ORIGINAL RESEARCH

To compare the fracture resistance of cobalt chromium four unit fixed dental prosthesis manufactured by CAD/CAM milling, 3D printed resin pattern casting and direct metal laser sintering

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ABSTRACT
Introduction: The purpose of this study is to evaluate and compare the fracture resistance of cobalt chromium four unit fixed dental prosthesis framework obtained by CAD/CAM milling, 3D printed resin pattern casting and direct metal laser sintering (DMLS). The null hypothesis was that no differences would be found in fracture resistance of cobalt chromium four unit fixed dental prosthesis framework obtained by CAD/ CAM milling, 3D printed resin pattern casting and direct metal laser sintering (DMLS). **Material and methods**: A total number 30 patients were selected and they are further divided into three groups as follows. 10 copings of CAD/CAM Milling. 10 copings of 3D Printed Resin Pattern Casting. 10 copings of Direct Metal Laser Sintering (DMLS). **Results:** The results only showed a difference in fracture resistance between the CAD/CAM and the DMLS-group ($p < 0.05$). Despite the fact that there was a lot of variation in the readings of resin pattern casting. Its mean fracture resistance was 6298.5, which was higher than the other two groups (CAD/CAM Milling and DMLS). The average of the fracture load came of resin pattern casting was in comparison with CAD/CAM Milling (6247.2) and higher than that of DMLS (5153.0). **Conclusion:** There is a lot of variations in the values of 3 D Printed Resin Pattern Casting, hence it does not have statistically significant difference in fracture resistance. Hence, the fracture resistance amongst all the groups was exhibited in the following order: a) CAD/CAM Milling b)Direct Metal Laser Sintering (DMLS). The fracture resistance for CAD/CAM Milling was better than all the other groups.
Keywords: fracture resistance, cobalt chromium four unit fixed dental prosthesis, cad/cam milling, 3d printed resin pattern casting and direct metal laser sintering

INTRODUCTION

The rampant consumption of soft diet worldwide has multiplied the occurrence of dental caries leading to eventual loss of teeth. Such a scenario warrants rehabilitation. Fixed prosthodontic treatment involves the replacement and restoration of teeth by artificial substitutes that are not readily removable from the mouth. Its focus is to restore function, aesthetics, and comfort. Fixed prosthodontics can offer exceptional satisfaction for both patient and dentist. Though implant therapy has seen rapid strides in fixed prosthodontics, it still remains unavailable to the general population due to high prices. Fixed dental prosthesis form the bulk of prosthesis inserted universally. Porcelain fused to metal (PFM) is thus the prosthesis of choice in the so-called “esthetic” zone and “posterior masticatory zones” and some even prefer fixed metal crown (FMC) in the posterior region of mouth. [1]

Even with the rise in popularity of all-ceramic dental restorations, which give the impression of superior aesthetics, biocompatibility and the increase in demand of metal free materials, all-ceramic systems have poor mechanical properties when compared with metal ceramic fixed dental prosthesis. [2] Therefore, metal-ceramics are still among the most commonly used material combinations for the prosthetic restoration of teeth.

[3] They are generally used when large number of teeth are to be replaced, due to their clinical longevity. [4]

In past Nobel alloys were used for the fabrication of fixed dental prosthesis. But the fluctuating prices of gold and other precious metals have encouraged the use of less costly alternative alloys. [5] The transition from high noble alloys to the much more economical base metal alloys, allow the latter to remain a formidable competitor. [6,7]

The alloys can be cast using the lost wax technique or milled by the computer-aided designing and computer-aided manufacturing (CAD-CAM) technique. Casting has been the most common method to fabricate dental alloy for many decades, but errors accumulated in the series of laboratory steps are inevitable. [3] In recent decades, digitalized technologies have been employed for the production of metallic structures, mainly in prosthetic dentistry. These technologies can be classified as based on subtractive manufacturing, such as the milling of pre-manufactured materials assisted by computer-aided design/computer-aided manufacturing (CAD/CAM) systems, or on additive manufacturing, such as the recently developed selective laser melting (SLM) technique [4] or direct metal laser sintering (DMLS) or additive prototyping technique (3D printing) is being used to design and then print a wax pattern or a resin pattern and then traditional casting is done.[5]

CAD/CAM technique was introduced in dentistry more than 20 years ago. [8] Year 1985 is the key to the introduction of CAD / CAM technology in dentistry. It was introduced after lost wax technique. [9]

CAD/CAM hard milling is one of the subtractive processes for producing Co-Cr restorations. The formation of casting-induced flaws and porosities are minimized by using Co-Cr alloy blanks manufactured under standardized industrial conditions. [10]

In recent years additive manufacturing technology such as selective laser melting (SLM) or direct metal laser sintering, has also become a viable option in place of conventional lost wax technique. All these methods are presently in commercial use for the fabrication of dental restorations including fixed prosthodontics. [1]

The powdered form of Co-Cr alloy can be used in DMLS technique. As described earlier, even this powered form of Co-Cr have high strength, are heat-resistant and non-magnetic, and have favourable resistance to wear, corrosion, and tarnish. [2]

The DLMS method produces a three-dimensional prosthesis in a layer by-layer fashion by selectively fusing together metal powder particles using a computer-controlled laser using a high-energy carbon dioxide laser beam. [11] As one layer is finished, the table where the finished part of the reconstruction is placed is lowered so that a new layer of green granules can be added over the sintered granules. Subsequently, the laser welds the new granules to the old ones, building up the reconstruction grain by grain until finished. The technique allows for many prosthetic units to be produced at the same time and the powder granules which are not exposed to the laser beam can be re-used. Both are the important advantages of the technology. [12] [13]

Observing the DMLS specimens, it can be noticed that the different layers are no more visible. This indicates that the particles are strongly joined together not only within each layer, but also between different layers. [14]In comparison with traditional casting, the DMLS technique also changes the surface microstructure of the alloy. [15] CAD/CAM milling and DMLS systems can save time and facilitate dental laboratory procedures so that they can be used routinely as alternatives to casting. Besides, laboratory procedures can be standardized by computerized technology. [16]

The selection of metallic powders is a crucial consideration in the preparation of DMLS prostheses for dental applications. Allergy reports have indicated that intraoral exposure to some elements contained in base metal alloys (i.e., nickel and beryllium) are of significant concern to human health. A new Co–Cr alloy, based on SLM technology, was recently introduced for manufacturing a new form of base metal copings. The immediate advantage of this SLM Co–Cr alloy is comparable performance to other base metal alloys, but without an allergenic nickel component. [17]

The term ‘3D printing’, however, is relatively new, and has captured the public imagination. This process is more correctly described as additive manufacturing, and is also referred to as rapid prototyping. Additive Manufacturing is the means of creating an object by adding material to the object layer by layer. The computer directs the 3-D printer to add each new layer as a precise cross-section of the final object. In Rapid Prototyping, designers create models using computer-aided design (CAD) software, and then machines follow that software model to determine how to construct the object. The process of building that object by "printing" its cross-sections layer by layer became known as 3-D printing. [18]

With the use of intraoral optical scanners or laboratory scanners it is possible to develop a precise virtual model of the prepared tooth and the dental arch. In fixed and removable prosthodontics, treatment may be planned and restorations designed in CAD software. This scan data and CAD design may be used to print crown or bridge copings and bridge structures. 3D printing may be harnessed for the fabrication of metal structures, either indirectly by printing in burn-out resins or waxes for a lost-wax process, or directly in metals or metal alloys. The advantage of printing in resin/wax and then using a traditional casting approach is that there is much less post-processing involved than in the direct 3D printing of metals and printing directly in metals requires the use of more costly technologies which have their own very specific health and safety requirements. Casting and casting alloys and facilities are familiar and widely available. [18]

Fixed prosthodontic treatment can range from the restoration of a single tooth to the full mouth rehabilitation. Missing teeth can be replaced with prostheses that will improve patient comfort and masticatory efficiency and maintain the health and integrity of the dental arches. The maximum number of posterior teeth that allows replacing with a fixed dental prosthesis is usually two. In rare circumstances, three can be replaced, but that should be attempted only under ideal conditions. [19]

The shape of an fixed dental prosthesis is not uniform. Its contour has a complex combination of multiple convexities and concavities, depending on the geometry of the teeth and their alignment. In particular, the connector area has a narrow constriction for biologic and esthetic reasons, and these sites in 3- and 4-unit fixed dental prosthesis represent stress concentrations relative to the average stress levels within other areas of the prosthesis. [20]

Fracture toughness and fracture resistance of the material is important if the restoration is expected to serve the patient for a long period of time. Fracture toughness value is important for more accurate prediction for long-term clinical performance of restoration.[1]

There are few studies about comparison of fracture resistance of Co-Cr alloys manufactured by different methods. Four unit fixed dental prosthesis is a long span bridge and for long survival span, its fracture resistance is important. Most studies involving fracture resistance of fixed dental prosthesis have focused on those with only three units, with very few targeting on four unit fixed dental prosthesis.

Therefore, the aim of this in vitro study is to evaluate and compare the fracture resistance of cobalt chromium four unit fixed dental prosthesis framework obtained by CAD/CAM milling ,3D printed resin pattern casting and direct metal laser sintering (DMLS). The null hypothesis was that no differences would be found in fracture resistance of cobalt chromium four unit fixed dental prosthesis framework obtained by CAD/ CAM milling, 3D printed resin pattern casting and direct metal laser sintering (DMLS).

METHODOLOGY

A total number 30 patients were selected and they are further divided into three groups as follows.
10 copings of CAD/CAM Milling.
10 copings of 3D Printed Resin Pattern Casting.
10 copings of Direct Metal Laser Sintering (DMLS).

INCLUSION CRITERIA

1. Four unit fixed partial denture
2. Cobalt chromium alloy.
3. Defect free copings.
4. Defect free Cobalt Chromium model.

EXCLUSION CRITERIA

1. Nickel, titanium alloy.
2. Coping with voids and nodules.

METHODOLOGY FOR THE STUDY

The methodology for the study undertaken has been divided under the following titles for the ease of understanding:

1. Fabrication of the custom-made Cobalt Chromium master die.
2. Fabrication of Group A.
3. Fabrication of Group B.
4. Fabrication of Group C.
5. Measurement of the Fracture Resistance.
6. Statistical analysis.

Fabrication of the custom-made Cobalt-Chromium master die

A plastic first premolar and second molar were prepared with a circumferential rounded chamfer margin in a typhodont. The finish line was 0.5 mm in width. During occlusal preparation, functional cusp was reduced by 1.5 mm and non-functional cusp was reduced by 1mm approximately. The functional cusp bevel was given. The crown preparation was done with 6-degree total axial wall taper. (fig.1)

The teeth prepared on the typhodont were scanned using Shining 3 D desktop scanner. (fig.2)

Custom-made Co-Cr master die was fabricated simulating the shape and dimensions of tooth preparation using a CAD/CAM Milling Machine.

The metal master die was fabricated and it was used as the basic template model.

Fabrication of Group A.

