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Comparison of the mechanical properties of polyetheretherketone and polyphenylene sulfone produced by fused filament fabrication and conventional manufacturing methods for dental applications: an in vitro study

Fatma Güneş[1*] and Turan Korkmaz[1]

Abstract

Background Additive manufacturing of thermoplastic polymers via fused filament fabrication (FFF) is gaining popularity because of its time and cost efficiency. Semicrystalline polyetheretherketone (PEEK) requires extensive postprocessing, whereas amorphous polyphenylene sulfone (PPSU) offers greater predictability and reliability. The purpose of this study was to compare the mechanical properties of PPSU, which is a new material for dental applications, and PEEK to evaluate the effects of different manufacturing methods on the mechanical properties of these materials.

Methods This study used a PEEK disc (Juvora Dental Disc, Juvora, UK) for subtractive CAD-CAM method-fabricated samples (PEEK-CG), a pure PEEK filament (Vestakeep i4 3DF filament, Evonik Industries AG, Germany) for FFF-fabricated samples (PEEK-3D), a pure PPSU filament (FIL-A GEHR, Gehr Plastics Inc., Germany) for FFF-fabricated PPSU samples (PPSU-3D), and a bulk PPSU material (Radel R-5000 NT, Solvay, France) for injection-molded samples (PPSU-INJ). A total of 80 samples ($n = 20$ per group) were tested. Half were evaluated for three-point bending ($n = 10$ per material), and the rest were evaluated for Vickers hardness ($n = 10$ per material). The Kruskal–Wallis H test was used for between-group comparisons, and the Mann–Whitney U test was used for pairwise comparisons ($\alpha = 0.05$).

Results PEEK-CG, with a flexural strength of 299.69 ± 12.26 MPa, a Young's modulus of 8.92 ± 0.28 GPa, a flexural stress at 3.5% strain of 6.97 ± 0.65 MPa, and a Vickers hardness of 40.02 ± 10.38 kg/mm^2, significantly outperformed PEEK-3D ($P < 0.05$). PPSU-INJ outperformed PPSU-3D in all tests, with significant differences only in Young's modulus ($P = 0.013$), with other differences not significant ($P > 0.05$). PEEK-3D outperformed PPSU-3D with a flexural strength

*Correspondence:
Fatma Güneş
fatmagullu@gazi.edu.tr

Full list of author information is available at the end of the article

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of 220.75 ± 12.07 MPa, a Young's modulus of 7.05 ± 0.94 GPa, a flexural stress at 3.5% strain of 3.94 ± 0.42 MPa, and a Vickers hardness of 21.48 ± 2.69 kg/mm^2 ($P < 0.05$).

Conclusions PPSU, with mechanical properties analogous to those of PEEK, meets the standards established for use in dental applications. These applications include its use as a fixed denture infrastructure, implant material, and removable denture clasps. The mechanical properties of 3D-printed PEEK and PPSU were found to be inferior to those of other methods, highlighting the need for improved additive manufacturing parameters.

Keywords High-performance polymer, Fused filament fabrication, Polyetheretherketone, Polyphenylene sulfone, Additive manufacturing

Background

Polymers can be classified into three distinct categories: engineering, commercial, and high-performance polymers. This classification is based on the degree of development of the polymers, which can be found in amorphous or crystalline forms [1]. PEEK, a semicrystalline high-performance polymer, is a member of the polyaryletherketone (PAEK) family. Compared with metal restorations, PEEK has several advantageous properties, including its lightness, modulus of elasticity similar to that of bone, capacity for shock absorption, resistance to corrosion, and low plaque retention potential [2–4]. PEEK has been widely applied in dental implants, as an abutment material and healing cap in implant-supported prostheses, as a precision connector in implant-supported overdentures, as a primary connector and clasp material in removable prostheses, and as an infrastructure material in fixed prostheses [2, 5, 6].

Thanks to its excellent mechanical strength, surface-modified PEEK implants have drawn attention as potential alternatives to titanium implants [2]. In addition to its esthetic advantages, one of the most significant benefits of PEEK is its physical properties, which closely resemble those of human bone [7]. Therefore, it has been suggested that PEEK appliances distribute forces more efficiently than acrylic and metal materials [7]. More recently, modified forms of PEEK with antimicrobial and bioactive properties have also been developed [8]. Considering these favorable characteristics, PEEK-based prosthetic appliances and obturators have been investigated as solutions to overcome the disadvantages of conventional prosthetic materials [9].

Unfilled PEEK has been reported to exhibit an elastic modulus of approximately 4 GPa, which is lower than that of dentin, but it demonstrates high compressive strength, withstanding plastic deformation up to approximately 1200 N [10]. These mechanical properties, combined with its biocompatibility and broad manufacturing and processing capabilities, make PEEK a suitable material for the production of customized post-core systems [10].

