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FRACTURE RESISTANCES OF CAD/CAM MONOLITHIC CROWNS WITH DIFFERENT OCCLUSAL THICKNESS

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ABSTRACT

The aim of this study was to examine the effect of different occlusal thickness on fracture resistances of different monolithic crowns.

Materials and methods: Forty molar crowns were fabricated from four different types of monolithic ceramic blocks (n=10): Group I: Vita Enamic (V.enamic), Group II: IPS E-max CAD (e.max), Group III: Celtra Duo(CD), Group IV: Functional explore (f.explore); by means of CAD/CAM system with two occlusal thickness (1 and 1.5mm). All crowns were cemented on implant abutments using self adhesive resin cement. Combined thermocycling and mechanical loading was performed under a chewing simulator, fracture resistance was examined by universal testing machine. The fractured specimens were inspected by scanning electron microscopy (SEM).

Results: The highest fracture resistance was recorded with f.explore group followed by CD group then e.max group while the lowest fracture resistance mean value recorded with V.enamic with statistical significance ($P=0.0001 < 0.05$) for both occlusal thicknesses. The 1.5 mm occlusal thickness recorded statistically non-significant ($P=0.2267 > 0.05$) higher fracture resistance mean value than 1mm occlusal thickness except for f.explore group.

Conclusions: The fracture resistance of monolithic crowns was differently affected by the ceramic materials and changes in occlusal thickness. Within the limitations of this study, all the tested crowns resisted the physiological range of masticatory forces at 1.5mm and reduced (1.0 mm) occlusal thickness.

KEY WORDS: Fracture resistances, monolithic crowns, occlusal thickness.

INTRODUCTION

Implant-supported prostheses are usually made using computer-aided design/computer-assisted manufacturing (CAD/CAM) technologies.[1] This CAD/CAM process assume to guarantee acceptable marginal gap and acceptable strength due to the absence of air bubbles inside the material.[2] Accordingly, the fracture resistance of restorations made from CAD/CAM blocks is high compared to those hand-processed by a technician.[3] The higher

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** Associate Professor, Fixed Prosthodontics, College of Dentistry, MTI University fracture resistance is of particular importance in the posterior regions of the jaw, where the average masticatory force can be up to 600 Newton. [4]

High survival rates of ceramics used in posterior fixed prosthesis was shown to be comparable to those of conservative single restorations in short-term studies.(5,6) Among the most commonly used ceramic restorations in posterior region are monolithic lithium disilicate and yttria-stabilized zirconia. Both were shown to have satisfactory clinical outcomes.(7)

Lithium disilicate glass ceramics have expanded popularity among dentists because of their high mechanical properties, great esthetics potential and wear behavior very similar to opposing dental enamel. (8)

LD ceramics can be used with both conventional pressing techniques and Computer Aided Design-Computer Aided Manufacturing (CAD/CAM) fabrication techniques; the latter permit for standardized processing of the material, decrease manufacture time and improve cost effectiveness. (9)

A recently developed ceramic called zirconia-containing lithium silicate (ZLS) relies on the addition of 10 mass percent zirconium oxide to lithium silicate glass compositions. Zirconia acts as nucleating agent but remains in solution in the glassy matrix, with two main consequences: A dual microstructure consisting of very fine lithium metasilicate (Li_2SiO_3) and lithium disilicate ($Li_2Si_2O_5$) crystals is obtained. Ceramic blocks of this material are available for CAD/CAM construction.(10)

Recently, a hybrid ceramic with interpenetrating polymer and ceramic networks has been technologically innovative in order to combine the advantages of ceramics which simulate the physical properties of natural teeth and the advantages of polymers that have less brittleness and avoid causing wear on antagonistic tooth.(11) The polymer infiltrated ceramic network (PICN) shows equivalent modulus of elasitcity to human dentin.[16,17] It has a high hardness due to the limitation of crack propagation caused by the two interconnected phases.(11), [12] The energy of external force exerted on the surface can be absorbed by plastic deformation and creep of the PICN. (13)

Numerous readings have described that the fracture resistance of posterior ceramics restorations is detrimental to treatment achievement. (14,15) However, there is no agreement on a minimum recommended thickness for monolithic lithium disilicate single crowns that is confirmed by scientific data yet and there is no agreement on how thin restorations can be made (14,16). To date, few laboratory data about the mechanical predictability of monolithic crowns are available in the literature, particularly for the so called "ultrathin" configuration (i.e. up to a thickness of 0.5 mm), as well as the validation of their clinical performances in the oral environment (8,9)

However, limitation in interocclusal space may dictate a compromise in abutment height as well as occlusal thickness of the final restoration. A materials that show sufficient fracture strength in reduced thickness, could be desirable in various clinical situations. For evaluation of new materials, in vitro tests is required combining reproducible laboratory conditions with basic requirements (occlusal loading, thermocycling) of the clinical situation.(17)The purpose of this study was to investigate the fatigue and fracture resistances of different monolithic crowns fabricated by computer-aided design and computer-aided manufacturing (CAD/CAM) with different occlusal thickness. The null hypothesis of the present study was: the fracture loads of the CAD/CAM-fabricated monolithic crowns will not differ by ceramic type and occlusal thickness.

