Journal of Engineering and Applied Sciences 13 (23): 9868-9879, 2018
ISSN: 1816-949X

Manufacturing of Zirconia Ceramics for Dental Applications by Computer Aided Design and Manufacturing (CAD/CAM)

Iman Jabber Abed and Doha Bassim
Department of Materials Engineering, University of Kufa, Najaf, Iraq

Abstract: The parts and components can be manufactring by computer aided design and manufacturing (CAD/CAM) and machined with very accuracy using a computer with joined software linked to a milling machine. CAD/CAM technology manufactures prosthesis from the data obtained from the patient's mouth. The aim of the current research is to fabricating crowns and bridge made of different brands of Yttria stabilized Tetragonal Zirconia Polycrystals (3-5 mol% Y-TZP) and employed different cutting speed by Computer Aided Design and Manufacturing Techniques (CAD/CAM). The experimental work involved the fabricating of crowns only from Yttria stabilized tetragonal zirconia polycrystals (dental direct block) and bridges made from three different brands of Yttria stabilized Tetragonal zirconia polycrystal (Kerox dental block, dental direct block and VITA YZ HT block) by computer aided design and manufacturing techniques and studying the effect the cutting speed on the surface properties and mechanical properties of zirconia such as flexural strength and microhardnes. Other characteristic techniques have been performed such as XRD and SEM to characterization of zirconia. It is found from the results of X-ray diffraction patterns carried out on presintered zirconia blocks the predominant phase is Tetragonal (T) and the appearance of the monolithic phase at some the peaks but after milling and sintering only Tetragonal phase (T) was identified for all of the samples except in Kerox dental block the monolithic phase was appear at 150 m/min cutting speed. The results scanning electron microscope of the surface morphology for each brands reveals milling traces and scratching induced by the milling tool also crack and microchips and pores appeared on the surface machining due to densification process. The results of mechanical tests were found flexural and hardness decrease with increasing cutting speed.

Key words: CAD/CAM, crown, bridge, flexural strength and sinterin, tetragonal, properties

INTRODUCTION

Zirconia start in the medical implant field due to excellent biomechanical properties over alumina such as higher fracture toughness, strength and best wear performance. The original employ of zirconia in the medical field was for total hip replacements in about 1985. It has since, gained popularity in the dental field as an alternative to alumina for employ as endodontic posts, crown and bridge restorations, implant abutmentsand orthodontic brackets (Lucas, 2015). The zirconia has produced considerable interest in the dental community as a dental substance also zirconia have perfect characteristics such as dimensional stability, chemical properties and young modulus of elasticity (210 GPa) identical to that found in of stainless steel (193 GPa), high mechanical strength and toughness, therefore, it is using widely build prosthetic devices. Always the mechanical properties of zirconia high amount for any dental ceramicszirconia has a high temperature stability and melting point (2680°C), low thermal conductivity (<1 W/mK), high thermal expansion (>10×10^{-6} 1/K), high hardness (1200-1350 HVN) and a good thermo-shock resistance (T = 400-500°C) (Ozkurt and Kazazoglu, 2010; Rosentritt, 2008). The zirconia use in dental filed currently content higher than 90% include ceramic with a Zirconium Dioxide (ZrO_2) such as glass infiltrated ceramics with 35% Partially Stabilized tetragonal Zirconia (PSZ) and Yttrium stabilized Tetragonal Zirconia (Y-TZP), these materials have a large domain of clinical uses, from implant abutments and elementary to Fixed Partial Dentures (FPDs) tooth restorations including several elements because of the superior mechanical properties of Y-TZP ceramics (Cavalcanti *et al*., 2009a, b).

Zirconia has properties that overcome the problem associated with alumina bioceramics, i.e., the brittleness of that material. This is most evident in partially stabilized zirconia (3Y-ZrO_2) which has a phase transformation toughening mechanism that helps prevent cracks from propagating. However, partially stabilized zirconia in the presence of bodily fluids undergoes an undesirable roughening of the surface and micro-cracking (Vythilingum, 2013).

Corresponding Author: Iman Jabber Abed, Department of Materials Engineering, University of Kufa, Najaf, Iraq

Table 1: Summary of materials used their manufacturer and chemical components

Abbreviations	Zirconia materials	Manufactures	Chemical components (%) (main component)	Typical properties of sintered body
V	VITA YZ HT	VITA Zahnfabrik	ZrO_2+HfO_2	Cofficient of thermal expansion
		Sackingen Germany	$Y_2O_3$4.95-5.26	(20- 500°C) approx. 10.5 10^{-6}. (K^{-1})
			Al_2O_3 = 0.15-0.32	Density = 6.05 (g/m^3)
			SiO_2 Max. 0.002	Chemical solubility<20 ($\mu g/cm^2$)
			Fe_2O_3 Max. 0.01	Bending strength>1200 (MPa)
			Na_2O Max. 0.04	Surface hardness>1250
				Flexural strength approx. 1.200 (MPa)
Z	Dental zirconia	Kerox Dental	$ZrO_2$90.2-94.3	
		Hungary	Y_2O_3 5.7-9.8	Density = 6.05 (g/cm^3)
			Al_2O3<0.25	Bending strength = 1500 (MPa)
			SiO_2<0.02	Surface hardness = 1250 HV
			Fe_2O_3<0.02	Fracture toughness upto 16 [Mpa√m]
			Na_2O<0.02	
D	DD bio ZW high	Dental Direkt	ZrO_2+HfO_2+	Density>6.05 (g/cm^3)
	strength zirconia	Industire Zentrum	3Y_2O_3>99	Fracture toughness>9.5 [MPa√m]
		Germany	Al_2O_3<0.5	Flexural strength 1200±250 (MPa)
			other oxide<1	Emodulus>200 (GPa)

By using CAD/CAM technology, most zirconia frameworks of FPD and single crowns can be milled from blocks of zirconia (Helvey, 2008; Hmedat and Ibraheem, 2013; Othmanm, 2017). Zirconia-based Fixed Partial Dentures (FPDs) because they can be used on molars, they have a wider application than other ceramics zirconia allows the bulding of resistance structures for chewing stresseson posterior tooth. Zirconia (ZrO_2) refers to fixed partial dentures restorations supported by implants or teeth. Although, to complete denture restorations, some manufacturers suggest are maximum potential them 5-unit FPDs (Ozkurt and Kazazoglu, 2010). Zirconia can be used for a variety of applications because we can modify the properties of the composite as required by the application (Kandaswamy, 2010). The computers are design and manufacturing (CAD/CAM) were inserted in the 1960's for use initially in the manufacturing industry and first applied to dentistry (Ding, 2016).