For group A, 10 Co Cr copings are to be fabricated with the help of CAD/CAM technology.

The CAD/CAM unit used is Yenadent DC 40. (fig.4)

First the Co-Cr model was scanned using the shining 3 D desktop scanner.

Then the scan was exported as an open file in the STL format to the CAD software.

The data obtained after scanning was converted into CAD data using which the fixed dental prosthesis of the respective groups were designed by the Exocad software.

In Exocad software, we have selected lower first premolar, second premolar, first molar and second molar. Then for each teeth restoration type was selected. For first premolar and second molar, restoration type selected was anatomic crown and for second premolar and first molar restoration type selected was anatomic pontic. Then the material selection was done. We have selected cobalt chromium.

After saving, we have chosen the scan of the master model. Then we have clicked on the design FPD icon to start the CAD application. The software has automatically started a wizard that guided us through all the work steps. It has started with the margin line detection. We could navigate the model in 3D space using mouse.

First the margin line was made starting from the buccal side by clicking on the points. Then after the margin lines were drawn. The fixed partial denture insertion directions were verified and the teeth were copied from the library.

Then the tooth placements were checked and the designing of the connector was done. The width, height and cross-sectional area of the connector and the shape and the position of the connector was fixed.

Like this a virtual coping was constructed on the screen.

Thickness of the coping was designed to be 1.5 mm thick for the functional cusp and 1mm for the non-functional cusp and 0.5 mm at the margin for optimal strength.

The connector width, length and cross-sectional area was standardized. The width was kept 4mm and length 4mm and the cross-sectional area was kept 16mm square.

Then again, the transformation to STL format was carried out and sent to CAM software for manufacturing of the CAD-CAM Co-Cr copings.

Like this, 10 Co-Cr copings were fabricated with CAD/CAM milling technology.

Fabrication of Group B.

For group B, 10 Co-Cr copings are to be fabricated with the help of 3D Printed Resin Pattern Casting with photon 3D printer and Unident centrifugal casting unit. (fig.11)

First the Co-Cr master die was scanned using the shining 3D desktop scanner. Then the scan was exported as an open file in the STL format to the CAD software.

The data obtained after scanning was converted into CAD data using which the copings of the respective groups were designed by the Exocad software.

In Exocad software, we have selected lower first premolar, second premolar, first molar and second molar. Then for each teeth restoration type was selected. For first premolar and second molar, restoration type selected was anatomic crown and for second premolar and first molar restoration type selected was anatomic pontic. Then the material selection was done. We have selected wax on the Exocad software and finished the restoration definition.

After saving, we have chosen the scan of the master model. Then we have clicked on the design FDP icon to start the CAD application. The software has automatically started a wizard that guided us through all the work steps. It has started with the margin line detection. We could navigate the model in 3D space using mouse. First the margin line was made starting from the buccal side by clicking on the points. Then after the margin lines were drawn. The fixed partial denture insertion directions were verified and the teeth were copied from the library.

Then the tooth placements were checked and the designing of the connector was done. The width, height and cross-sectional area of the connector and the shape and the position of the connector was fixed.

Like this a virtual coping was constructed on the screen.

Thickness of the coping was designed to be 1.5 mm thick for the functional cusp and 1mm for the non-functional cusp and 0.5 mm at the margin for optimal strength.

The connector width, length and cross-sectional area was standardized. The width was kept 4mm and length 4mm and the cross-sectional area was kept 16mm square.

This creates a 3-dimensional CAD model, creating a file. Then again, the transformation to STL format was carried out and sent to 3D printer software for 3D printing.

With the help of this data 3D printer printed the fixed dental prosthesis. The prosthesis was printed in burn-out resin material.

Then the copings were fabricated in Co-Cr material with the help of traditional centrifugal casting approach.

Like this, 10 Co-Cr copings were fabricated with 3 D Printed Resin Pattern Casting technology.

Fabrication of Group C.

For group C, 10 Co Cr copings are to be fabricated with the help of direct metal laser sintering technology with the help of EOS M 100 sintering unit. (fig.22,23)

First the Co-Cr master die was scanned using the shining 3D desktop scanner. Then the scan was exported as an open file in the STL format to the CAD software.

The data obtained after scanning was converted into CAD data using which the copings of the respective groups were designed by the Exocad software.

In Exocad software we have selected lower first premolar, second premolar, first molar and second molar. Then for each teeth restoration type was selected. For first premolar and second molar, restoration type selected was anatomic crown and for second premolar and first molar restoration type selected was anatomic pontic. Then the material selection was done. We have selected cobalt chromium and finished the restoration definition.

After saving, we have chosen the scan of the master model and we have clicked on the design FDP icon to start the CAD application. The software has automatically started a wizard that guided us through all the work steps. It has started with the margin line detection. We could navigate the model in 3D space using mouse. First the margin line was made starting from the buccal side by clicking on the points and the margin lines were drawn.

Then the fixed partial denture insertion directions were verified and the teeth were copied from the library. Then the tooth placements were checked.

Then the designing of the connector was done. The width, height and cross-sectional area of the connector and the shape and the position of the connector was fixed. Like this a virtual coping was constructed on the screen.

Thickness of the coping was designed to be 1.5 mm thick for the functional cusp and 1mm for the non-functional cusp and 0.5 mm at the margin for optimal strength.

The connector width, length and cross-sectional area was standardized. The width was kept 4mm and length 4mm and the cross-sectional area was kept 16mm square.

This creates a 3-dimensional CAD model, creating a file. Then again, the transformation to STL format was carried out and sent to the direct metal laser sintering unit.

With the help of this data DMLS machine fabricated the copings in layer-by-layer fashion.

Like this, 10 Co-Cr copings were fabricated with direct metal laser sintering (DMLS) technology. (fig.26)

Measurement of the Fracture Resistance.

All the 30 samples of Group A, Group B and Group C were evaluated for fracture resistance.

Static load-bearing tests will be conducted with a universal testing machine.

The Fixed dental prosthesis which are seated on Co-Cr master die were loaded in compression centrally on connector between the two pontic with 8-mm-diameter stainless steel ball until fracture in a universal testing machine with speed of 1mm/minutes. Loading was continued till fracture of the fixed dental prosthesis occurred and failure loads were recorded. Like this the fracture resistance was measured.

RESULTS

Evaluating the Fracture Resistance for Group A:

For Group A, the basic data for evaluating the fracture resistance and the mean for Group A was calculated.

The mean fracture resistance obtained for Group A was 6247.2 with a standard deviation of 929.757 and a standard error of 280.332. The minimum and maximum values were 4322 and 7493 respectively.

Evaluating the Fracture Resistance for Group B:

For Group B, the basic data for evaluating the fracture resistance and the mean for Group B was calculated.

The mean fracture resistance obtained for Group B was 6298.5 with a standard deviation of 1691.844 and a standard error of 510.110. The minimum and maximum values were 3778 and 8201 respectively.

Evaluating the Fracture Resistance for Group C:

For Group C, the basic data for evaluating the fracture resistance and the mean for Group C was calculated.

The mean fracture resistance obtained for Group C was 5153 with a standard deviation of 432.193 and a standard error of 130.311. The minimum and maximum values were 4543 and 5860 respectively.

Comparing the Fracture Resistance exhibited by Group A, Group B and Group C:

Comparison of the mean values of all the three groups was done where the means of Group A, Group B and Group C were 6247, 6298 and 5153 respectively.

Test of Normality (Shapiro Wilk Test) was done. According to the test, the observed data was distributed normally. Therefore, ANOVA test was used to compare the fracture resistance.

ANOVA test was applied to compare the mean values between the three groups. According to the ANOVA test it was confirmed that the mean Fracture Resistance for Group A, Group B and Group C differed significantly with the F value being 3.530 and the significance being .042

There is lot of variation in the values for 3D Printed Resin Casting and hence it does not have statistically significant difference in fracture resistance. Hence, the result came was CAD/CAM has more resistance as compared to DMLS.

The highest fracture resistance was exhibited by Group A (mean fracture resistance = 6247.2), followed by Group C (mean fracture resistance = 5153.0) and Group B is not statistically significant (mean fracture resistance = 6298.5). The mean fracture resistance for Group A, Group B and Group C were within the clinically acceptable range.

Post Hoc test was applied to do multiple comparisons between Group A, Group B and Group C. Only there is a statistically significant difference between CAD/CAM and DMLS, CAD/CAM has more fracture resistance than DMLS technique.

The null hypothesis was rejected as there was a statistically significant difference between the fracture resistance of the two study groups.

Table 1- Test of Normality (Shapiro Wilk Test)

Test of Normality (Shapiro Wilk Test)		Statistic	df	Sig.
Fracture Resistance	CAD/CAM Milling	.951	10	.052
	3D Printed Resin Casting	.867	10	.071
	DMLS	.857	10	.654

*The observed data is distributed normally. Therefore, ANOVA test is used to compare the fracture resistance.

Table 2- Descriptive Statistics for Fracture Resistance (N)

Type of Process	N	Mean	Std. Deviation	Std. Error	95% Confidence Interval for Mean		Minimum	Maximum
					Lower Bound	Upper Bound		
CAD/ CAM Milling	10	5153.00	432.193	130.311	4862.65	5443.35	4322	7493
3D Printed Resin Casting	10	6298.45	1691.844	510.110	5161.86	7435.05	3778	8201
DMLS	10	6247.18	929.757	280.332	5622.56	6871.80	4543	5860

Table 3- ANOVA test to compare the mean fracture resistance between the different method

ANOVA Test	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	9210405.818	2	4605202.909	3.530	.042
Within Groups	39135770.364	30	1304525.679		
Total	48346176.182	32			

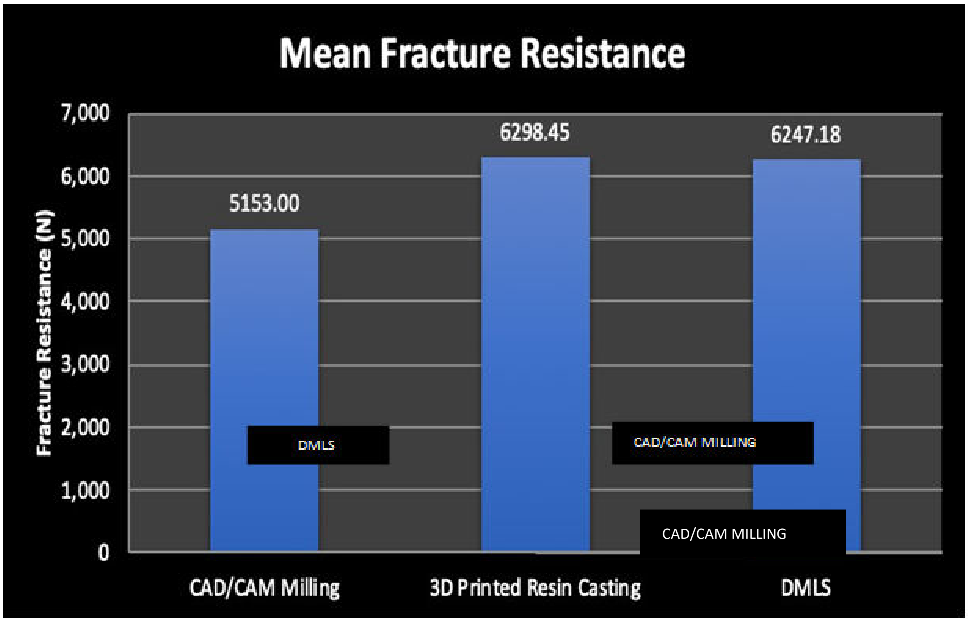
*The mean fracture resistance difference is statistically significant at 5% level of significance (P < 0.05)