PPSU is a thermoplastic polymer new to dentistry. PPSU is a classification of high-performance polymers with an amorphous phase composition. This thermoplastic is highly resistant to water solubility [11]. Initial investigations [12–14] investigating the mechanical properties of this material revealed that it is promising for dental applications. Initial studies [12–14] have indicated that PPSU is a promising alternative to PEEK for use as a substructure material, as an implant material in single- or multiple-component fixed dentures, and as a clasp and substructure material in removable dentures. Amorphous thermoplastics exhibit low resistance to chemicals and fatigue [11]. Semicrystalline thermoplastics contain both amorphous and regular crystal structures. Compared with amorphous thermoplastics, semicrystalline thermoplastics exhibit improved resistance to chemicals, abrasion, and fatigue [1, 15]. Amorphous thermoplastics, owing to their transparent properties, provide better results in aesthetic restorations where translucency is important [11]. However, owing to the presence of crystals larger than the wavelength of light in their structure, semicrystalline thermoplastics do not transmit light [1] and are not suitable for monolithic use in aesthetic applications because of their opacity [11].

Thermoplastic polymers can be manufactured via conventional pressing methods or computer-aided manufacturing technologies. The conventional manufacturing processes are classified into three categories on the basis of the pressing method used: injection molding, extrusion, and compression molding [16]. In the context of computer-aided design and computer-aided manufacturing (CAD-CAM) technology, production relies on the "subtractive method," where the desired geometry is achieved by engraving a CAD-CAM disc. However, this approach is costly because of the excessive material waste involved. To address the need for cost reduction, additive manufacturing was developed [17]. This method allows for the production of restorations that mimic complex dental anatomy with less material than traditional methods do [17].

The use of 3D printing technology facilitates a wide range of dental applications, including surgical guides, dental models, custom implants, temporary restorations, partial and full dentures, maxillofacial prostheses, custom impression trays, occlusal splints, and orthodontic

appliances [18]. The broad spectrum of applications and practicality of this technology have led to its significant attention in the field of dentistry [18]. Additive manufacturing offers design flexibility, enabling rapid production of custom implants and restorations. While the subtractive method has higher material consumption, additive manufacturing uses materials efficiently, minimizing waste.

Currently, the most widely used additive manufacturing method for PEEK is FFF, which is a low-cost technology with material savings [19–22]. Since PEEK restorations require extensive posttreatment for marginal fit, surface quality, and long-term performance properties, the search for suitable materials for additive manufacturing continues [13]. PPSU material has been shown to meet the necessary criteria for additive manufacturing of dental restorations [13]. The manufacturing process for a three-dimensional (3D) printed object from a PPSU involves two distinct stages. Initially, the raw material undergoes extrusion into a filament, necessitating precise temperature regulation and predrying of the material. Second, the filament is produced by FFF, which carefully builds a 3D structure layer by layer through controlled heating and extrusion. In this process, the chosen printing parameters are critical. Studies [6, 7, 13, 14] investigating the additive manufacturing method of polymers have reported that parameters such as the nozzle temperature, plate temperature, layer thickness, printing speed, infill ratio, or raster angle strongly affect the mechanical and physical properties of the material [13, 22, 23].

The mechanical and optical properties of additive-manufactured PEEK materials are influenced by the amorphous-to-crystalline phase ratio [13]. The crystalline–amorphous phase cycle, which undergoes phase transformation during cooling and reaches a semicrystalline state after complete solidification, is highly dependent on adjusting the cooling parameters [22]. Rapid cooling inhibits the formation of crystal structures, whereas PPSU exists solely in an amorphous solid-state [12]. In contrast to the significant impact of the crystal phase ratio on the material properties of PEEK, PPSU, owing to its fully amorphous nature, offers enhanced predictability [22].

Flexural strength is defined as the resistance of a material to plastic deformation and fracture. In situations where a material is subjected to masticatory forces that may result in permanent deformation, a high flexural strength is advantageous [24]. Flexural stress at 3.5% strain is a measure of how much a material bends or flexes under an applied force. This measurement is used to determine the extent of stretching a material can undergo during bending and its ability to return to its original shape after the removal of the load. Additionally, the bending strength of a material can be calculated as the load capacity before failure. The elasticity modulus, which is defined as a material's resistance to deformation under stress, is another crucial mechanical property in the selection of dental implants and prosthetic infrastructure materials. PEEK, with a Young's modulus comparable to that of bone, is an ideal material for use as an implant because of its biocompatibility and optimal mechanical properties [25]. The wear resistance of a material is determined by its surface hardness, which is a key factor in its durability and performance under dynamic conditions [26]. In the context of PEEK utilization as an implant material, the surface properties and surface quality of the implant assume significant importance. The hardness of the surface layer is a crucial parameter for the wear resistance of prosthetic restorations [26].