Duret and colleagues develop or be the first to use the Computers for Design and Manufacturing (CAD/CAM) in the early 1970s. Based on the research of the researcher system, many researchers in the world began to development system to manufacture a crown in the 1980s, this technology uses computer togather information, design and manufacture a wide range of products (Uzun, 2008). Prefabricated ceramic monoblocks is using in the intraoperatory application for one-appointment restoration fabrication. In parallel, commercial production centers and expanded the range of the materials used and the types of restorations that are produced due to the emergence of design and manufacturing computer-aided system in dental laboratories. In CAD/CAM systems, zirconia blanks are milled both, immediately after cold isostatic pressing as green body sample and after compressed sintering as a hardened (sintered) blank (Tinschert *et al*., 2007; Strub *et al*., 2006).

Computer aided design and manufacturing of crownand fixed partial denture contain many advantages such as reduced labor, application of new materials, quality control and cost effectiveness, also, some limitation for using CAD/CAM technology in fabrication dentistry have to be mentioned, the first investment of machines was high overextend the budget of smaller laboratories and due to software and production procedures, some applications are limited (Miyazaki and Hotta, 2011; Beuer *et al*., 2009).

This research aims to fabricate crowns and bridge made of from different brands of Yttria stabilized Tetragonal Zirconia Polycrystal (3-5 mol% Y-TZP) by computer aided design and manufacturing techniques and analyze effect of cutting speed on mechanical and microstructural properties of zirconia crowns and bridges constructed using dental CAD/CAM.

Experimental part: In this research three type of presintered zirconia were employed according to manufacturing procedure:

- Block VITA YZ HT block (VITA Zahnfabrik Sackingen Germany) consist from Yttria stabilized tetragonal zirconia polycrystal, ZrO_2 (3 Y_2O_3 mol%) with factor of shrinkage 1.230
- Block zirconia blank (Kerox dental zirconia- Germany) with factor of shrinkage 1.231
- Block dental direct zirconia (high strength zirconia DD Bio ZW ISO- Germany) with factor of shrinkage 1.248 (Table 1)

Tools

CoriTec i3Dscan eco: The new i3D scan eco is an extremely compact, fully automatic dental scanner. In addition to the basic functions like crowns and bridges,

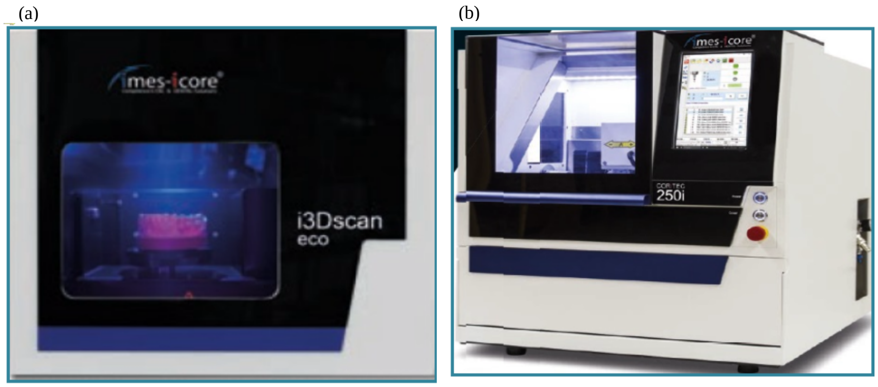

Fig. 1: Illustration of; a) The three dimension scan of imes-icore system; b) The CorTec 250i machine systems

expansion modules are readily available such as articulator inlay, onlay veneer, splints, telescope or individual abutments as show in Fig. 1a.

CoriTec 250i machine systems: The CorTec 250i dry machine systems is the most widely used CAD/CAM systems for applications of zirconium dioxide, PMMA or wax. Thanks to the 5-axis technology, these machines can also produce complex dentures with diverging stumps and undercut areas without further rework as shown in Fig. 1b.

MATERIALS AND METHODS

The dentoform tooth was prepared from gypsum and consider it as a modelof the master die to complete the construction of the zirconia bridge and crown by the CAD/CAM imes-icon machine. Then put the master die with the gypsum base on the scanning table for the 3D optical scanner and switch on the scanner and CAD computer to start the scanning as a following procedure:

Open the program exoCAD, the first step open the page to enter the information of the case such as: (patient name, technician name, address and dentist name) and save data. Choice the type of the restorations (bridge and crown with minimum thickness 0.5 mm) and select type of material.

Press scanning imes-icons to start the 2D scanning to determine the position of the bridges and crowns and then continue with the 3D scanning, the 3D scanner will tack multipicture and then press match icon to get the 3D picture of the master die. 3D pictures were shown on the PC screen.

Design of the bridge and crown by press the design icon to open the design window and start the designing of bridge and crown, the first step determined the finishing line after complete the design of thebridge and crown copy it as a STL file (Standard Tessellation Language). It describes a physical space mathematically as a series of small triangles using a three-dimensional coordinate systemand send it to the CAM computer to amount the design bridge and crown in to the zirconia blank. The bridge and crown fixed in to the blank by connect and then calculate the bridge and crown to the milling computer. The milling computer will draw the calculated bridge and crown from the CAM computer for bridge and crown milling.

The milling computer will receive the calculated crown and bridge from the CAM computer for crown and bridge milling at the same manner copy (6) crown STL files for two groups and (18) bridge STL files for six groups, six zirconia crowns for each group and 18 zirconia bridge for each group. Zirconia crowns and bridge have a one 3D scanning and one design and then the complete designed crown STL file copied (6) STL files and bridge STL file copied (18) STL files, so, we have a standardizes in 3D scanning, designing and thickness of the crowns and bridge. After completion of the milling process and before sintering, all the restoration carefully cut off from the block holder or the disc using a tungsten carbide bur and then the attachment point must be smoothed. Finally, placed the bridge and crown in furnace to full sintering.