Table 4 Multiple Comparison (Post Hoc test)

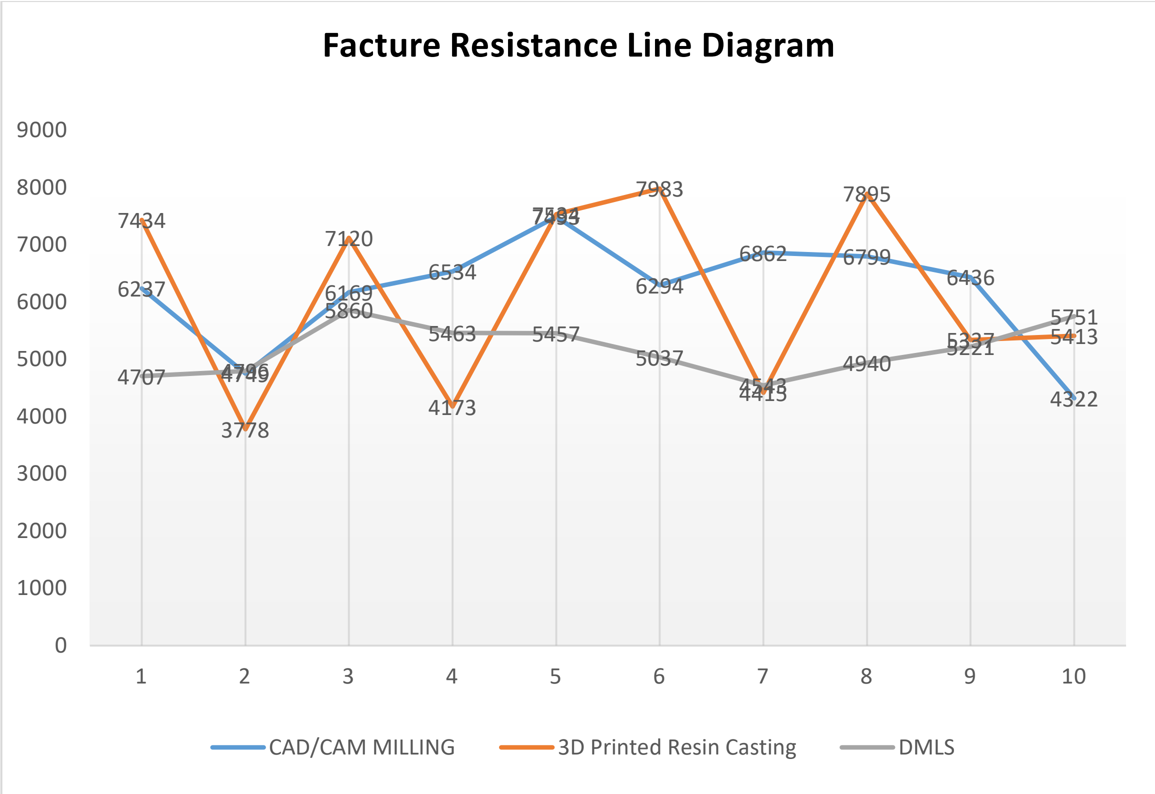
Group (I)	Group (J)	Mean Difference (I-J)	Std. Error	Sig.	% Confidence Interval		Result
					Lower Bound	Upper Bound	
DMLS	3D Printed Resin Casting	-1145.455	526.492	.139	-2604.98	314.07	Non Significant
	CAD/ CAM Milling	-1094.182*	309.139	.009	-1925.16	-263.21	Significant
3D Printed Resin Casting	CAD/ CAM Milling	51.273	582.064	1.000	-1496.16	1598.71	Not Significant

*Only there is a significant difference between DMLS and CAD/CAM, and CAD/CAM has more fracture resistance than DMLS technique

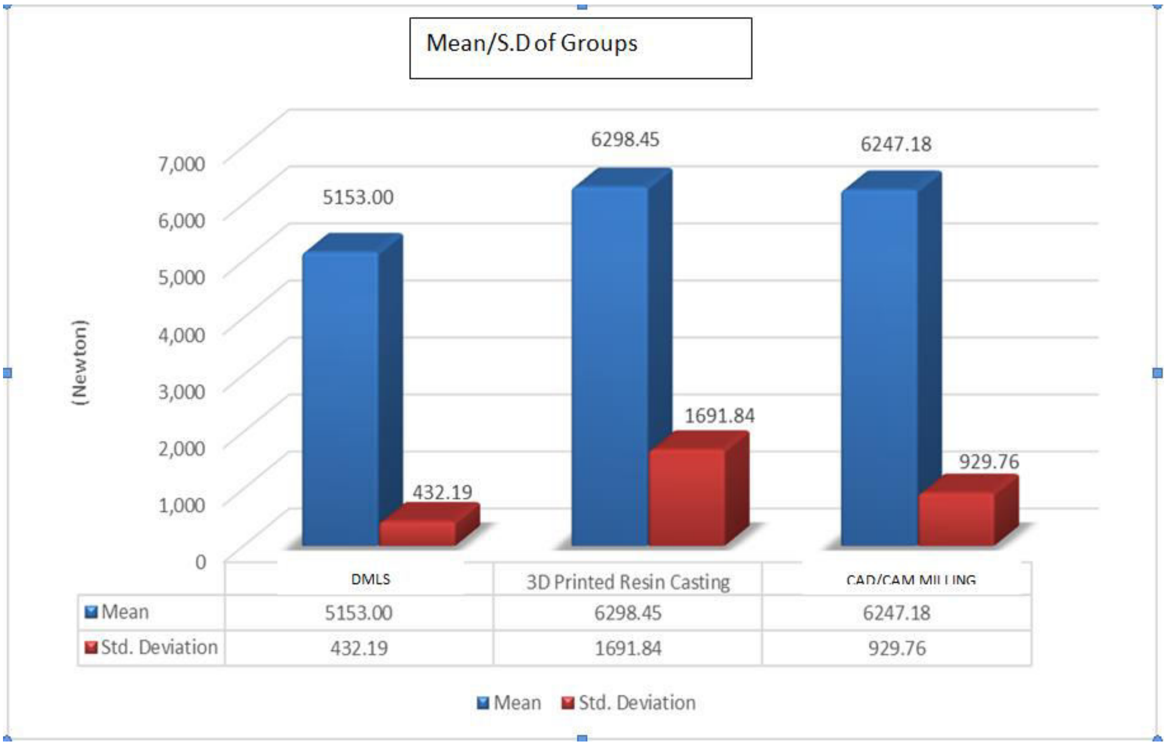
GRAPHS



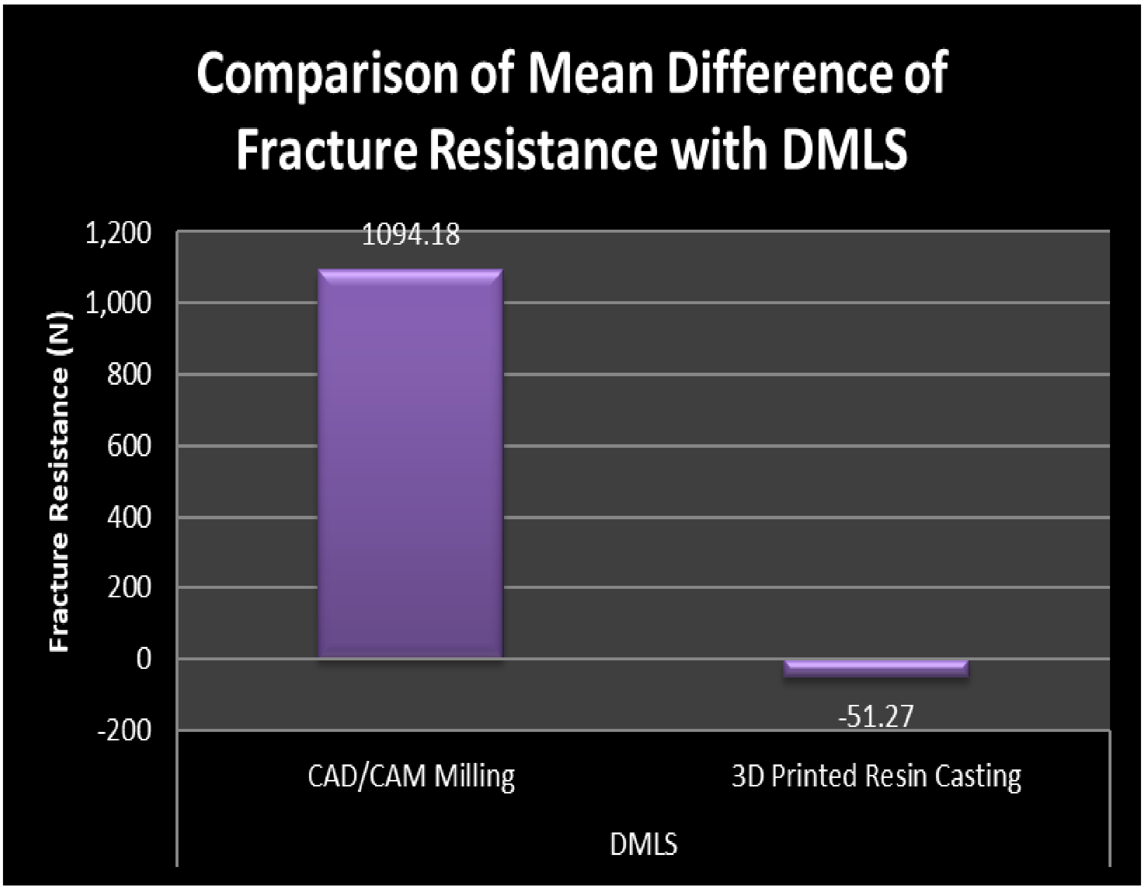
Graph 1: Descriptive Statistics for Fracture Resistance (N)