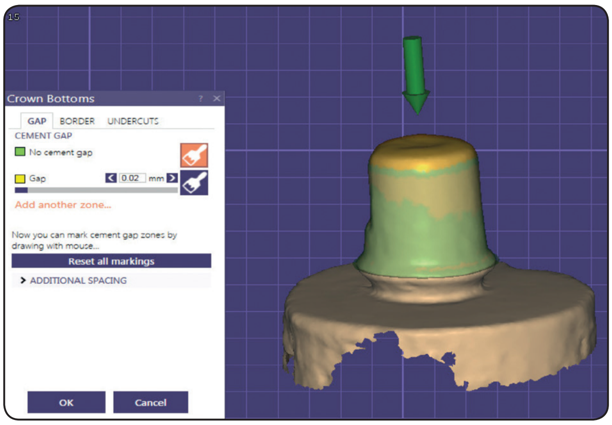
MATERIALS AND METHODS

Forty implant lab analog (Neobiotech C0., Ltd. Guro-gu, Seoul, Korea) [1] were attached to implant impression copings (IS Hexed Pick-up Impression coping 4.0/Long) and the implant analogs were mounted vertically in resin blocks (Cold cure denture base material, Acrostone Dental Factory) (Fig.:1). After setting of the resin mold, the impression copings were removed and straight IS Cemented Abutment (4.5*1.0mm, L: 7.0mm Hex) (Neobiotech C0., Ltd. Guro-gu, Seoul, Korea) were screw attached according to manufacturer's instructions. The abutments were shortened to 4 mm using diamond disc mounted on a low-speed hand piece held by a parallelometr. The screw holes of the abutments were filled with visible light activated composite (Filtek Z350 XT Universal Restorative System (3M-ESPE AG Dental products ST. Paul, MN 55144 USA), (Fig.2) light-cured for 40 s with a LED curing unit (Elipar S10, 3M ESPE, Seefeld, Germany).

Fig. (1): Implant analog attached to impression copings, mounted vertically in resin block.

Fig. (2): The abutment was shortened to 4 mm using diamond disc(a) The screw holes of the abutments were filled with visible light activated composite(b)

All abutments were sprayed with Scan spray (Renfert GmbH Industriegbiet 78247 Hilzingen/ Germany), scanned using Scan Box (Smart Optics Sensortechnik GmbH, Germany) and imported into the CAD/CAM software (InLab version 4.0; Sirona) except f. explore it was milled using (Roland, MDX-40; Japan) (Fig.: 3).



Gap 0.02 mm

Now you can mark cement gap zones by drawing with mouse...

Fig. (3): Scanned abutment.

The ideal anatomical tooth form in EXOCAD Valetta 2.2 software library (Exocad GmbH) was used as the reference design, (Fig. 4). Two designs were made to make two groups of crowns (table 1):

Group A (N=20): The reference design of anatomical form of lower right first molar with 1.5 mm thickness at the central fossa.

Group B (N=20): The reference design of anatomical form of lower right first molar with 1.0 mm thickness at the central fossa.

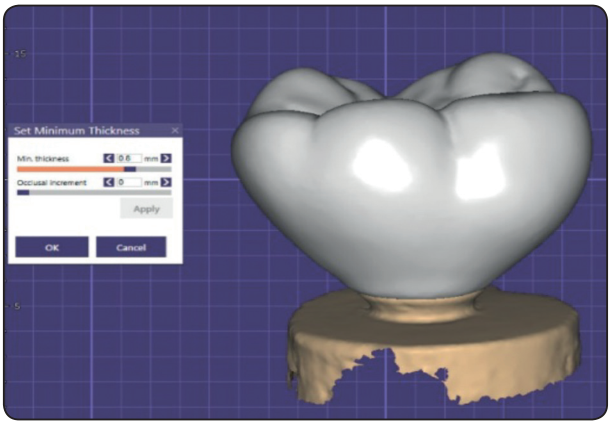
Each group included 4 subgroups, according to the material of which the crows were milled:

Subgroup 1 (N=5): Vita Enamic (V.enamic) (VITA Zahnfabrik, Bad Säckingen, Germany).

Subgroup 2 (N=5): IPS E-max CAD(e.max) (Ivoclar Vivadent, Schaan, Liechtenstein).

Subgroup 3 (N=10): Celtra Duo(CD) (Dentsply DeTrey, Konstanz, Germany)

Subgroup 4 (N=10): Functional explore (f.explore) (Shenzhen Upcera Dental Technology



Min. thickness 0.6 mm

Occlusal increment 0 mm

Fig. (4): The design of anatomical tooth form of lower right first molar crown in EXOCAD Valetta 2.2 software.

Co High-tech Industry Park, Nanshan District, Shenzhen, Guangdong, China. Provided by Ivory Trade International, Egypt).

TABLE (1): Samples grouping:

	Group A (1.5 mm)	Group B (1.0 mm)	
Subgroup 1 (V.enamic)	N=5	N=5	N=10
Subgroup 2 (e-max)	N=5	N=5	N=10
Subgroup 3 (CD)	N=5	N=5	N=10
Subgroup 4 (f.explore)	N=5	N=5	N=10
Total			N=40

For all groups the cement layer thickness was set to 50 µm. The manufacturer's information of the monolithic ceramics used in this study was shown in Table 2. The design data of the CAD-based lower right first molar crowns were used to establish the crowns. Eight experimental groups with two different occlusal thicknesses: 1.5 mm (recommended condition) and 1.0 mm (experimental condition).