Manufacturing of sample

Manufacturing of zirconia three-unit fixed dental prosthesis: In this research, 18 specimen with computer aided design and manufacturing framework of 3-unit bridges were machined from commercially three different brands of zirconia presintered blocks with two different velocity of the bur 100 and 150 m/min. The first group consist of six samples machining from vita zirconia block, second group (six samples) machined from zirconia block and third group(six samples) machined from dental direct zirconia block. For the three-unit Fixed Dental Prosthesis (FPDs) elaboration. The user model contains these details the dental first premolar and first molar teeth were employed as retainers to assist a three unit bridge. Teeth have been put on the sample with shoulder circumferential thickness of 2 mm, space required for the structure of the bottom of the zirconia (0.4 mm).Then the dental prostheses were fully sintered to 1500°C for 2 h holding time with the heating rate 8°C/min and 3 h cooling rate for sintering for zirconia block and dental direct zirconia block. For the fully sintering vita YT zirconia block.

Zirconia crown manufacturing: The dentoform was prepared for left upper first premolar to get each ceramic crown. In this study 6 crowns from dental direct presintered zirconia block were machine by CAD/CAM system (full anatomy crown with minimum thickness 0.5 mm). This samples divided to two groups with different cutting speed 100 and 150 m/min. The crown with the following features shall have a minimum wall thickness 1 mm and a thickness of 0.04 mm for cement gap, the cement space started at 0.25 mm after complete dental prostheses the samples were fully sintering to 1500°C for 2 h at 8°C/min. The full sintering treatment was accomplished in (Zirkonofen 600 furnace, Zirkon Zahn company).

Surface characterization

X-Ray Diffraction (XRD): All processed surface under each condition were X-ray scanned using an X-Ray Diffractometer (XRD) (The Shimadzu Lab XRD-6000, Japan X-ray) to obtain X-ray patterns, recognize crystalline phases existing in the zirconia presintered pieces and sinter pattern next of machining operation. The X-ray tube was operated at 40 kV and 20 mA at a wavelength of 0.154 nm. The scanning range was 10-70° (2θ) using the step size of 0.02° and 0.5 sec/step.

Scanning Electron Microscopy of zirconia (SEM): After completed sintering the microstructure, the morphology of crown and bridge surfaces also, the fracture surface morphology was examined directly without any treatment after the mechanical tests were observed by employing SEM (inspect S50 SEM, Japan made). The test accomplished in Faculty of Sciences, University of Kufa. The patterns were covered with gold utilizing a sputter coating machine.

Mechanical tests: The tests were carried out on samples zirconia after machining by CAD/CAM and sintering to compute mechanical properties such as flexural strength test and microhardness test.

Flexural strength of zirconia: The three-point bending experiment were done in mechanical testing machine with loading speed 0.5 mm/min. In the prosthesis test, a 6.5 mm steel ball, set on a jut was utilized to lift the contact surface. All loads were applied centrally at the point of focus of the pontic.

The FDPs were considered beam supported by two strips at an equal distance from the loading screws for calculate the flexural force. The bending strength (σfs) was computed utilizing the Eq. 1:

$$\sigma_{fs} = 3FI/2BH^2 \quad (1)$$

Where:

F = The maximum load (N)
I = The distance between the two rests on the surface under the tensile force (mm)
B = The width (mm)
H = The Height of the specimen between the surface under force (Callister and David, 2014)

Width and height of specimen were 4 and 3 mm, respectively, the distance between support points was 15 mm and the span length was 22 mm. The test were performed using the microcomputer controlled electronic universal testing machine (WDW-100E, China made) at Materials Engineering Department/Faculty of Engineering, University of Kufa.

Microhardness: Vickers hardness values were measured on samples after machining by CAD/CAM and sintering, using a Vickers hardness diamond indenter (TH-714 Digital Micro Vickers hardness tester) at Materials Engineering Department/Faculty of Engineering, University of Kufa. The 9.8N load was used and the time of holding was 30 sec. The average of 4 measurements was taken as an average value of HV bridge restoration samples.

RESULTS AND DISCUSSION

X-ray diffraction analysis: X-ray diffraction samples had been received from zirconia prior and after milling and sintering to 1500°C for 2 h holding time with the heating rate 8°C/min for Yttria partially stabilized zirconia block (Kerox dental block) and high strength zirconia DD Bio ZW ISO(dental direct block), the cooling rate was 3 h, also, the full sintering to 1530°C for Yttria stabilized tetragonal zirconia polycrystals (VITA YZ HT block). Figure 2 shows the X-ray diffraction for three different brands of pre sintered zirconia block. In VITA YT-ZrO_2, zirconia blank and dental direct block presintered blocks, the predominant phase is Tetragonal (T) but the appearance of the monolithic phase at the peaks

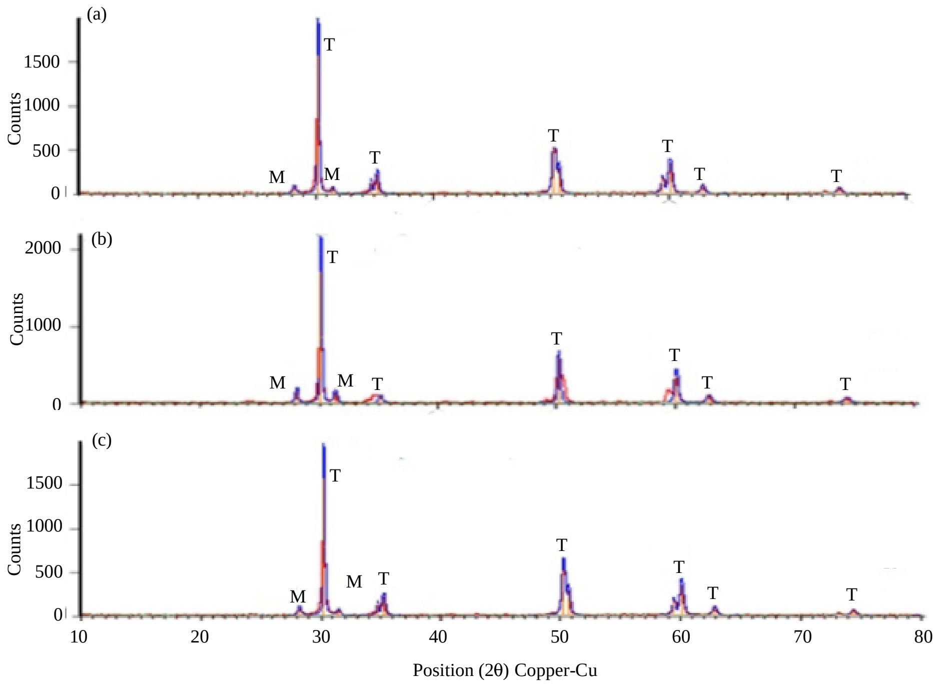

Fig. 2: X-ray diffraction for three different brands of pre sintered zirconia block; a) VITA YZ HT ; b) Kerox dental zirconia and c) Dental direct zirconia