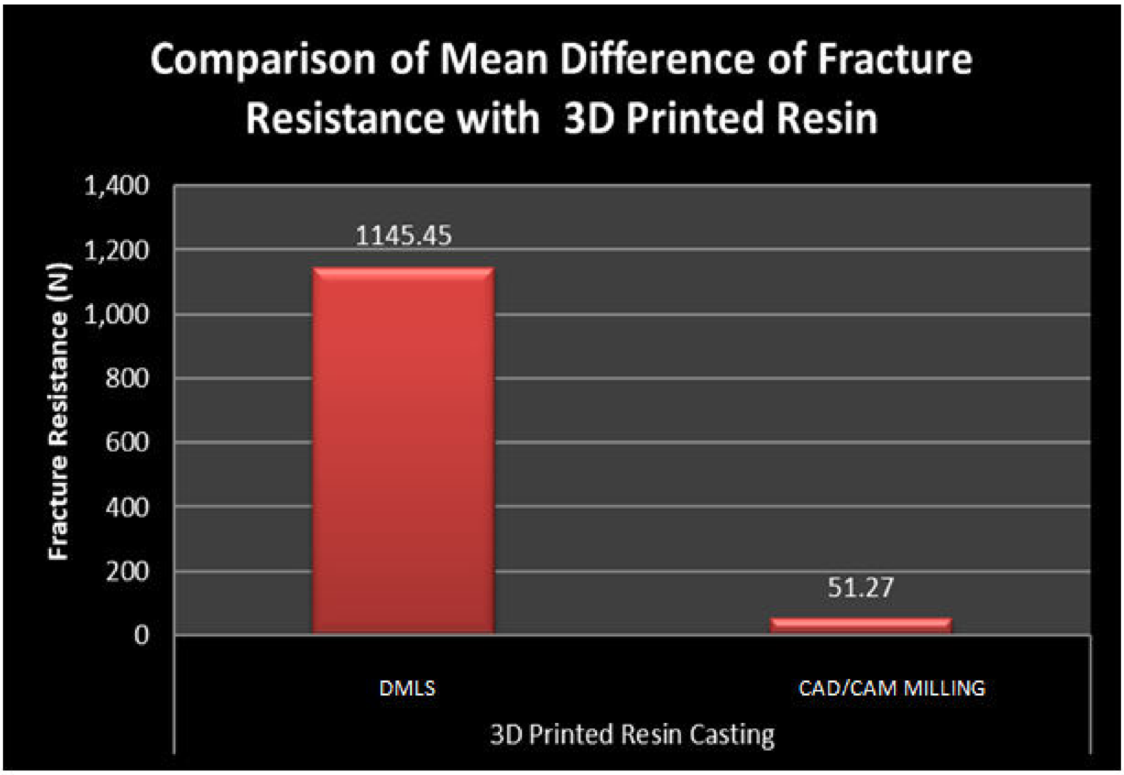
Graph 2: Facture Resistance Line Diagram



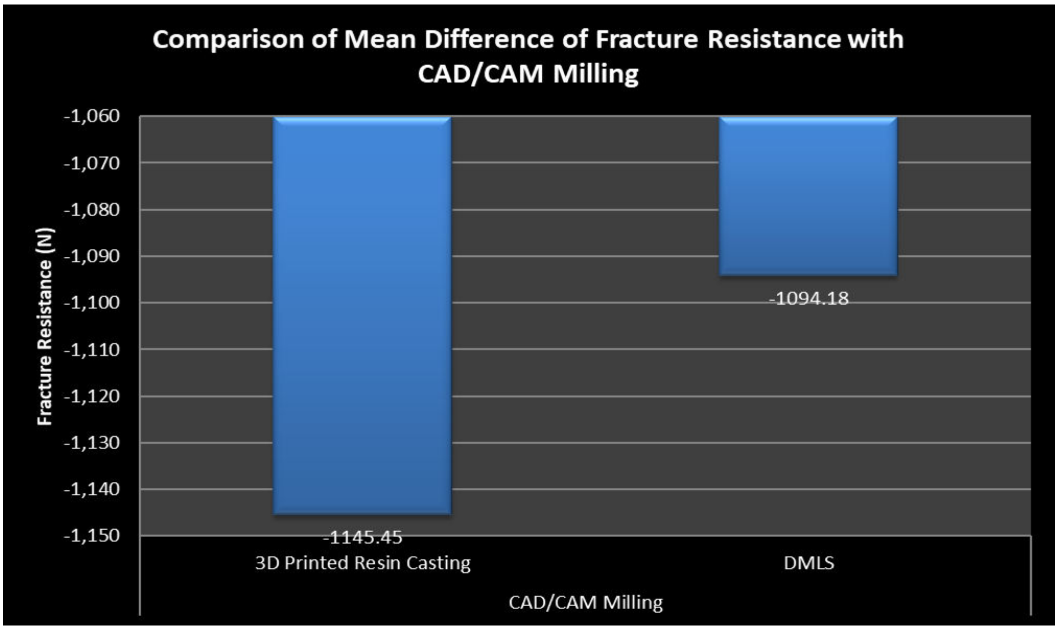
Graph 3: Mean/S.D of Groups



Graph 4: Comparison of Mean Difference of Fracture Resistance with CAD/CAM Milling



Graph 5: Comparison of Mean Difference of Fracture Resistance with 3D Printed Resin Casting



Graph 6: Comparison of Mean Difference of Fracture Resistance with DMLS

Figure 1: Crown Preparation on Typhodont.