Four different CAD/CAM ceramic blocks including ; the PICN; Vita Enamic (v.enamic); VITA Zahnfabrik, Bad Säckingen, Germany) finished with rubber (Eve America; Naples, FL); the lithium disilicate; IPS E.max CAD (e.max); (Ivoclar Vivadent, Schaan, Liechtenstein), sintered and finished a with a glaze cycle (Programat P300; Ivoclar Vivadent) ; zirconia- reinforced lithium silicate, Celtra Duo (CD); (Dentsply DeTrey, Konstanz, Germany) Crystallization and glazing process were performed in a furnace (Programat P310; Ivoclar Vivadent) according to the corresponding manufacturer's protocols and f.explore were fabricated into molar-shaped crowns by using four-axis milling machine (Trione G; Dio, Busan,Korea) by Advanced Dental Studio - Egypt. And sintering in (Tabeo, Mihm Vogt, Germany).

All abutments were sand blasted by 50-μm aluminum oxide particles (Renfert, Hilzingen, Germany) at a distance of 10 mm. The specimens were cleaned in distilled water for 3 minutes in ultrasonic bath (Ultrasonic Cleaner; Biem Ultrasonic Makina San. Ltd. Şti., Turkey) in order to remove any surface deposits.

Prior to the cementation, the inner surface of each ceramic crown was treated according to the manufacturer's recommendations. For f.explore group; the intaglio surfaces of the zirconia crowns were air born particle abraded by 50 µm Al_2O_3 particles at 1 bar pressure then ultrasonically cleaned and air dried.

For CD groups, the inner surfaces of the crowns were washed with ethanol, treated with 5% hydrofluoric acid gel (Ceramic etching gel; Ivoclar Vivadent) for 20 seconds, and then washed with water. After removing the residual acid with water, they were air-dried and coated with silane (Monobond N; Ivoclar Vivadent).

For v.enamic group, the inner surfaces were washed with ethanol, treated with 5% hydrofluoric acid gel (Ceramic etching gel; Ivoclar Vivadent) for 60 seconds, and then washed with water for 60 seconds. Afterwards, they were air-dried for 20 seconds and coated with silane (Monobond N; Ivoclar Vivadent).

TABLE (2): All-ceramic crown materials used in the study

Material	**Description, properties & composition**	**Manufacturer**
hybrid ceramic (Vita Enamic) (groupI)	Double network hybrid ceramic block, Shade 3M2-HT EM-14 **Flexural strength** : 150-160 MPa **Fracture toughness:** 1.5 MPa m½ **Elastic modulus :** 30 GPa ***Composition of the ceramic part : (86 wt% / 75 vol%)*** Silicon dioxide SiO2 58 – 63% Aluminum oxide Al2O3 20 – 23% Sodium oxide Na2O 9 – 11% Potassium oxide K2O 4 – 6% Boron trioxide B2O3 0.5 – 2% Zirconium dioxide ZrO2 < 1% Calcium oxide KaO < 1% ***Composition of the polymer part : (14 wt% / 25 vol%)*** UDMA (urethane dimethacrylate) TEGDMA (triethylene glycol dimethacrylate)	VITA Zahnfabrik Germany
(group II) Lithium-disilicate glass ceramic (IPS e. max CAD) (group II)	Lithium disilicate glass-ceramic block **Flexural strength** : 360 ± 60 MPa **Fracture toughness**: 2.0 – 2.5 **Elastic modulus** : 95 ± 5 GPa ***Composition in Wt%:*** Silicon dioxide SiO2 57.0 – 80.0 Lithium dioxide Li2O 11.0 – 19.0 Potassium oxide K2O 0.0 – 13.0 Phosphorus pentoxide P2O5 0.0 – 11.0 Zirconium dioxide ZrO2 0.0 – 8.0 Zinc oxide ZnO 0.0 – 8.0 Other & coloring oxides 0.0 – 12.0	Ivoclar Vivadent Schaan, Liechtenstein
Zirconia-reinforced Lithium Silicate (ZLS)Celtra duo	10% zirconium dioxide (ZrO2) in highly dispersed form in the glass phase of the ceramic blocks. **Flexural strength** : 370 MPa **Fracture toughness** : 2.56MPa m½ **Elastic modulus** : 70 GPa ***Composition of the ceramic part*** Zirconia 10% Silica, Lithium metasilicate and phosphate crystals 58%	Dentsply DeTrey, Konstanz, Germany
Yttria -stabilized tetragonal zirconia polycrystalline (Upcera) (group VI)	Upcera Zirconia , specialized for Full Contour Zirconia restorations Multi Layered Zirconia disc with 98.5mm diameter and 18 mm thickness, shade A light **Flexural strength** : 1125 MPa **Fracture toughness** :5.1 MPa m½ **Elastic modulus** : 214 GPa ***Composition in Wt%:*** Zirconium dioxide ZrO2 +HfO2 90-95% Yttrium Oxide Y2O3 5-8 % Other oxides CaO, MgO 0-2%	Shenzhen Upcera Dental Technology Co, China

For e. max group, the inner surfaces were treated with 5% hydrofluoric acid gel (Ceramic etching gel; Ivoclar Vivadent) for 20 seconds, washed with water and air-dried, and coated with silane coupling agent (Monobond N; Ivoclar Vivadent).