2θ = 28.2431 and 31.4539 for VITA YT-ZrO_2 in Kerox dental block appeared in 2θ = 28.105 and 31.3536 also in dental direct block the monolithic phase at the peaks 2θ = 28.1524 and 31.3541. It can be notice that the monoclinic was disappearancein the rest of the peaks. This agrees with Elias *et al.* (2014) when he studied the crystalline phases of the sintered and presintered samples where they found that the presintered standardizes samples contained the monoclinic (10.5%) and tetragonal (85.8%) phases. The presence of the monoclinic (10.5%) phase is actually associated with the cutting method of the samples prior to the milling which allowed the conversion of the tetragonal phase into a monoclinic phase (Elias *et al.* 2014).

While Fig. 3 and 4 shows the X-ray diffraction pattern of the samples after machining by CAD/CAM at different speed 150-100 m/min and then sintered. The peak locations and their relative intensities for the ZrO_2 phases were cited from the International Center for Diffraction data (ICDD) database. Only Tetragonal phase (T) (ICDD 00-060-0505) of ZrO_2 was identified for VITA YZ HT and dental direct zirconia samples. In addition to it was noticed that the cutting speed not affect on transformation of monoclinic phase to tetragonal phase but when sintering to high temperatures, the monoclinic phase transformation to Tetragonal (T) phase, this agreement with Shaik E H when study XRD patterns of the ceramic sintered at various temperatures. The XRD pattern for the sample sintered at 1250°C and above shows a fully stabilized tetragonal structure (Hoosain, 2011). The CAD/CAM milling do not lead to phase transformation but sintering processes after milled process transformation monoclinic to tetragonal phase (Arroyave *et al.*, 2002). Further, the complete transformation of monoclinic to tetragonal phase following the sintering process after milling. This is expected because the sintering temperature (1530°C) was high enough to induce this transformation (Alao *et al.*, 2017). But in Kerox dental brand with cutting speed 150 m/min as shown in Fig. 4 does not agree with Shaik E H, the monoclinic was appear when 2θ = 31.0248 then the monolithic phase disappears in the rest of the pick peaks, the appears the monolithic phase in this case due to machining process by CAD/CAM and residual stressesduring the cooling processes, this result agreement with Elias *et al.* (2014). The content of monoclinic phase identified in the sintered samples is probably related to the residual thermal stresses developed during the cooling process and also by spontaneous transformation into the surface of the zirconia phase Elias *et al.* (2014).

Scanning electron microscope of samples: Scanning Electron Microscope (SEM) was used to determine the morphology of the zirconia block from three different brands. Figure 5 shows the surface morphology at different magnification powerfor VITA YZ block after

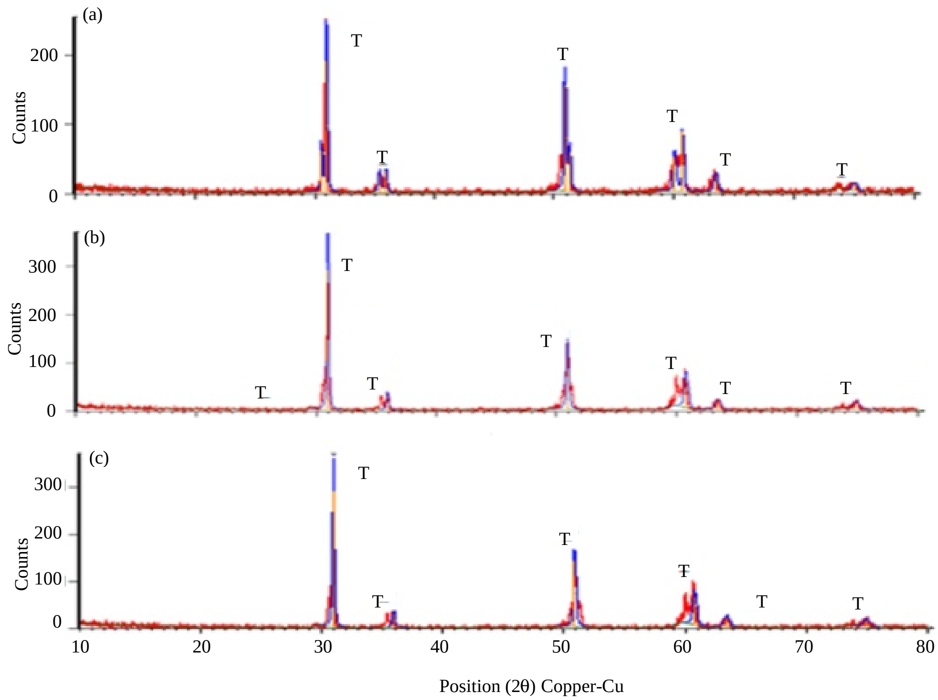

Fig. 3: X-ray diffraction pattern of for three different brands of after machining by CAD/CAM at cutting speed 150 m/min and and sintering; a) VITA YZ HT; b) Dental direct zirconia and c) Kerox dental zirconia