Figure 2: Desktop Scanner - Scanning Typhodont

Figure 3: Co-Cr Master Model

Figure 4: CAD / CAM System

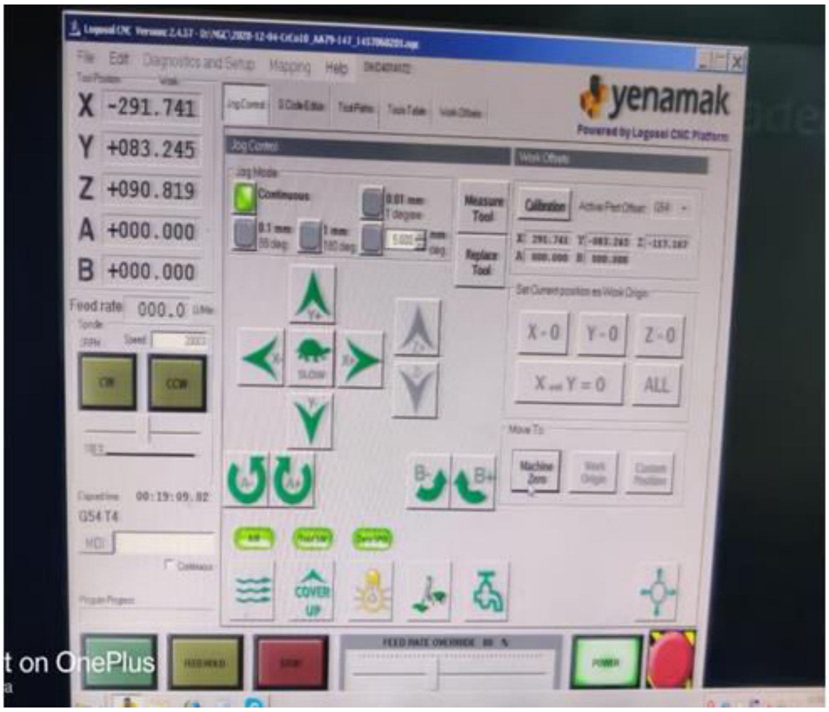


Figure 5: Programming of the CAD/CAM System - part 1 of 2

Figure 6: Programming of the CAD/CAM System - part 2 of 2

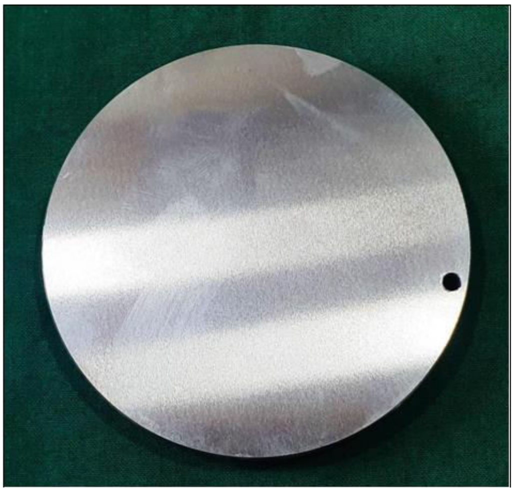


Figure 7: CAD/CAM Blank

Figure 8: Co-Cr Blank in Milling Uni

Figure 9: Milled Samples on Co-Cr Blank

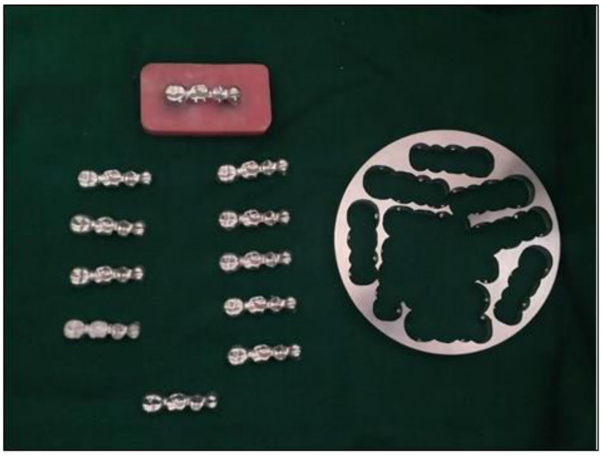


Figure 10: Group A - 11 CAD/CAM Samples

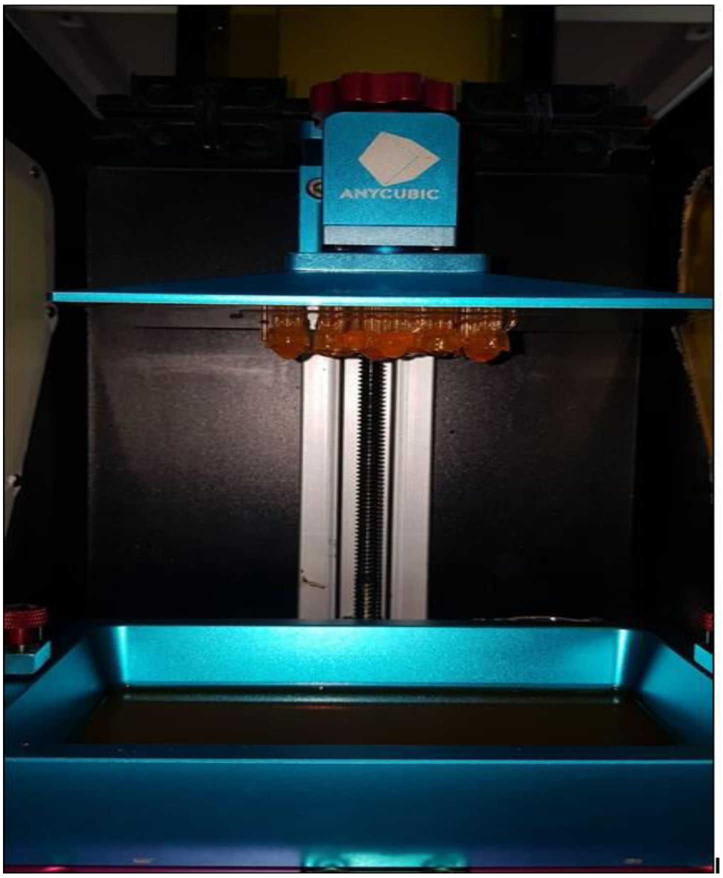


Figure 11: 3D Printer Un

Figure 12: Castable Resin

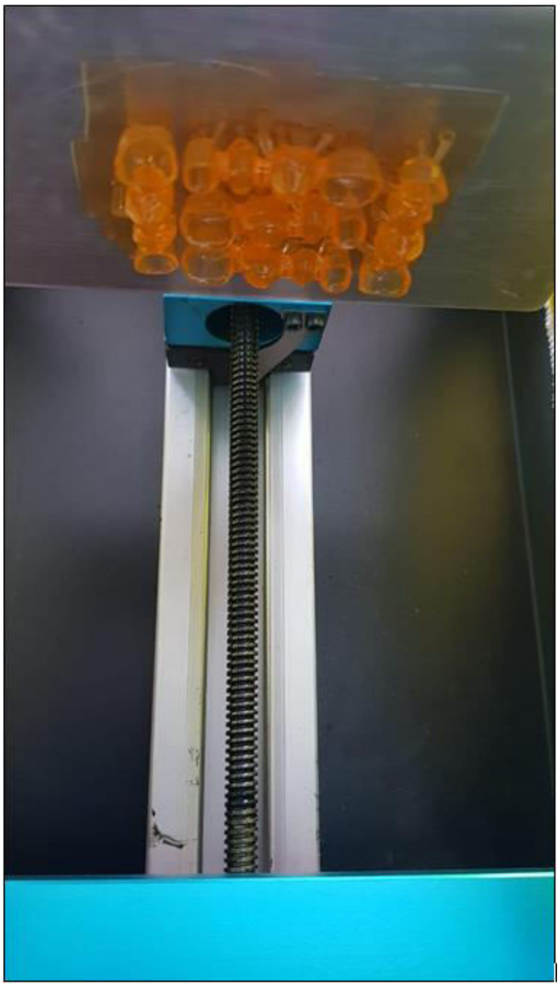


Figure 13: Fabricated Resin Pattern on 3D Printer - Full View

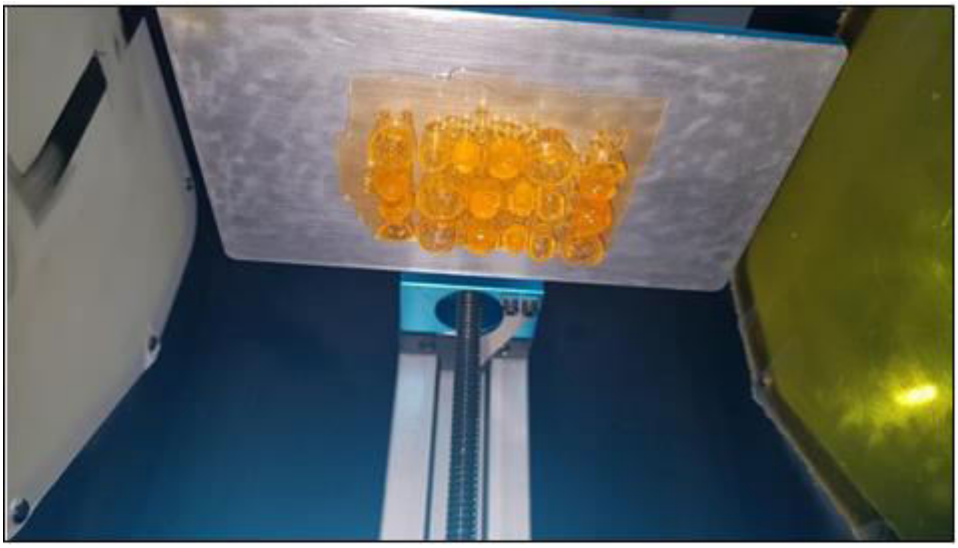


Figure 14: Fabricated Resin Pattern on 3D Printer - Closer View

Figure 15: Resin Pattern on Master Model - Occlusal View

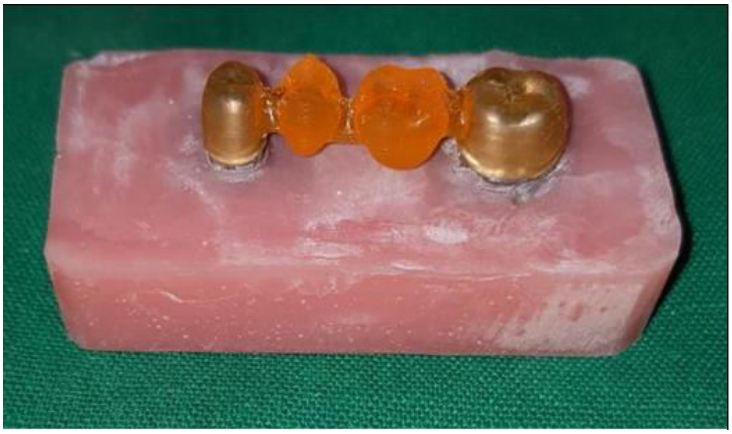


Figure 16: Resin Pattern on Master Model - Axial View

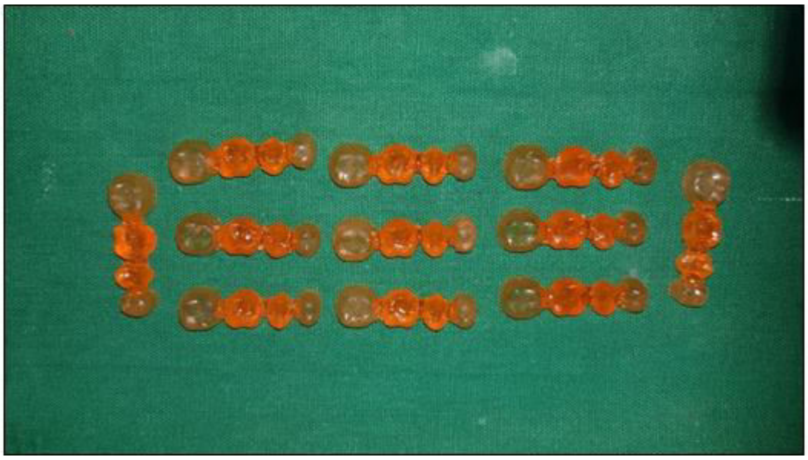


Figure 17: Group B - 11 3D Printed Resin Pattern Samples

Figure 18: Centrifugal Casting Unit

Figure 19: Casting Alloy

Figure 20: Phosphate Bonded Investment Material

Figure 21: Group B - 11 3D Printed Resin Pattern Casting Samples

Figure 22: DMLS Unit

Figure 23: DMLS Unit Closer View

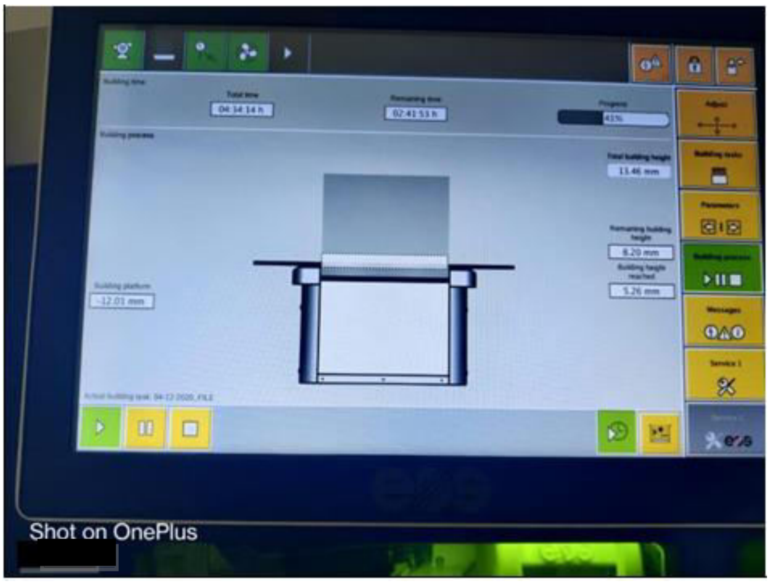


Figure 24: Programming of the DMLS Unit

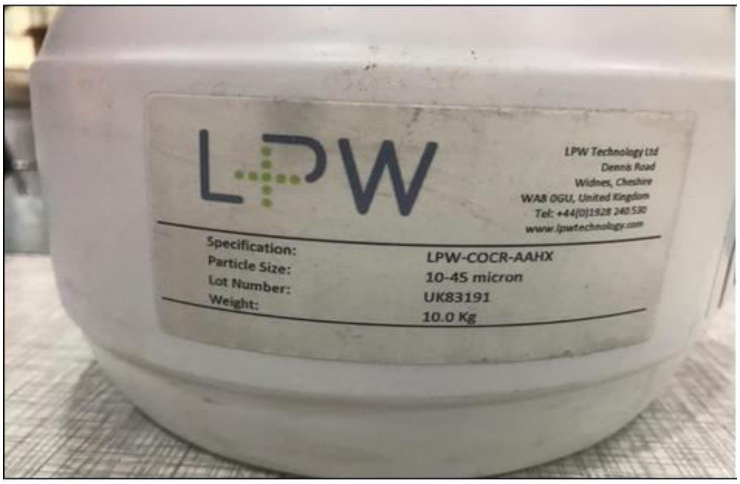


Figure 25: Co-Cr DMLS Powder

Figure 26: Group C - 11 DMLS Samples

Figure 27: Universal Testing Machine

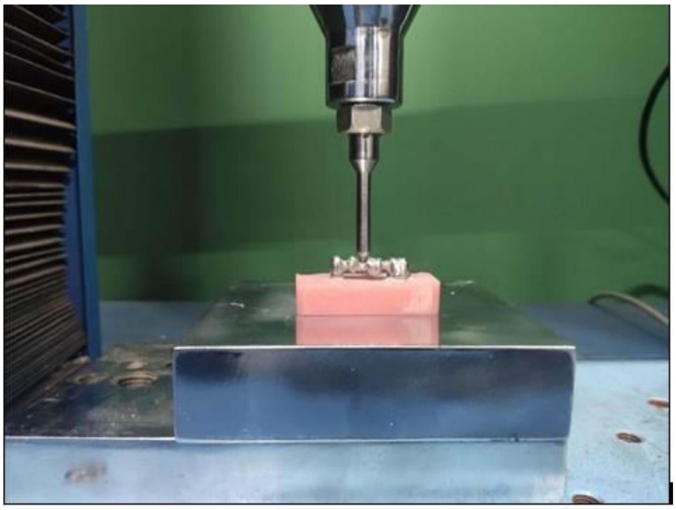


Figure 28: Load Applied on Centre of Connector Area

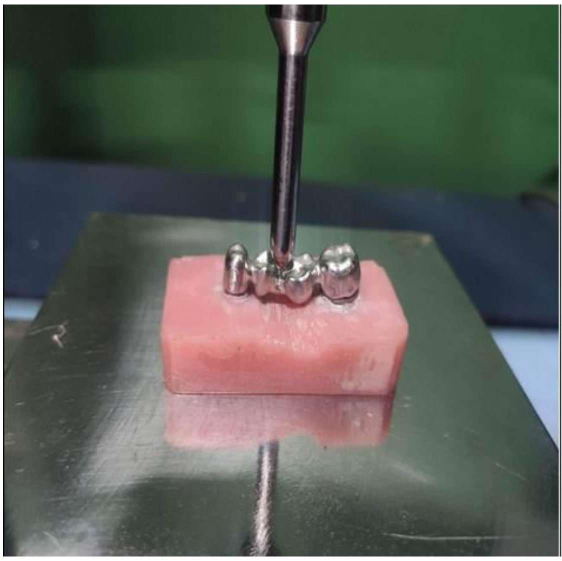


Figure 29: Flexion of the Connector Seen Before Facture

Figure 30: Load vs Displacement Graph on UTM

Figure 31: Group A - CAD/CAM Milled Fracture Samples

Figure 32: Group B - Resin Pattern Casting Fractured Samples with Comparative Higher Resistance