Subsequently, self-adhesive resin cement (RelyX U200; 3M ESPE) was used to bond the crowns to the abutments according to the manufacturer's protocol.

The crowns were seated onto the implant abutment with finger pressure and then 5 kg were applied onto each crown using a specially designed loading device, for 10 minutes. Cement excess was removed with a micro brush and each surface was light-cured for 40 s with a LED curing unit (Elipar S10, 3M ESPE, Seefeld, Germany).

The bonded crown/abutment specimens were stored in distilled water at 37°C for 24 hours prior to thermal and mechanical cycling. [5]

Thermo-cycling Mechanical loading (TCML) was performed using a programmable equipment; the newly developed four stations multimodal ROBOTA chewing simulator integrated with thermo-cyclic protocol operated on servo-motor (Model ACH-09075DC-T, AD-TECH TECHNOLOGY CO., LTD., GERMANY) ROBOT. A chewing simulator which has four chambers simulating the vertical and horizontal movements simultaneously in the thermodynamic condition. Each of the chambers consists of an upper Jackob's chuck as hardened steel antagonist holder that can be tightened with a screw and a lower plastic sample holder in which the specimen can be embedded. The specimens were embedded in chemical cured acrylic mold which in turn fixed by tightening screw to teflon holder in the lower part of simulator. A weight of 5 kg, comparable to 49 N of chewing force was exerted. The test was repeated 75,000 times to clinically simulate the 6 months chewing condition, according to previous studies. (18) Chewing simulation test parameters Vertical movement: 3 mm. Horizontal movement: 1 mm. Rising speed: 90 mm/s. Forward speed: 90 mm/s. Descending speed: 40 mm/s. Backward speed: 40 mm/s. Cycle frequency 1.6 Hz Weight per sample: 5 kg Torque; 2.4 N.m. During TCML all crowns were controlled daily for failures and failed crowns were excluded from further simulation and testing.

Fracture resistance testing

All samples were individually mounted on a computer-controlled material testing machine (Model 3345; Instron Industrial Products, Norwood, MA, USA) (Fig. 5) with 5 kN load cell and data were documented using computer software (Instron® Bluehill Lite Software). Samples were secured to the lower fixed compartment of testing machine by tightening screws. Fracture test was done by compressive mode of load applied occlusally using a metallic rod with spherical tip (5.8 mm diameter) attached to the upper movable compartment of testing machine traveling at cross-head speed of 1mm/min with tin foil sheet in-between to achieve homogenous stress distribution and minimization of the transmission of local force peaks. The load required to fracture was recorded in Newton. After the testing, the fractured surface of each specimen

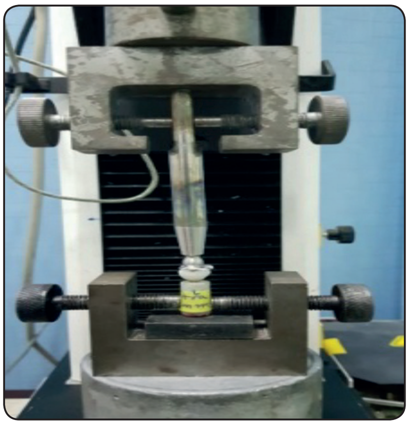


Fig. (5): sample mounted on a computer-controlled material testing machine.

was observed under a stereomicroscope at a magnification of (36-50X), and one representative specimen from each group was randomly selected for the fractographic analysis. The fractographic analysis was conducted to examine the fracture characteristics of the various ceramic crowns and to identify the crack propagation direction, under a scanning electron microscope (SEM) (JEOL JSM-T 20. Japan) operating at 20.0 Kv, at (36-50X) for fracture pattern examination.

Statistical analysis

The results were analyzed using Graph Pad Instat (Graph Pad, Inc.) software for windows. A value of $P \leq 0.05$ was considered statistically significant. Continuous variables were expressed as the mean and standard deviation. After homogeneity of variance and normal distribution of errors had been confirmed, Analysis of variance was used to compare materials. Student t-test was done for compared pairs. Two-factors analysis of variance for each factor. Sample size (n=5) was large enough to detect large effect sizes for main effects and pair-wise comparisons, with the satisfactory level of power set at 80% and a 95% confidence level.

RESULTS

Fracture resistance

The results of fracture resistance (N) mean, standard deviation (SD) values for all groups with both occlusal thickness after mechanical aging are summarized in table (3) and graphically drawn in figure (6).

1mm occlusal thickness

The highest fracture resistance mean value recorded with f.explore group (3951 N) followed by CD group (1356.2 N) followed by e.max group (1008.6 N) while the lowest fracture resistance mean value recorded with V. Enamic group (811.7 N) and this was statistically significant as indicated by one-way ANOVA followed by Tukey's pair-wise tests (P=<0.0001< 0.05), as shown in table (3) and figure (6).

1.5 mm occlusal thickness

It was found that the highest fracture resistance mean value recorded with f.explore group (4663 N) followed by Celtra group (1617 N) then e.max group (962.23 N) while the lowest fracture resistance

TABLE (3): Fracture resistance results (Mean values in Newton ±SDs) for all groups with both occlusal thicknesses.