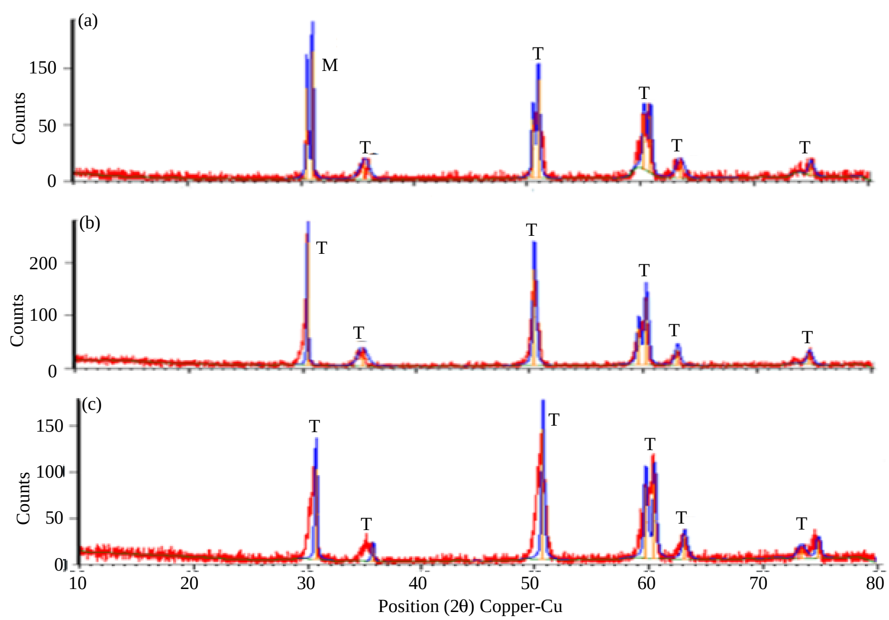

Fig. 4: X-ray diffraction pattern of for three different brands of after machining by CAD/CAM at cutting speed 100 m/min and sintering; a) Vita YZ HT; b) Dental direct zirconia and c) Kerox dental zirconia

milling at cutting speeds 150 m/min and then sintering. Figure 5a reveals machining groove associated with plastic deformation and scratchinginduced by the milling tool. Figure 5b shows the detailscrack and microchips located in the surface induced by ploughedmilled surface, indicating the sintering cannot completely microcracks produced in the CAD/CAM milling. Figure 5c show pores appeared on the surface machining due to densification

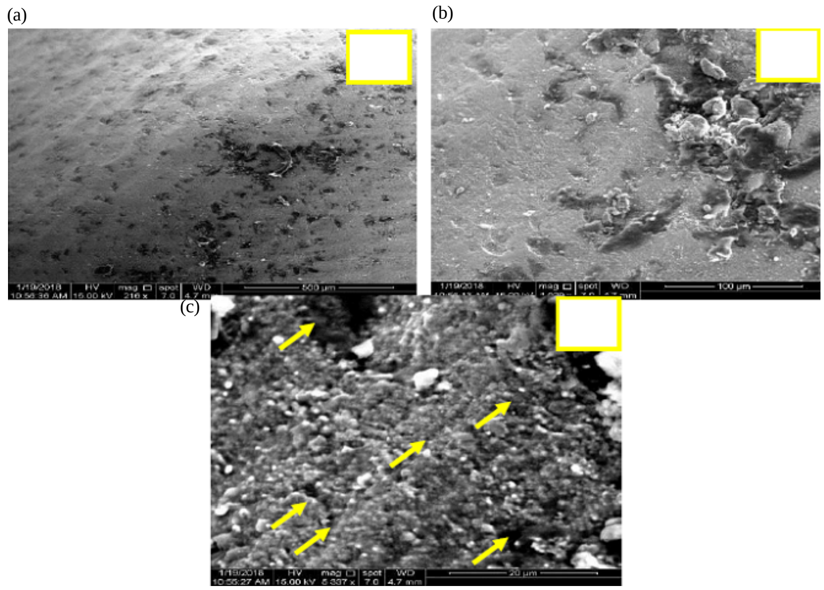

Fig. 5a-c): SEM images for machining VITA HT YZ block and sintering by used cutting speed 150 m/min at different magnification power, the arrow shows the pores

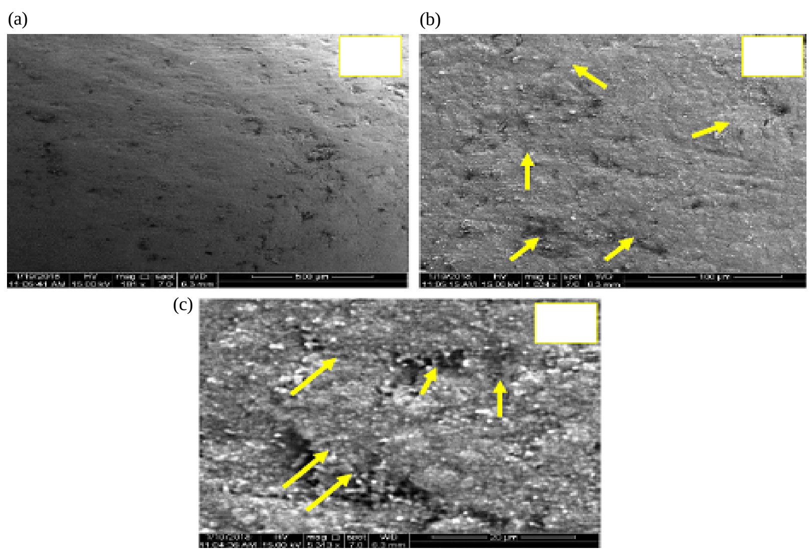

Fig. 6a-c): SEM images for machining VITA HT YZ block and sintering by used cutting speed 100 m/min at different magnification power, the arrows show the cracks and pores

process, also, observed residual pulverized debris in form very fine particle on the machined surface. Figure 6 shows the surface morphology at different magnification powersfor machining VITA HTYZ block after milling at cutting speeds 100 m/min and then sintering. Figure 6a shows that the milled surface with machining traces and scratches. Figure 6b shows surface details at higher magnification in which visible surface defects such asfractures and microcracks were observed. Figure 6c reveals pores appeared due to densification process on the surface machining. The SEM micrographs produced in CAD/CAM machining process revealed the surface morphologyof machining Kerox dental blockafter milling at cutting speeds 150 m/min and sintering with different magnification powers. Figure 7a reveals milling traces and the detailscrack and debris located in the surface induced by milled surface. Figure 7b shows a high magnification power surface image cracks and the adhesion of fine