Figure 33: Group B - Resin Pattern Casting Fracture Samples with Comparative Lower Resistance

Figure 34: Group C - DMLS Fractured Sample

DISCUSSION

Long term success of the Fixed Partial Denture, depends on the fracture resistance of fixed partial denture. The causes that compel patients to seek any dental restorations including fixed dental prosthesis, inlays, veneers and crowns could be divided into the following: (1) dental disease including caries or periodontal causes; (2) trauma such as accidents and (3) aesthetics to improve the appearance of their smile. When we consider these causes then the main aim while planning a fixed dental restoration i.e., a crown or a bridge is to replace missing teeth, restore function and aesthetics without compromising the oral health. [21]

The present study investigated the fracture resistance of Co-Cr fixed dental prosthesis produced using different manufacturing techniques: 3 D printed resin pattern casting, CAD/CAM milling and Direct metal laser sintering techniques. The fracture resistance of each sample of each group was calculated. Then the mean of each group was calculated. Then the test results were statistically analyzed with ANOVA test. The results only showed a difference in fracture resistance between the CAD/CAM and the DMLS-group ($p < 0.05$). Despite the fact that there was a lot of variation in the readings of resin pattern casting. Its mean fracture resistance was 6298.5, which was higher than the other two groups (CAD/CAM Milling and DMLS). The average of the fracture load came of resin pattern casting was in comparison with CAD/CAM Milling (6247.2) and higher than that of DMLS (5153.0). The lowest fracture load of DMLS was 4543 and of resin pattern casting was 3778 and of CAD/CAM Milling was 4322. When we compare from the lowest fracture load, the resin pattern casting has got the lowest fracture resistance which was followed by CAD/CAM Milling and then DMLS. Whereas the highest fracture load of DMLS was 5860 and of resin pattern casting was 8201 and of CAD/CAM milling was 7493. So here, the resin pattern casting has got the highest fracture resistance followed by CAD/CAM milling and then DMLS.

The fracture load of 10 CAD/CAM milled samples are 6237, 4749, 6169, 6534, 7493,6294,6862,6799 and 6436 Newton. The fracture load of 10 Resin pattern casting samples is 7434, 3778, 7120, 4173, 7534, 7983, 4415, 7895 and 5337 Newton. The fracture load of 10 DMLS samples are 4707, 4707, 4796, 5860, 5463, 5457, 5037, 4543, 4940 and 5221 Newton. As you can observe there is very little variation in CAD/CAM samples then in DMLS samples and then in resin pattern casting samples.

Fracture resistance is the nominal property of a material when this is resisting the development of a fracture.

Casting metals by the lost wax process have been recognized in the industry and the arts for many years. No record exists to indicate exactly when and where this type of casting procedure was first developed. All that is known is that somewhere along the line — it may have been in ancient Egypt or in ancient china – someone conceived of the idea of making a wax replica of an item to be cast, surrounding this replica with an investment material, letting the investment harden, then melting and burning out the wax, thus producing a mold having highly intricate and accurate cavity. The next step was melting the metal and pouring it into cavity. [22]

It has long been the dentist's desire, however, to be able to produce a restoration in the laboratory and place it in the patient's mouth in a relatively short time. In the literature, credit is given to Dr. Swasey

(1890), who introduced a technique where a solid gold inlay could be prepared. In his technique, the gold foil was first adapted to the shape and contour of the tooth, then removed, invested, and filled with 20-karat gold. Wax was used for making gold inlays for the first time by Martin (1891). According to Martin's technique, there was no need to use a foil liner. He filled the cavity with wax, removed it after hardening, invested it, burned it out, and then poured the molten gold into the mold formed in the investment. [22]

A few years later, Dr. Philbrook (1896) introduced a pressure-casting method of producing gold inlays. His technique was essentially the same as that used today. In his technique, the wax pattern was formed directly in the mouth, mounted on a sprue pin, and invested in a metal ring, with plaster of Paris and silex used as an investment medium. The metal ring was placed in an oven for wax elimination. Finally, the alloy was melted in a crucible formed in the investment, and cast by means of air pressure. At the time, the profession failed to realize the significance of Philbrook's contribution. It is Philbrook whose name should be associated with the method of casting dental castings under pressure. About 10 years later, Dr. Taggart (1907) presented a paper before the New York Odontological Group, in which he discussed his casting technique and machine. Taggart's success was mostly due to his improved casting machine, since his casting technique was not original; the idea of using wax to form the pattern was that of Martin (1891), and using pressure to cast the alloy was that of Philbrook (1896). Whether Taggart or Philbrook or Martin should be called the father of today's dental casting procedure is debatable. [22]

Taggart was successful in using wax to form the pattern and used pressure to cast the alloys which formed the basis of conventional casting technique. [8]

The conventional way of fabricating metal substructure is lost wax technique.[6] Wax has been chosen for this process because it can be well manipulated, precisely shaped, and completely eliminated from the mold by heating. Fabrication of wax patterns is the most crucial and labor-intensive step in making metal ceramic restoration. [9]

Success in dental casting restorations for fixed dental prosthesis depends on the castability. Castability is described as the ability of an alloy to faithfully reproduce sharp detail and fine margins of a wax pattern. The goal of a prosthodontist is to provide the patient with restorations that fit precisely. Regardless of the alloy used for casting, the casting technique should yield a casted alloy, which should possess sufficient mass, surface hardness and minimal porosity after casting. [10] But, according to studies casting defects can be minimized, but they cannot be removed.

In general, just having a good dental alloy is not sufficient to improve dental castings. Besides a properly formulated and carefully manufactured dental alloy, one needs a good investment, the proper equipment, and well-defined, easily usable technique. The high level of accuracy of the fit that dental castings of today have reached is the result of improvements made in all of these areas — namely, alloy, investment, equipment, and casting technique. [22]

The major disadvantage of casting is casting defects. One of the casting defects is distortion. Any marked distortion of the casting is probably related to a distortion of the wax pattern. In the present study, this distortion is prevented by using 3D printed resin patterns. A concentrated beam of UV light is focused on the surface of a tank filled with liquid photopolymer. As the light beam draws the object on the surface of the liquid, each time, a layer of resin is polymerized or cross-linked until the real object is obtained.[23]

The other casting defect and the defect which affects the results of the current study is porosity. Porosity may occur both within the interior region of a casting and on the external surface. Here we are concerned more with the internal region. The internal porosity weakens the casting. The porosities in the casting cannot be prevented entirely.

This may be the reason for a lot of variation in the values of fracture resistance of fixed dental prosthesis made from 3D Printed Resin Casting was seen. Casting is an art and not a science. Even highly trained technicians can cause casting defects due to human error.

In 3D printing there are two types, one is Direct 3D printing and the other one is Binder 3D printing. Direct 3D printing uses inkjet technology, which has been available for 2-D printing since the 1960s. Like in a 2D inkjet printer, nozzles in a 3D printer move back and forth dispensing a fluid. Unlike 2D printing, though, the nozzles or the printing surface move up and down so multiple layers of material can cover the same surface. Moreover, these printers don't use ink; they dispense thick waxes and plastic polymers, which solidify to form each new cross-section of the sturdy 3-D object. [22] This is also known as stereo lithography (SLA). SLA solidifies the liquid resin layer by layer based on section size, using laser or ultraviolent light. [23] In our study we have used SLA technology and solidification was done with ultraviolet light.

Binder 3-D printing, like direct 3-D printing, uses inkjet nozzles to apply a liquid and form each new layer. Unlike direct printing, though, binder printing uses two separate materials that come together to form each printed layer: a fine dry powder plus a liquid glue, or binder. Binder 3-D printers make two passes to form each layer. The first pass rolls out a thin coating of the powder, and the second pass uses the nozzles to apply the binder. The building platform then lowers slightly to accommodate a new layer of powder, and the entire process repeats until the model is finished. [23]

From a mechanical perspective, 3D printers are often quite simple robotic devices. The apparatus would be nothing without the computer-aided design (CAD) software that allows objects, and indeed whole assemblies to be designed in a virtual environment.

CAD software is commonplace in industrial design, engineering, and manufacturing environments, and is also common in the dental laboratory. Developments in computer technology and software applications are very much a part of the groundswell of technological change that has taken 3D printing to where it is today. For 3D printing to have value we need to be able to create objects to print; CAD software allows us to create objects from scratch. [24]

Kim et al in his study said that the microstructure of Co-Cr sample, cast with 3D- printed pattern is a typical cast microstructure—inhomogeneous, consisting of large grains with dendrite morphology. A large amount of lamellar and blocky carbides of different sizes are located mainly along the grain boundaries. The dendrites consist of γ phase, while the inter-dendritic regions consist of microeutectic (Co solid solution with intermetallic precipitations) and carbides of the (Cr, Mo) 23C6 type. The hardness of the Co-Cr samples, cast with 3D-printed patterns, is in the range 327 HV–343 HV with uneven distribution due to the inhomogeneous microstructure. The dendritic grains with intermetallic phases, precipitated along the grain boundaries, were observed in the microstructure of the samples, cast with 3D-printed patterns.