Variables			*Occlusal thickness*		*t-test*
			1 mm	*1.5 mm*	*P value*
Material type	*V. Enamic*	*Mean*	$\boldsymbol{811.7^{D}}$	$\boldsymbol{891.62^{C}}$	0.1914 ns
		SD	±54.58	±69.205	
	e.max	*Mean*	$\boldsymbol{1008.6^{C}}$	$\boldsymbol{962.23^{C}}$	0.6488 ns
		SD	±98.45	±130.4	
	Celtra	*Mean*	$\boldsymbol{1356.2^{B}}$	$\boldsymbol{1617^{B}}$	0.1221 ns
		SD	±152.1	±173.8	
	f.explore	*Mean*	$\boldsymbol{3951^{A}}$	$\boldsymbol{4663^{A}}$	0.0432*
		SD	±227.225	±356	
	ANOVA	*P value*	<0.0001*	<0.0001*	

Different letters in same column indicating significant (p<0.05) *; significant (p<0.05) ns; non-significant (p>0.05)

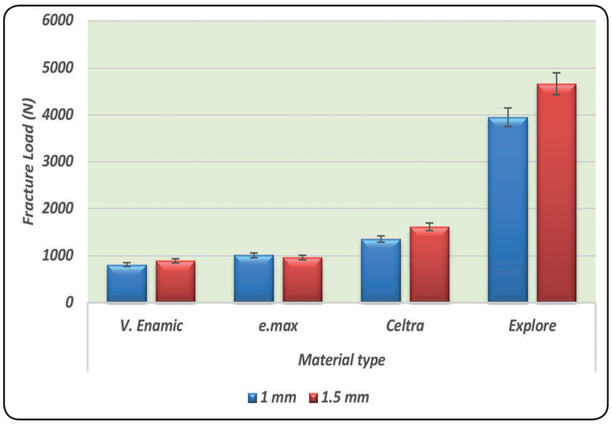


Fig. (6) Column chart of fracture load mean values for all groups with both occlusal thicknesses.

mean value recorded with V. Enamic group (891.62 N) and this was statistically significant as indicated by one-way ANOVA test (P=<0.0001 < 0.05). Pair-wise Tukey's post-hoc test showed non-significant (p>0.05) difference between (V. Enamic and e.max) as shown in table (3) and figure (6)

1 mm vs. 1.5 mm occlusal thickness

For V. enamic group; 1.5 mm occlusal thickness recorded statistically non-significant higher fracture resistance mean value (891.6 N) than 1mm occlusal thickness type mean (811.7 N) as proved by paired t-test (p=0.1914 > 0.05).

E.max group: 1 mm occlusal thickness recorded statistically non-significant higher fracture resistance mean value (1008.6 N) than 1.5 mm occlusal thickness type mean (962.23 N) as proved by paired t-test (p=0.6488 > 0.05)

CD group: 1.5 mm occlusal thickness recorded statistically non-significant higher fracture resistance mean value (1617 N) than 1 mm occlusal thickness type mean (1356.2 N) as proved by paired t-test (p=0.1221 > 0.05).

F.explore group: 1.5 mm occlusal thickness recorded statistically significant higher fracture resistance mean value (4663 N) than 1 mm occlusal thickness type mean (3951 N) as proved by paired t-test (p=0.0432< 0.05).

Influence of material group on fracture resistance

Regardless to cementation approach, totally it was found that the highest fracture resistance was recorded with f.explore group followed by CD group then e.max group while the lowest fracture resistance mean value recorded with v. Enamic with statistical significance as indicated by two-way ANOVA test (P=0.0001< 0.05). Pair-wise Tukey's post-hoc test showed non-significant (p>0.05) difference between (e.max and v. enamic) .

Effect of occlusal thickness on fracture resistance

Irrespective of material groups, totally it was found that 1.5 mm occlusal thickness recorded statistically non-significant higher fracture resistance mean value than 1mm occlusal thickness except for f.explore group as demonstrated by two-factors ANOVA test (P=0.2267 > 0.05).

Scanning Electron Microscope analysis

The failure mode of the samples was a combination of fracture and debonding, some of the f.explore crowns with occlusal thickness of 1.5 mm were not fractured even at 10 kN and associated with abutment fracture.

The SEM image of the fractured surface showed multiple cracks together with arrest lines, some compression curls, and twist hackle marks (Figs. 7 to 10). The origin of fracture seems to arise at the occlusal surface of the E.max and Enamic crowns, while for f. explore and CD groups the fracture seems to arise from the intaglio surface of the crowns. For the Enamic crowns, the cracks were shown to arise from the occlusal surface and then end before reaching the fitting surface. For the Emax crown, a tuft of cracks were arising from the loading point at the occlusal surface. For the f.explore and CD crowns, a main fracture line seems to arise from the fitting surface of the crown and the cracks extend to the outer proximal and cervical area.

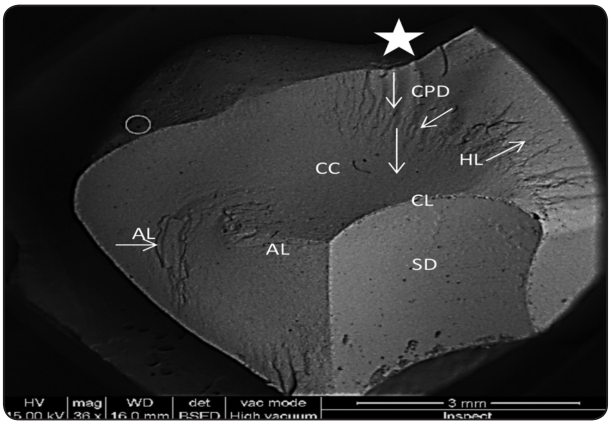


Fig. (7): SEM image of the fractured surface of v. enamic crown (group A1) with occlusal thickness of 1.5 mm. hackle line (HL), arrest line (AL), the star: where the fracture began, the arrows: the direction where cracks propagated, CC: compression curle (CPD): crack propagation direction, (CL) cementation layer, and (SD) supporting die.