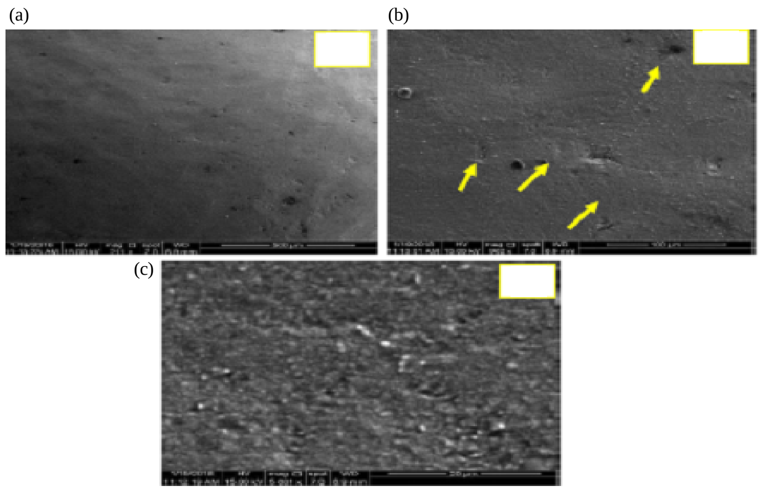

Fig. 7a-c): SEM images for machining Kerox dental block and sintering by used cutting speed 150 m/min at different magnification power

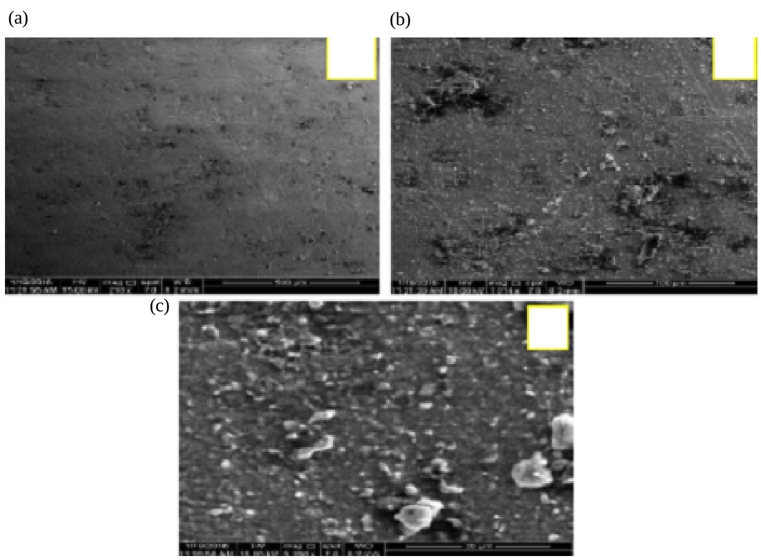

Fig. 8a-c: SEM images for machining Kerox dental block and sintering by used cutting speed 100 m/min at different magnification power

particle of debris. Figure 7c shows the details of the grain coarsening. Combining to X-ray diffraction in Fig. 6a is considered that the grain coarsening resulted from the transformation of monoclinic zirconia to tetragonal zirconia magnification power surface image on which no visiblesurface defects can be observed but only the adhesion of fine particle of debris.

Figure 8 shows the surface morphology at different magnification powersfor machining Kerox dental blockafter milling at cutting speeds 100 m/min and then sintering. Figure 8a reveals scratching produced by the milling tool and milling traces appeared on the surface machining. Figure 8b shows the detailscrack and microchips located in the surface induced by milled surface. Figure 8c shows the detailed smearing of zirconia grains also shows the sintering-induced grain coarsening on the milledsurface. Figure 9 shows the surface morphology at different magnification powers for machining dental direct blockafter milling at cutting speeds 150 m/min and then sintering. Figure 9a reveals machining groove and trace induced by the milling tool. While Fig. 9b shows the detailscrack and microchips located in the surface induced by ploughedmilled surface. Figure 9c shows the sintering-induced grain coarsening

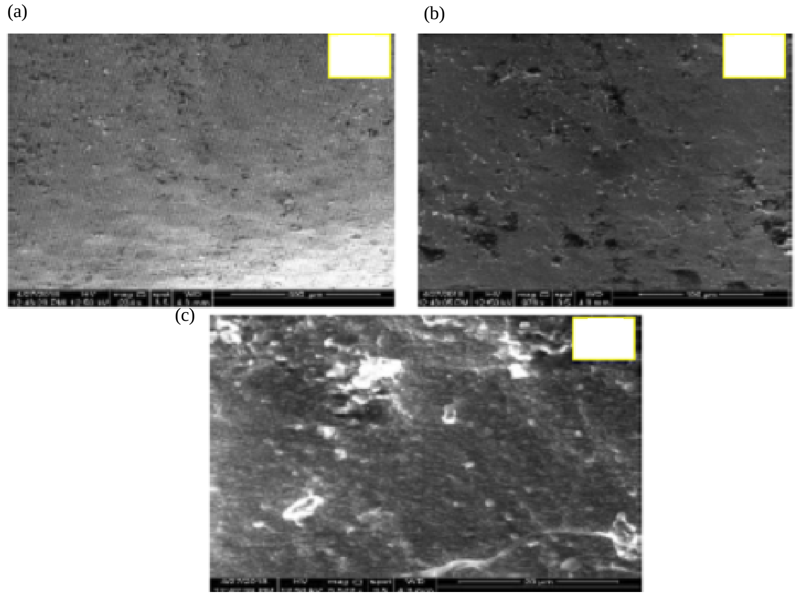

Fig. 9a-c): SEM images for machining dental direct block and sintering by used cutting speed 150 m/min at different magnification power

