported to have excellent mechanical and biologic properties,^(59,60) but their impact on peri-implant bone and on the whole masticatory system has not yet been investigated.

According to Skalak,⁽³¹⁾ the viscoelastic behavior of an acrylic resin as occlusal material would be enough to delay the transmission of force and reduce its peak compared with materials with greater elastic moduli. An in vitro study by Gracis et al⁽⁵²⁾ concluded that the harder and stiffer the material, the higher the force transmitted onto the implant. In fact, according to Hooke's law, the higher the modulus of elasticity of a material, the less the material will deform under pressure and the more likely the force will be transferred through the material.⁽⁶¹⁾ On the other hand, a review of the literature over the last 20 years demonstrated that many articles contradict the existence of a shock absorption capacity of resilient dental materials.⁽⁶²⁻⁶⁹⁾

The choice of the restorative material to be used in implant restorations should be made in light of newly introduced concepts of osseosufficiency and osseoseparation⁽⁷⁰⁾: as long as the host, the implant, and the clinical procedures induce and allow for maintaining osseointegration, an osseosufficiency state is present. But some patient-related or nonpatient-related factors could induce osseoseparation, compromising the obtainment or maintenance of osseointegration.

As mentioned before all the previous studies were concerned about the material of the prosthesis. None was directed towards the stress-breaking action of the cement itself.

Studies on shear bond and tensile strength showed that it varied significantly depending on the resin cement used⁽⁷¹⁻⁷⁷⁾, bonding agents⁽⁷⁸⁻⁸¹⁾ or between enamel and dentine^(71,75). Lührs AK, et al, found that the shear bond strength of self-adhesive resin cements was inferior compared to conventional composite resin cements.⁽⁸²⁾ But less sensitive to variations in handling and aging and thermocycling.⁽⁸³⁾

Results of this study showed that there were no significant difference between groups in both tensile strength and shear bond strength. The highest mean tensile strength value was observed in group D (611.5 MPa) and the lowest mean value was observed for group C (12.86 MPa). The highest mean shear bond strength value was observed in group D (13.23 MPa) and the lowest mean value was observed for group C (0.2 MPa).

Results of the experimental medical grade rubber cement (EMGR) came as a disappointment as it is far below levels needed intraorally. Its current possible use as a cement is highly questionable. On the other, experimental medical grade silicon adhesive (EMGS) showed both higher shear bond strength and tensile strength than that of EMGR. The formula needs further improvement to cope with the oral stresses. There are no comparable studies done on both formulas.

The idea of stress breaking cement might work out after all. The biocompatibility issue of both used cements are solved as both of them are being under skin in contact with blood for medical appliances. They have tolerance against moisture as they are both used to seal medical tubes and electrodes. The issue of strength is the problem as both formulas are not designed to withstand masticatory forces. Further improvement in the silicon adhesive might be helpful to tailor the experimental cement to oral function.

CONCLUSIONS

Within the limitations of the study it could be concluded that; none of the two experimental medical grade adhesives showed comparable results to either GIC or resin cements.

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