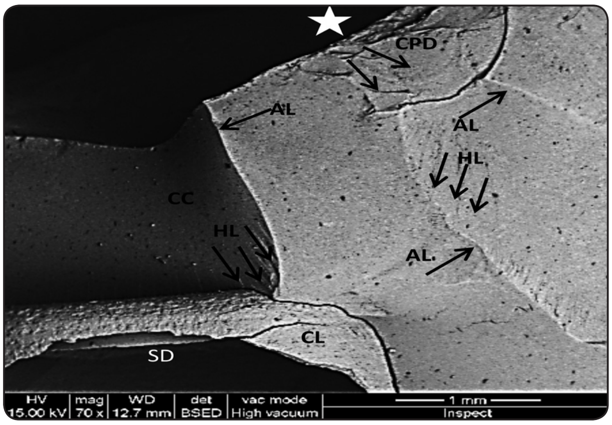


Fig. (8): SEM image of the fractured surface of e. max crown with occlusal thickness of 1.5 mm. HL: hackle line, AL: arrest line, the star: where the fracture began, TH: twist hackle, black solid arrow: crack propagation direction (CPD), CL: cementation layer, SD: supporting die

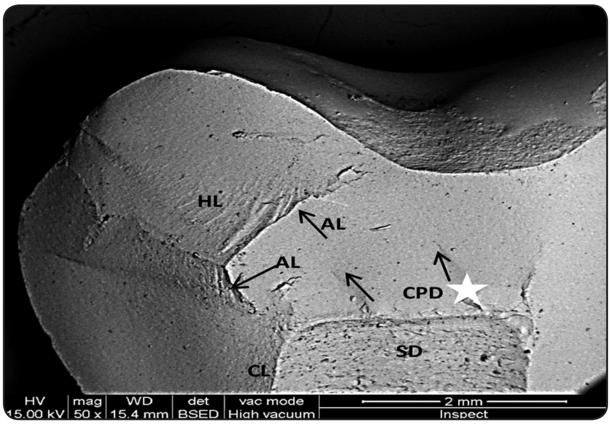


Fig. (9): SEM image of the fractured surface of CD crown with occlusal thickness of 1.5 mm. HL: hackle line, AL: arrest line, CC: compression curl, TH: twist hackle, white star: origin of fracture, black solid arrow: crack propagation direction (CPD), CL: cementation layer, SD: supporting die.

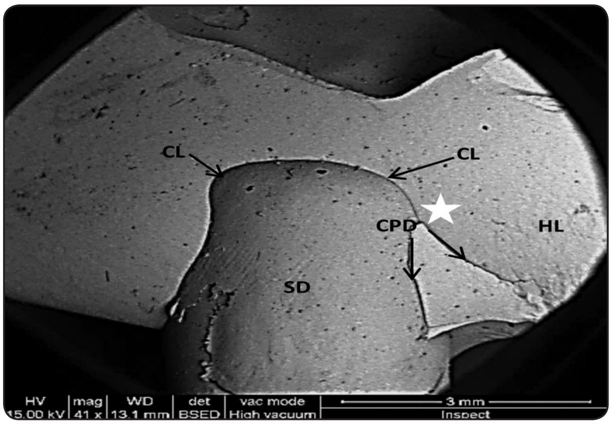


Fig. (10): SEM image of the fractured surface of f.explore crown with occlusal thickness of 1 mm with deeply located crack origin from the interface between the crown and cement layer.

DISCUSSION

Implant-supported crowns gained popularity as they were proved to have good clinical survival rate in long-term studies. (19) Specially when used with monolithic ceramic crowns showing aesthetic improvement above metal ceramic restorations, with adequate mechanical properties. (20) However, occlusal fractures occasionally were reported in clinical practice specifically with inadequate inter-arch space. This study evaluated four monolithic implant-supported crowns machined with CEREC system using the mandibular first molar as a reference.

In the present study, the CAD/CAM system was chosen, as it allowed for the use of high-quality materials, such as CAD/CAM-prefabricated blocks and this system also allowed for the standardization of manufactured crowns. In the monolithic restorations there is no further veneering layer, and hence the construction is simplified. (21)

Posterior restorations are subjected to high masticatory forces. The reported maximum biting forces in posterior region vary from the 600 to 900 N. The variations were shown to be influenced by facial morphology and age.[22]. In this study, the force needed to fracture the crowns surpassed the maximum biting force in the molar region regardless of their occlusal thickness, which means that all four materials tested are recommended for cemented implant-supported crowns.

Since the fracture resistance is influenced by the occlusal thickness of ceramic crowns, It was recommended to achieve 1.3 mm to 2.0 mm thickness at the occlusal surface of ceramic crowns. (23,24) This study assumed that clinical situations may dictate the use of 1.0 mm occlusal thickness; therefore ceramic crowns with 1.5 mm as generally accepted, as well as 1.0 mm were tested.