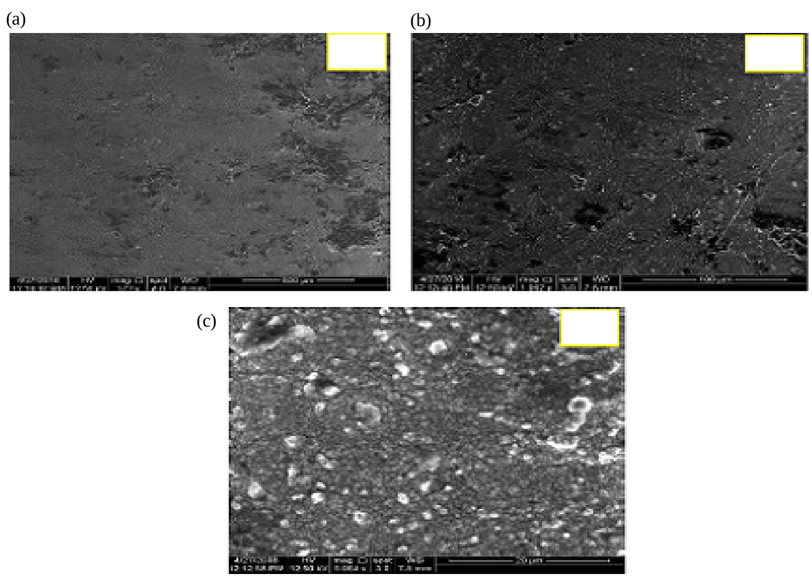

Fig. 10a-c): SEM images for machining dental direct block and sintering by used cutting speed 100 m/min at different magnification power

on the milledsurface, also, observed residual pulverized debris in form very fine particle on the machined surface. Figure 10 shows the surface morphologyat different magnification powers for machining dental direct blockafter milling at cutting speeds 100 m/min and then sintering. Figure 10a shows that the trace of milling surface and scratches due to milling tool. Figure 10b shows surface details at higher magnification in which visible surface defects such asfractures and microcracks were observed. Figure 10c reveals coarsening grain resulted from the transformation of monoclinic zirconia to tetragonal zirconia, also can be observed fine particle of debris.

Mechanical properties

Flexural strength: Figures 11 shows the relation between flexural strength values and type of machined material (dental direct block, Kerox dental block and VITA YZ HT

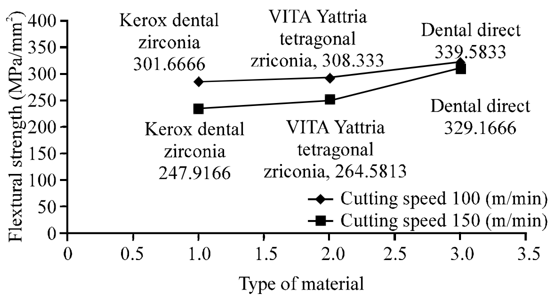

Fig. 11: Relation between flexural strength and type of machined material at two different cutting speed

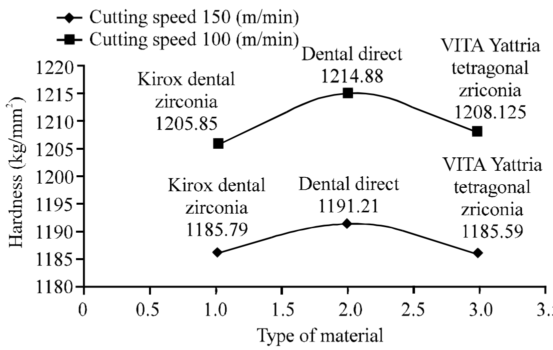

Fig. 12: Relation between hardness and type of machined material at two different cutting speed

block) of the prepared samples using universal testing machine at two different cutting speed (100 and 150) m/min without polishing. The results optioned from the three bending tests show that the values of flexural strength at cutting speed 100 m/min is higher than the cutting speed 150 m/min for three types of zirconia, these results agreement with Yusuke Ito *et al*. which show that the bending strength and fracture toughness of this material decrease with increasing speed andfollowing increase in temperature. It can be noticed the flexural strength values of Kerox dental block increasing from 247.9166 MPa up to 301.6666 MPa when cutting speed increasing from 100-150 m/min while the flexural strength values of VITA YZ HT block increase from 264.5813 MPa up to 308.3333 MPa and the flexural strength values of dental direct block increase from 329.1666 MPa up to 339.5833 MPa.

The best brand at two cutting speed was dental direct block have height values of flexural strength, next brand was dental direct brand have medium surface roughness and zirconia blank VITA YZ HT block have medium flexural strength and Kerox dental Zirconia have lowest values of flexural strength. During CAD-CAM machining, grooves are created on the surface which reduce the flexural resistance. Fine polishing may improve surface finishing but induces a compressively stressed layer and therefore decreases the mean flexural strength. Sandblasting may be useful to increase the strength of dental YTZP (Elias *et al*., 2014).

Hardness: Figure 12 shows Vickers hardness of three different brand of Yttria-zirconia samples after sintering with two cutting speed. The micro hardness values increases with decreasing cutting speed. The hardness values of Kerox dental Zirconia change from 1185.59 up to 1205.85 and the HV values of dental direct, increase from 1191.21 up to 1214.88 while the hardness values of VITA YT Zirconia increase from 1185.58 up to 1208.125. From these results show that the best brand was Kerox dental Zirconia has highest hardness, next VITA YT Zirconia block have medium hardness and Zirconia block have lowest hardness. The results agreement with Yusuke Ito *et al*. It is known that the hardness of Yttria-stabilized zirconia decreases when the temperature increases because of the heat generated by high-speed machining (Kalpakjian and Schmid, 2013). Zirconia is very high hardness and need larger forces and speed less this weakens the tool used during operation and reduces the life of the tool. Which suggests that clinicians and technicians will get a longer life time from their milling burs when using these softer materials. Additionally, computer-aided milling cycles may be speed up if cutting the material requires less force (Lawson *et al*., 2016). Theoretically, the phase change from tetragonal to monoclinic of specimens could cause some compressive stresses to the surface of the specimens. This might have increased the surface microhardness. However, no difference in microhardness was seen in study (Hjerppe, 2010).

The intrinsic deformability and microstructural characteristic such as multiphase and grain size can influence the Vickers hardness of ceramic materials. Another reason can explain the increment in the microhardnes is based on the Hall-Petch relation, since, the grain size is decreased:

$$Hv = H_0 + K.d^{-1/2} \tag{2}$$

Where:

H_0 = The intrinsic hardness
K = An empirical constant for each materials and d is thegrain size (Othmanm, 2017)

CONCLUSION

In VITA YZ HT, Kerox dental and dental direct presintered blocks, the predominant phase is Tetragonal (T) but the appearance of the monolithic phase at the some peaks.