In the development of the current dental CAD/CAM systems, there were three pioneers in particular who contributed. [13]

Dr. Duret was the first in the field of dental CAD/CAM development. From 1971 he began to fabricate crowns with the functional shape of the occlusal surface using a series of systems that started with an optical impression of the abutment tooth in the mouth, followed by designing an optimal crown considering functional movement, and milling a crown using a numerically controlled milling machine. Later he developed the Sopha System, which had an impact on the later development of dental CAD/CAM systems in the world. [13]

The second is Dr. Moermann, the developer of the CEREC system. He attempted to use new technology in a dental office clinically at the chairside of patients. He directly measured the prepared cavity with an intra-oral camera, which was followed by the design and carving of an inlay from a ceramic block using a compact machine set at chair-side. The emergence of this system was really innovative because it allowed same-day ceramic restorations. When this system was announced, it rapidly spread the term CAD/CAM to the dental profession. [13]

The third is Dr. Andersson, the developer of the Procera system. At the beginning of the 1980s, nickel-chromium alloys were used as a substitute for gold alloys because of the drastic increase of gold prices at that time. However, metal allergies became a problem, especially in Northern Europe, and a transition to allergy-free titanium was proposed. Since the precision casting of titanium was still difficult at that time, Dr. Andersson attempted to fabricate titanium copings by spark erosion and introduced CAD/CAM technology into the process of composite veneered restorations. This was the application of CAD/CAM in a specialized procedure as part of a total processing system. This system later developed as a processing center networked with satellite digitizers around the world for the fabrication of all-ceramic frameworks. Such networked production systems are currently being introduced by a number of companies worldwide. Meanwhile, a number of Japanese universities started research and development of dental CAD/ CAM systems in the latter half of the 1980s and several CAD/CAM systems have been available on the domestic Japanese market. However, there was the unfortunate situation in Japan that, even though there was abundant research on dental CAD/CAM at universities and companies during the past 20 years, dental service was largely provided by the health insurance system, which have resisted the routine application of dental CAD/CAM in clinics. Nevertheless, considering the globalization of dental services, it appeared promising that dental CAD/CAM would obtain approval for the application of all-ceramic crowns and FPDs in Japan in the near future [13]

CAD/CAM milling uses milling tools of different shapes and sizes to mill or grind reconstructions from solid blocks of materials. Many different materials can be processed, e.g., titanium, various polymers, ceramic materials and Co-Cr [25,26]. One major advantage of using milling technology is that some disadvantages of casting, such as casting-induced flaws and porosities which can degrade the quality of the reconstructions, can be avoided as the blanks are manufactured under highly standardized industrial conditions. Therefore, this might be the reason there is less discrepancy between the fracture resistance of 11 CAD CAM fixed dental prosthesis.

Direct metal laser sintering (DMLS) was developed jointly by Rapid Product Innovations (RPI) and EOS (Electro Optical System – Germany), starting in 1994, as the first commercial rapid prototyping method to produce metal parts in a single process.

[27] In dentistry it was first introduced by Deckard and Beaman. [28]

In direct metal laser sintering technique, the Co-Cr or powder granules are fused together by laser welding in layers using a high-energy carbon dioxide laser beam. As small granules are fused together with laser and the layer are also fused together by laser, it has comparatively less porosities than are seen in casting. This might be the reason there is comparatively less discrepancy between the fracture resistance of 11 DMLS fixed dental prosthesis than 11 resin pattern casting fixed dental prosthesis. The technique even allows many prosthetic units to be produced at the same time and the powder granules not exposed to the laser beam can be reused, both important advantages of the technology. Laser sintering is a relatively new method compared to both conventional casting technology and CAD/CAM milling.

The difference between direct metal laser sintering (DMLS) and selective laser melting (SLM) is that, SLM heats the metal powder until it fully melts into a liquid and DMLS does not melt the metal powder, so less energy is needed. Sintering heats metal enough so that their surfaces weld together. Because it does not melt the metal powder, DMLS works with metal alloys (metal mixtures), while SLM works best with pure metals. DMLS and SLM are like sisters. They both have good mechanical properties, because they sinter or melt fine powder particles, therefore the chances of voids are less.[29]

During SLM, layers of metal powder are fused into a real object by a computer- directed laser. The process characterizes with high heating and cooling rates, leading to fine-grained microstructure converted into solidified layer. The specific features of the SLM process, characterizing with high heating and cooling rates, define unique microstructure of Co-Cr dental alloys and mechanical properties higher than that of the cast alloys.

Meacock and Vilar reported that the microstructure of biomedical Co-Cr-Mo alloy, produced by laser powder micro-deposition, is homogeneously composed of fine cellular dendrites. The average hardness was 460 HV0.2 which was tested on hardness testing machine, which is higher than the values obtained by the other fabrication process.

In investigation of Co-Cr-Mo parts, produced by direct metal laser sintering, Barucca et al. established that the microstructure consists of γ and ε phases. The ε phase is formed by a thermal martensitic transformation, and it is distributed as network of thin lamellae inside the γ phase. The higher hardness (47 HRC according to rockwell C scale) is attributed to the presence of the ε-lamellae grown on the γ planes that restricts the dislocation movement in the γ phase.

Lu et al. investigated the microstructure, hardness, mechanical properties, electrochemical behavior, and metal release of Co-Cr-W alloy fabricated by SLM in two different scanning strategies—line and island. They established the coexistence of the γ and ε phases in the microstructure and nearly the same hardness, 570 HV for line-formed alloy and 564 HV for island-formed alloy. Their research shows that the results of tensile strength (1158.22 ± 21 MPa for line scheme and 1115.56 ± 19 MPa for island scheme), hardness, density, electrochemical, and metal release tests are independent of the scanning strategy, and the yield strength of both samples meets the ISO 22764:2006 standard for dental restorations.

The tensile strength of two Co-Cr dental alloys—cast remanium GM and SLM F75— was investigated in the work of Jevremovic et al. They established that SLM samples have more than 1.5 times higher tensile strength compared to the cast ones (1363–1472 MPa and 900 MPa, respectively).

Vandenbroucke and Kruth did complex investigation of SLM titanium and Co- Cr-Mo alloys. The mechanical tests proved that the SLM Co-Cr fulfil the requirements for hardness, strength, and stiffness. Concerning to the corrosion, the SLM Co-Cr samples showed lower emission than the cast ones due to the more homogeneous and finer microstructure of the laser molten material.

The research of Kim et al. shows that the SLM Co-Cr alloy exhibits the laser scan traces, as in higher magnification, the presence of fine grains with sizes about 35μm can be recognized. The SLM samples showed the highest mean ultimate tensile strength, followed by milled/post-sintered, cast, and milled samples. The yield strength and mean percent elongation of SLM alloy were higher than the cast alloy, while Young's modulus was lower. A high yield strength and relatively low but sufficient modulus of elasticity of the SLM samples allow this technology to be used for manufacturing dental constructions, such as removable partial dentures, clasps, thin- veneered crowns, and wide-span bridges.

Averyanova et al. investigated three unit fixed dental prosthesis of Co-Cr alloy, manufactured by SLM. The hardness of the SLM Co-Cr alloy is in the range of 400 ± 14 HV10 and 462 ± 22 HV0.05, while the average tensile strength is 1157 MPa, which is equal to that of the wrought alloy and about twice higher than the cast Co-Cr alloy (655 MPa).

The investigation of Dikova et al. established similar hardness of Co-Cr four unit fixed dental prosthesis, produced by SLM (407–460 HV), which is higher than the cast alloy and has nearly even distribution along the depth of each crown. Their subsequent investigations show that the tensile strength and the yield strength of the SLM Co-Cr alloy are higher than the cast alloy ($R_{0.2}$ = 720 MPa and $R_{0.2}$ = 410 MPa, respectively), while the elastic modulus is comparable (213 and 209 GPa, accordingly). The higher hardness and more homogeneous microstructure of SLM Co-Cr dental alloys determine their higher wear and corrosion resistance.

The Co-Cr dental alloys, fabricated by SLM, characterize with homogeneous fine-grained microstructure, consisting mainly of γ and ε phases in different ratios. Their unique microstructure defines higher mechanical properties, —hardness, yield and tensile strength, fatigue strength as well as higher wear and corrosion resistance— compared to the cast alloys.

Xian zhen xin, jie chen, nan xiang and bin wei did a study on Surface Properties and Corrosion Behavior of Co-Cr Alloy Fabricated with Selective Laser Melting Technique and concluded that within the scope of their study, SLM-fabricated restorations revealed good surface properties, such as proper hardness, homogeneous microstructure, and also showed sufficient corrosion resistance which could meet the needs of dental clinics. They founded that

microstructure of SLM specimens was more homogeneous than that of cast specimens. [30]
In the present study, cobalt chromium alloy is used. Cobalt chrome is the metal alloy of cobalt and chromium. It has a very high specific strength. It was first discovered by Elwood Haynes in the early 1900s by fusing cobalt and chromium. He said that the alloy is capable of resisting oxidation and corrosion and tarnish. It has high wear resistance and it has a high melting point. Co-Cr alloys show high resistance to corrosion due to the spontaneous formation of a protective passive film composed of mostly $Cr2O3$, and minor amounts of cobalt and other metal oxides on the surface.[6] As its wide application in biomedical industry indicates, Co-Cr alloys are well known for their biocompatibility. Biocompatibility also depends on the film and how this oxidized surface interacts with physiological environment.[7] Good mechanical properties that are similar to stainless steel are a result of a multiphase structure and precipitation of carbides, which increase the hardness of Co-Cr alloys tremendously. The hardness of Co-Cr alloys varies ranging 550-800 MPa, and tensile strength varies ranging 145-270 MPa.[8] Moreover, tensile and fatigue strength increases radically as they are heat- treated.[9] However, Co-Cr alloys tend to have low ductility, which can cause component fracture. In order to overcome the low ductility, nickel, carbon, and/or nitrogen are added. These elements stabilize the γ phase, which has better mechanical properties compared to other phases of Co-Cr alloys. Co-Cr alloy dentures and cast partial dentures have been commonly manufactured since 1929 due to lower cost and lower density compared to gold alloys; however, Co-Cr alloys tend to exhibit a higher modulus of elasticity and cyclic fatigue resistance, which are significant factors for dental prosthesis. Co Cr is used more commonly than Ni Cr because nickel is the most common metal sensitizer in the human body. Corrosion resistance of cobalt chromium is more than that of nickel chromium. Nickel has been considered carcinogenic to humans since the work of International agency for Research on cancer (IARC) of the WHO. Cobalt Chromium is preferred in place of gold, as it is cost effective.
Cobalt chromium alloys contain at least 60 wt% cobalt and are always multiple- phase. The chromium content is generally at least 30 wt%, and carbon is often added to strengthen the alloy. Although these alloys can be used for full cast, metal-ceramic, or removable partial denture restorations. [31]
Viennot S, Dalard F in his study concluded that cobalt chromium alloy has very high resistance to corrosion. He even concluded that due to many beneficial mechanical properties of cobalt chromium alloy, it can be used to manufacture fixed dental prosthesis. [32]
Although fusing porcelain to a base-metal alloy is more technique-sensitive than fusing porcelain to a noble alloy, in vitro studies have shown that the bond strength of porcelain to base-metal alloys may be equal, or greater than, the bond strength to high- noble alloys. Moreover, the higher melting range of base-metal alloys compared with high-noble alloys is advantageous because it reduces the risk of distortion and sagging during porcelain firing. [5] For metal-ceramic dental restorations, cobalt-chromium alloys are currently considered the best dental base metals in terms of cost, biocompatibility, handling properties, and clinical performance. [2] Although also according to wear resistance, the cobalt chrome is less abrasive to enamel than a soft gold alloy. [4] Strengthening of alloys can be done for more mechanical advantage. It is
obtained through combination of solid-solution hardening by the addition of carbon, chromium, molybdenum, tungsten or nickel to the pure cobalt matrix [7].
The castability of dental alloys should also be considered, since it is directly related to restoration success. Poor or inconsistent castability often leads to costly laboratory remakes of restorations. Although the castability, as a property of dental casting alloys, may be affected by several external factors like composition of the alloy. Castability values obtained by all methods, may be reflected by the properties of dental alloys. [28] Thus, improvement of the alloy composition and its handling play a fundamental role not only in its physical and mechanical properties, but also in the quality of casting. Within this context, the use of casting machines that enables controlling of casting conditions may represent a significant contribution to improving the quality of dental restorations with base metal alloys. Manufacturers generally recommend an ideal casting temperature or temperature range. However, no information has been given concerning casting atmosphere, considering that casting in the presence of oxygen may cause oxidation of metal elements with substantial reaction power. [33]
When dental alloys are cast by direct gas-oxygen flame, the visual control of this casting method is the only criterion with which to decide the precise moment to start centrifugation. Therefore, the procedure is vitally dependent on the technician's ability to find that critical moment. In such situations, the defect may represent to be more of a failure in the casting procedure. [33]
In the present study it is seen that the fracture resistance of cobalt chromium four unit fixed dental prosthesis manufactured with CAD/CAM Milling is more than DMLS and 3 D Printed resin pattern casting is not statistically significant due to large differences in fracture resistance in 10 samples made by this group. According to studies when seen under high magnification microscope cast specimens shows multiple porosities and few porosities in milled specimens and very few porosities in DMLS specimens. Studies even reveal that casting process does not achieve sufficiently homogeneous alloy. This may be the result in restorations with weaker areas and thereby reduced clinical success. The observed