The standardization in the present study was simplified by using implant abutments because in natural teeth standardization regarding to properties and forms could be problemist.

In the current study, aging of the samples was done by using chewing simulator to exert intermittent occlusal equivalent to chewing force together with thermos-cycling. The load was repeated 75.000 times to clinically simulate a six months chewing condition, according to previous studies. (18)

Temperature changes and exposure to water were found to distress the strength properties of ceramic restorations, (25) and static fatigues with slow flaw propagation were reported. (26) Thus, samples of this study were subjected to thermocycling to simulate the clinical situation as carry out by many authors. (25-27)Dynamic fatigue was shown to decrease fracture resistance of several materials including dental ceramics that can fracture during function.

In the existent study crowns were cemented using self-adhesive resin cement (RelyX U200; 3M ESPE). This shortened delivery procedure makes it clinically applicable, fast, and appreciated by clinicians.

The results of this study showed that; the fracture resistance of monolithic crowns was differently affected by the type of ceramic material; therefore, the first null hypothesis is rejected. F.explore group verified the highest fracture resistance followed by CD group then e.max group whereas the lowest fracture resistance means value were verified for v.enamic group when the occlusal thickness was identical . However, differences between the four materials fracture resistance can be ascribed to the material structure and the mechanical properties (Table 2). (28) The outcomes of the current study were in accordance with those of Kok et al (29) who examined the mechanical behavior of different implant-supported posterior crowns and found that, the highest initial load to failure was for anatomic contour zirconia crowns followed by e. max CAD then Vita enamic. The results were also in agreement with those of Kim et al (30) who reported that implant supported zirconia posterior crowns had significantly higher fracture resistance comparing to e. max CAD crowns.

Interestingly, the present study displayed that an increase in occlusal thickness from 1 to 1.5mm didn't affect the fracture resistance for all material used except the f.explore group, therefore the second null hypothesis is partly rejected. F.explore group with 1.5 mm occlusal thickness has significantly higher fracture resistance than that with 1.0 mm occlusal thickness, this comes in consistence with Nakamura N et al(31) who found that the monolithic zirconia crowns strength was dependent on the occlusal thickness. Moreover, in the load-to-failure test, some of the f.explore crowns with occlusal thickness of 1.5 mm were not fractured even at 10 kN. This result is consistent with a previous report wherein Beuer et al. (32) demonstrated that 11 out of 12 monolithic zirconia crowns did not fail at 10.5 kN. This could be attributed to the special formulation of f. explore zirconia blanks which consist of 5 layers having different types of zirconia powder within the same blank; 1st layer (4Y-TZP), 2nd, 3rd, and 4th layers are the combination of (4Y-TZP) and (3Y-TZP) so that each layer has different portions of (4Y-TZP) and (3-TZP), therefor different combination of strength and translucency. The 5th layer is (3Y-TZP). (33)

In this study, v.enamic crowns required the least load to fracture (891N). This comes in accordance with the relatively low mechanical properties of this material including low flexural strength (150-160 MPa) and low fracture toughness (1.5 MPa). It can be explained by the hybrid nature of this material as it is composed of interconnected networks of ceramic and polymer, the different behavior of ceramic and polymer during the grinding and polishing processes may result in micro-cracks in the network boundaries, and this is assumed to affect the mechanical properties of the material (34, 35). Moreover, it was shown that, in a hybrid material, failure can be initiated from any weak point of the microstructure, like the polymer in polymer infiltrated ceramic (36). This comes in accordance with the findings of Bilkhair [27] who had a study to compare the fracture strength of monolithic hybrid ceramic crowns with lithium disilicate and feldspathic ceramic crowns. In his study it was shown that the hybrid dental ceramic crowns had lower fracture resistance than that for lithium disilicate crowns. In other study by Sieper et al. (37) that compared the fracture strength of hybrid dental ceramic, lithium disilicate and zirconia-reinforced lithium silicate. It was shown that hybrid ceramic crowns recorded the lowest fracture strength among the tested materials.

Moreover, the outcomes of this study presented that occlusal thickness variations in v.enamic crowns had none significant influence its fracture resistance. The results comes in consistence with the work of Choi S et al. [38] and Chen C et al [39]; the later studied a standardized molar onlays using resin nano-ceramics and they found no significant difference in fracture resistance between 1.0 mm and 1.5 mm occlusal thickness. The authors advocated that difference in mechanical properties of resin surrounding substance and ceramic particles created fracture resistance comparative to thickness variations inside a certain range. Similarly, the stress distribution and fracture manner of v.enamic was dissimilar from glass ceramics owing to elastic polymer complex. Also this result was in accordance with Sallam H. who found that V. enamic crowns on implant abutments used recorded the lowest fracture resistance among the tested ceramic crowns. The author explained this result as; hybrid ceramics are not as strong as ceramics. (40) Vita enamic is based on a network structure made of aluminum oxide-enriched, fine structure feldspar ceramic combined with a proportion of polymer material consisting of UDMA and TEG-DMA. However, its lower modulus of elasticity made it a suitable material for implant supported restorations. (41-43)

On the other hand, according to the current study; it was found that an increase in occlusal thickness of v.enamic to 1.5mm lead to fracture resistance increase to be insignificantly different from e.max at the same occlusal thickness. This result is agreed with Menini M and Conserva E. (43), they have explained that resin-based materials and composites have higher shock absorbing capacity than ceramics. (20, 21) This was in agreement with findings by Kok et al. (29) who found higher fracture values for materials with a lower modulus of elasticity. On the other side, the results were contradicting with those of Rosentritt et al (44) and Weyhrauch et al (45) who tested the fracture resistance of several CAD/CAM fabricated all ceramic implant supported posterior crowns and found that IPS e max CAD crowns registered statistically significant higher fracture resistance values than v.enamic crowns at 1.5mm occlusal thickness.