Only Tetragonal phase (T) of ZrO_2 was identified for all the samples for three different brands after millingat two cutting speeds 100 and 150 m/min and sintering but at machining of Kerox dental brand with cutting speed 150 m/min, the monoclinic was appear when $2\theta = 31.0248$.

The cutting speed has not effect on transformation of monoclinic phase to tetragonal phase but when sintering to high temperatures, the monoclinic phase transformation to Tetragonal (T) phase

The use of high cutting speed during the milling processes reduces the mechanical properties of three different brands of zirconia such as flexural strength and microhardnes.

After flexural strength show the pore and flaws on the surface and interior of three different brands which resulted to a lower flexural strength than theoretical value. The best brand was dental direct have high flexural strength, microhardnes and the best surface roughness.

ACKNOWLEDGEMENTS

The researchers would like to express their appreciation to the University of Kufa, College of Engineering/Materials Engineering Department for its support of this research. The researchers would like to thank all the technicians in the laboratories for their valuable help.

REFERENCES

Alao, A.R., R. Stoll, X.F. Song, T. Miyazaki and Y. Hotta *et al*., 2017. Surface quality of Yttria-stabilized tetragonal Zirconia polycrystal in CAD/CAM milling, sintering, polishing and sandblasting processes. J. Mech. Behav. Biomed. Mater., 65: 102-116.

Arroyave, R., L. Kaufman and T.W. Eagar, 2002. Thermodynamic modeling of the Zr-O system. Calphad, 26: 95-118.

Beuer, F., P. Neumeier and M. Naumann, 2009. Marginal fit of 14-unit zirconia fixed dental prosthesis retainers. J. Oral Rehabil., 36: 142-149.

Callister, W.D. and G.R. David, 2014. Fundamentals of Materials Science and Engineering: An Integrated Approach. 5th Edn., John Wiley & Sons, Hoboken, New Jersey, USA., ISBN:9781119230403, Pages: 933.

Cavalcanti, A.N., R.M. Foxton, T.F. Watson, M.T. Oliveira and M. Giannini *et al*., 2009. Bond strength of resin cements to a zirconia ceramic with different surface treatments. Operative Dent., 34: 280-287.

Cavalcanti, A.N., R.M. Foxton, T.F. Watson, M.T. Oliveira and M. Giannini *et al*., 2009b. Y-TZP ceramics: Key concepts for clinical application. Operative Dent., 34: 344-351.

Ding, H., 2016. Recycling Zirconia dental prostheses waste: A feasibility study. Master Thesis, The University of Hong Kong, Hong Kong.

Elias, C.N., A.M. Melo, D.H.E.S. Santos and D.C. Santos, 2014. Degradation and mechanical properties of Zirconia 3-unit fixed dental prostheses machined on a CAD/CAM system. Intl. J. Appl. Ceram. Technol., 11: 513-523.

Helvey, G.A., 2008. Zirconia and Computer-aided design/computer-aided manufacturing (CAD/CAM) dentistry. Inside Dent., 4: 72-79.

Hjerppe, J., 2010. The Influence of Certain Processing Factors on the Durability of Yttrium Stabilized Zirconia used as Dental Biomaterial. University of Turku, Turku, Finland, ISBN:978-951-29-4384-5, Pages: 74.

Hmedat, S.J. and A.F. Ibraheem, 2013. An *In vitro* evaluation of fit of the crowns fabricated by Zirconium Oxide-based ceramic CAD/CAM systems, before and after porcelain firing cycles and after glaze cycles. J. Baghdad College Dent., 25: 43-48.

Hoosain, S.E., 2011. The sintering and optimization of stabilised Zirconia. Ph.D Thesis, University of the Witwatersrand, Johannesburg, South Africa.

Kalpakjian, S. and S. Schmid, 2013. Manufacturing Engineering and Technology. 7th Edn., Pearson, New Jersey, USA, ISBN:9780133151213, Pages: 1224.

Kandaswamy, R., 2010. Optimizing the mechanical properties of partially Yttria Atabilized Zirconia with Alumina additions. Master Thesis, Clemson University, Clemson, South Carolina.

Lawson, N.C., R. Bansal and J.O. Burgess, 2016. Wear, strength, modulus and hardness of CAD/CAM restorative materials. Dent. Mater., 32: e275-e283.

Lucas, T.J., 2015. Analysis of tetragonal to monoclinic phase transformation caused by accelerated artificial aging and the effects of microstructure in stabilized Zirconia. Ph.D Thesis, University of Alabama at Birmingham, Birmingham, Alabama.

Miyazaki, T. and Y. Hotta, 2011. CAD/CAM systems available for the fabrication of crown and bridge restorations. Aust. Dent. J., 56: 97-106.

Othmanm, A., 2017. The effect of Nano-Aluminum oxide on the mechanical and microstructure properties of 3-yttira tetragonal Zirconia polycrystal ceramic. MSc Thesis, University of Kufa, Kufa, Iraq.

Ozkurt, Z. and E. Kazazoglu, 2010. Clinical success of zirconia in dental applications. J. Prosthodontics Implant Esthetic Reconstructive Dent., 19: 64-68.

Rosentritt, M., 2008. A focus on Zirconia: An In-vitro lifetime prediction of Zirconia dental restorations. Ph.D Thesis, University of Amsterdam, Amsterdam, Netherlands.

Strub, J.R., E.D. Rekow and S. Witkowski, 2006. Computer-aided design and fabrication of dental restorations: Current systems and future possibilities. J. Am. Dent. Assoc., 137: 1289-1296.

Tinschert, J., G. Natt, N. Mohrbotter, H. Spiekermann and K.A. Schulze, 2007. Lifetime of Alumina- and Zirconia ceramics used for crown and bridge restorations. J. Biomed. Mater. Res. Part B. Appl. Biomater., 80: 317-321.

Uzun, G., 2008. An overview of dental CAD/CAM systems. Biotechnol. Equip., 22: 530-535.

Vythilingum, V., 2013. Zirconia nanocomposites for biomedical applications. Ph.D Thesis, University of the Witwatersrand, Johannesburg, South Africa.