differences are due to the differences in crystal growth during manufacturing. Casting results in a loose pattern of dendritic growth, and the material produced for milling has a denser structure and a more homogenous microstructure. Laser sintering results in coarse, granular structure, where the direction of the crystal growth varies throughout the material. [3]

According to studies, CAD/CAM is a subtractive technology. In CAD/CAM, large blanks of cobalt chromium are milled in milling unit with the help of burs. The blanks of cobalt chromium are fabricated in high industrial conditions. Now a days 0.01mm burs are also available. Therefore, fit and accuracy of the prosthesis has increased. The milling machine then executes commands for removing material that is not wanted in the final product. This method is also known as "subtractive manufacturing. In casting, the blocks of cobalt chromium are melted at particular temperature mentioned by the manufacturer of alloy and under controlled atmosphere. But preventing complete casting defects is not possible. DMLS, which is an additive technology, here the powder is sintered in layer-by-layer fashion. But preventing complete casting defects is not possible. Therefore, the chances of porosities and flaws are more here to. Therefore, the fracture resistance is comparatively less than CAD/CAM milling.

Mehmet selim bilgin compared the fracture resistance of Co-Cr post-cores fabricated with 3 different techniques: traditional casting (TC), computer-aided design and manufacturing (CAD/CAM) milling (CCM), direct metal laser sintering (DMLS). They concluded that Co-Cr posts fabricated by traditional casting and DMLS systems performed similarly in terms of fracture resistance, posts fabricated by milling techniques showed higher values of fracture resistance values. Co-Cr metal posts fabricated by milling and DMLS could be an alternative to traditional casting processing in daily clinical application. [34]

Nehir Ozden and Mohammed Abujalala compared the internal fit, accuracy, bond strength, marginal adaptations, connection between the veneer porcelain, fracture strength marginal and internal gaps of laser-sintered dental prosthesis with conventional casting techniques. They concluded that Laser sintering seems to be an alternative technique to conventional casting of dental alloys. Complex shapes from metal alloys can easily be produced with these devices. These systems can be helpful to obtain fixed restorations, facial prostheses, titanium implants, surgical models and stents. The new laser-sintering technique appears promising for dental application, but additional studies of properties of laser sintered FPD are necessary. [35]

Paulo Piloto in his article said that metal ceramic fixed dental prosthesis are suitable to increase fracture resistance presenting higher clinical longevity. This type of prosthesis is mainly used when a great number of teeth replacements are needed. [36]

Bjarni Elvar Pjetursson in his study concluded that survival rates of all types of all-ceramic FDPs were lower than those reported for metal-ceramic FDPs. The incidence of framework fractures was significantly higher for reinforced glass ceramic FDPs and infiltrated glass ceramic FDPs, and the incidence for ceramic fractures and loss of retention was significantly higher for densely sintered zirconia FDPs compared to metal- ceramic FDPs. [37]

Metal ceramic FPDs have the vital role to withstand occlusal forces applied in the posterior region. [28] Terry R Walton (2002) in his study concluded that Tooth-supported FPDs have an expected survival rate of 85% at 15 years when the described clinical and laboratory protocol is applied. [28]

Lucia Denti reported that studies on the longevity of FPDs fabricated using high- noble alloys have shown survival rates of 80 to 98%, 81 to 97%, and 74 to 85% after 5,
10, and 15 years, respectively [18]

NA Fernandes in his study concluded that gold restorations are still the "gold standard" with a 96% over 10 years survival rate, followed by porcelain-fused-to-metal crowns (PFM) (90% over 10 years), and all ceramic crowns (75-80% over 10 years). Amongst the ceramic restorations, eMax shows the longest survival rate (90% over 10 years), and Zirconia the lowest (88% over five years). [38]

Pruitt and Chakravartula in 2011 wrote in their article metallic structures exhibit superior fracture resistance properties (Pruitt and Chakravartula 2011) compared to all- ceramic based systems, despite the innovations and properties associated with yttria stabilized zirconia (Piconi and Maccauro 1999), which has relatively high fracture toughness due to the tetragonal to monoclinic phase toughening mechanism. [3]

Raquel Castillo-de Oyague's study suggests that selecting the most appropriate and predictable dental alloy and casting technique is a crucial factor in the long-term success of the restoration. This study concluded that fracture strength of metallic frameworks depended on the alloy type and investing and casting procedures. [30]

Hanaa I. Sallam in her study concluded that CAD/CAM soft milled/post sintered Co-Cr frameworks may be considered promising alternatives to conventional cast
frameworks for implant supported metal ceramic bridges in terms of marginal accuracy and fracture resistance. [14]

Radhakrishnan Prabhu study concluded that DMLS metal-ceramic fixed partial denture prosthesis had a survival rate of 95.5% and yielded promising results during the 5-year clinical study. [28]

Suliman and Steyern carried out an in vitro study in which they showed that there is no difference in fracture strength of metal ceramic restorations with copings fabricated from Co-Cr using different production technologies (casting, milling and laser sintering). [17]

Similarly, Krug et al reported that cast and CAD/CAM milled and sintered metal ceramic bridges were comparable in their fracture performance. [38]

Mehmet Selim Bilgin in his study concluded that Co-Cr posts fabricated by Traditional Casting and Direct Metal Laser Sintering systems performed similarly in terms of fracture resistance, but posts fabricated by CAD/CAM techniques showed higher fracture resistance values. [26]

Dental restorations are subjected to intermittent occlusal forces during mastication and swallowing. The chewing frequency and the places of maximal force on the occlusal surface were relatively constant [38]. The forces exerted on the occlusal surface seldom exceeded 10 to 15 pounds in normal chewing. Also, the maximum bite forces that the stomatognathic system can exert in the posterior area (300 to 880N) for the first molar. Some bruxers and clenchers present bite forces six-times higher than that of non-bruxers, while Patients with symptoms of dysfunction of the masticatory system show lower bite forces that increase as the symptoms disappear. Additionally, there are differences for biting forces among male and female where 807N for males and 650N for females in the molar region. The biting force decrease with increasing age, because of the age-dependent deterioration of the dentition. [23]

P seaton in his study wrote, movements caused by the application of chewing loads to a fixed partial denture (FPD) are predictable. There are translational and rotational movements of the FPD and flexure of the structure. According to this article, the magnitude of stress also depends on length of the fixed dental prosthesis. The longer the length of the prosthesis, the more the stress concentration on the prosthesis. [30]

It was a five unit zirconia bridge, therefore 45 was the middle pontic. According to this study, stress distribution is higher in middle areas of fixed partial denture. [21]

Rezaei SMM, Heidarifar H, Arezodar FF, Azary A, Mokhtarykhoee (2011) conducted a study to determine the effect of buccolingual increase of the connector width on the stress distribution in posterior fixed partial dentures made of IPS Empress. The buccolingual connector width varied from 3.0 to 5.0 mm. Bridges were vertically loaded with 600 N at one point on the central fossa of the pontic, at 12 points along the cusp-fossa contact (50 N each), or at eight points along the cusp-marginal ridge contact (75 N each). Alternatively, a load of 225 N was applied at a 45º angle from the lingual side. The results showed that stress concentrations were observed within or near the connectors. The von Mises stress decreased by increasing connector width, regardless of whether the loading was applied vertically or at an angle. They concluded that that increasing the connector width decreases the failure probability when a vertical or angled load is applied. [31]

Ayesha Aslam in her study said, for clinical longevity, connectors of a fixed partial denture should be thick enough to resist the occlusal loads. However, for optimal aesthetics, occlusal and gingival embrasures must be created.

Osama Saleh in his study concluded that, the 3D printed pattern resulted in an E-max crown with better marginal adaptation and internal fitness. The fracture resistance of E-max crown was improved as its internal adaptation was enhanced. [32] Four unit fixed partial dentures are commonly used in clinical practice. It is a long span fixed partial denture. Chances of failure of four unit fixed partial denture are more than three unit fixed partial denture. If all other parameters are kept constant a four-unit fixed partial denture replacing two teeth will flex (bend) eight times more as compared to a three-unit fixed partial denture with a single pontic. [40]

CONCLUSION

The present in-vitro study was carried out to comparatively evaluate the fracture resistance of CAD-CAM Milled, 3 D Printed Resin Pattern Casted and Direct Metal Laser Sintered Cobalt Chromium Copings.

The conclusions drawn within the limitations of the present in-vitro study were:

The fracture resistance for Group A (CAD-CAM milling) was evaluated as 6247.2 N. This value was within the clinically acceptable range. The fracture resistance for Group B (3D Printed Resin Pattern Casting) was evaluated as 6298.5 N. This value was within the clinically acceptable range. The fracture resistance for Group C (Direct Metal Laser Sintering) was evaluated as 5153 N. This value was within the clinically acceptable range. There is a lot of variations in the values of 3 D Printed Resin Pattern Casting, hence it does not have statistically significant difference in fracture resistance. Hence, the fracture resistance amongst all the groups was exhibited in the following order: a)CAD/CAM Milling b)Direct Metal Laser Sintering (DMLS). The fracture resistance for CAD/CAM Milling was better than all the other groups. 3 D Printed Resin Pattern Casting has the highest average fracture resistance amongst all the three groups. But according to statistical analysis, there is a lot of variations in the values of 3 D Printed Resin Pattern Casting, hence it does not have statistically significant difference in fracture resistance. There was a statistically significant difference between the fracture resistance of two groups; CAD/CAM Milling and 3 D Printed Resin Pattern Casting.

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