E.max and CD crowns recorded lower mean values of fracture strength than f.explore crowns with highly significant differences. This might be due to the characteristic lower mechanical properties as flexural strength, lower elastic moduli and lower fracture toughness of the two materials comparing with zirconia. The difference in occlusal thickness in the LD group had no significant effect on fracture resistance. This came in accordance with the work of Sorrentino R et al (46) who found that LD crowns with 1mm occlusal thickness showed higher fracture resistance than 0.5 and 1.5 occlusal thickness; consequently, in agreement with the manufacturer's instructions, the author stated that it is possible to design posterior LD crowns with 1mm occlusal thickness to guarantee the greatest mechanical performances under function. They attributed these results to the fact that, with less bulk of ceramic the stresses are transmitted and absorbed by the resin cement, on the other hand, the stress adsorbing capability of resin cements could not be entirely effective in the presence of bulk thicknesses of glass ceramics and this mechanical drawback could lead to intrinsic micro cracks of lithium disilicate causing clinical failures. This result comes in consistence with Sieper k et al (47) and Sydler B et al (48) who demonstrated that an occlusal thickness of 1.0 mm allowed monolithic lithium disilicate crowns to withstand occlusal forces in the molar areas. On the other hand, this result was contradicted with Choi S et al (50); who's displayed a direct relation between thickness and fracture load.

Results of the present study found that CD group recorded higher statistically difference fracture strength than e.max group this came in harmony with the results Of Preis et al. [51], Schwindling et al. [52] and Jassim Z M[28] et al. the authors attributed this to the following reasons; first: The diffusion of nano- zirconia particles in the CD glassy matrix (10%), looks to improve the fracture resistance comparing with e.max. The integrated zirconia particles boosted the production of a larger amount of minor crystallites (0.5-1μ) rather than the smaller number of large crystallites (1.5 μ) that are existent in the e.max, and this comes in accordance with microstructure appearance which demonstrates that the glass part of CD is present at a higher ratio comparing with conventional LD in spite of LD has higher percentage of crystal part (about 70%) comparing to C D (40-50%) (53). At the same time, the integrated zirconia will undergo phase transformation and increase the fracture toughness by preventing crack propagation [54]. This explains the outcomes of a SEM of this study which displayed distinct crescent arrest lines nearby the beginning of failure were presented in CD crowns, which makes the supposition that micro cracks may have an effect on the fracture strength of CD comparing with LD crowns which display leading hackles from the beginning of failure toward the die (55). Second: The lower modulus of elasticity of CD (70 GPa) comparing with e.max (95 GPa), which recommends that stress collected in the e.max crowns is more than that collected in CD crowns. However, the above outcome differs from the results of Sieper et al. [47] Gungor and Nemli [56] whom experienced the fracture resistance of crowns made-up from LD, zirconia reinforced lithium silicate and other ceramic materials and found that the fracture resistance of LD was higher than zirconia-reinforced lithium silicate crowns. Such disagreement may be due to the difference in the type of zirconia-reinforced lithium silicate material used as they used Vita Suprinity. This was contradicted with Choi S. et al (50) whereas the CD group exhibited a significant lower value comparing with e.max and alterations in the restoration thickness significantly affect the CD fracture resistance.

The mean fracture resistance of all tested ceramic groups in this study went from 811 N to 4663 N. This advocated that all the monolithic crowns used were acceptable for clinical usage because they can experimentally resist the average (700 N) or the maximum physiological forces (1000 N) exercised on posterior teeth.(57,58)

The v. enamic. and LD crowns in the existing study displayed that the fractures started in the load contact area, protracted to the cervical area, and lead to complete crown separation. In the v. enamic. sample, the spreading of minor crack was stopped within the deep occlusal surface, signifying that possible crack limitation due to respectable damage tolerance. On the contrary, the CD and f.explore crowns presented dissimilar forms of fractures, with deeply located crack starts near the interfaces between the crown and cement layer. The compression curls and twisted hackle results with bifurcations advocated probable crack propagation across the occlusal surface. As all the tried crowns were destructed catastrophically.

CONCLUSIONS

Within the restrictions of this study, the subsequent conclusions could be withdrawn:

1- Occlusal thickness significantly affects the fracture resistance of f.explore zirconia groups, while it has non-significant effect for all the other groups.

2- F.explore group with 1.5 mm occlusal thickness has significantly higher fracture resistance than that with 1.0 mm occlusal thickness, and both have significantly higher fracture resistance than all the other groups.

3- Celtra duo groups have significantly higher fracture resistance than IPS E-max CAD and Vita Enamic groups.

4- IPS E-max CAD groups have non-significantly higher fracture resistance than Vita Enamic group with 1.5 mm occlusal thickness.

5- Vita Enamic group with 1.0 mm occlusal thickness has significantly lower fracture resistance than all the other groups.

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