The purpose of this study was to compare the mechanical properties of PPSU, which is a new material for dental applications, and PEEK to evaluate the effects of different manufacturing methods on the mechanical properties of these materials. The following null hypotheses were investigated: (1) the material composition used in the additive manufacturing method does not affect the mechanical properties tested, and the manufacturing method used does not affect the flexural strength, Young's modulus, or hardness values of (2) PEEK and (3) PPSU polymers.

Methods

Specimen design and preparation

The samples in this study were prepared according to ISO 10477:2020 standards, with a total of 80 samples (20 from each group) and dimensions of 25 × 2 × 2 mm. A total of 40 samples (10 from each group) were used for the three-point bending test ($N = 40$, $n = 10$), whereas the remaining 40 samples were allocated for the Vickers hardness test ($N = 40$, $n = 10$). The PEEK and PPSU materials used are listed in Table 1. The sample size was determined via power analysis via the mean and standard deviation from a previous study [13] with a margin of error (α) of 0.05, a test power (1-β) of 0.80, and an effect size (f) of 0.30. The analysis suggested that a minimum of 64 samples would suffice; however, 80 were included to increase the statistical power and reliability.

The PEEK and PPSU samples were designed via computer software (Materialise 3-matic, Materialise NV, Leuven, Belgium) and converted into Standard Tessellation Language (STL) file format (Fig. 1) [22] To mitigate the impact of moisture on additive manufacturing quality [15] the PEEK filament (Vestakeep i4 3DF, Evonik Industries AG, Essen, Germany) was dried for 48 h at 160 °C, and the PPSU filament (FIL-A GEHR, Gehr Plastics Inc., Mannheim, Germany) was dried for 72 h at 120 °C, accounting for their different drying needs. During

Table 1 Materials used

Abbreviation	Material	Manufacturer	Composition	Geometry	Color
PEEK-CG	Juvora Dental Disc	Juvora	PEEK	98 × 20 mm round bar	Natural
PEEK-3D	Vestakeep i4 3DF	Evonik Industries AG	PEEK	1.75 mm diameter filament	Natural
PPSU-3D	FIL-AGEHR PPSU	Gehr Plastics Inc	PPSU	1.75 mm diameter filament	Natural
PPSU-INJ	Radel R-5000 NT	Solvay	PPSU	60 × 60 × 2 mm plate	Translucent

PEEK-CG PEEK by subtractive CAD-CAM method, *PEEK-3D* PEEK by 3D printing filament, *PPSU-3D* PPSU by 3D printing filament, *PPSU-INJ* PPSU cut from bulk material, *mm* millimeter

printing, the filaments were kept in a sealed box with a desiccant to minimize moisture [22].

PEEK and PPSU samples were fabricated via an FFF method on a 3D printer (Intamsys Funmat Pro 410, Intamsys Technology GmbH, Eden Prairie, USA). PPSU samples were printed with a 0.4 mm nozzle diameter [12], a 130 °C print bed temperature, a 390 °C nozzle temperature, a 90 °C ambient temperature, a 0.1 mm film thickness, and a 10 mm/s printing speed. PEEK samples were prepared at 0.4 mm nozzle diameter [22], 150 °C print bed temperature, 90 °C ambient temperature, 440 °C nozzle temperature [22], 0.1 mm film thickness, and 10 mm/s print speed [22]. Printing parameters were determined within the manufacturer's specifications and the limitations of the equipment used in production.

For the subtractive method, a PEEK CAD-CAM disc (Juvora Dental Disc, Juvora Ltd., United Kingdom), which is a 98 × 20 mm circular disc in natural color, was used [12]. Prior to engraving, the samples were designed via a computer program (Coritec Icam v4.6 CAM, Imes-icore GmBH, Eiterfeld, Germany) and converted to the STL file format. PEEK-CG samples were fabricated into 20 pieces via a milling machine (Imes-icore GmbH, Eiterfeld, Germany). For the injection molding method, 20 samples were fabricated from PPSU translucent ready plates (Radel R-5000 NT, Solvay, France) with dimensions of 60 × 60 × 2 mm via the injection molding method via a wet cutting precision cutting device (Micracut, Metkon Instruments Inc., Bursa, Türkiye) [1].

Flexural strength, young's modulus, flexural stress at 3.5% strain measurements

Uniaxial flexural tests were conducted via a universal testing machine (Lloyd Instron, Lloyd UK Ltd., United Kingdom) in accordance with ISO 10477:2020. The samples were subjected to a three-point bending test [12]. All the samples were placed exactly at the center point on circular section steel bars with a distance of 20 mm between them, and a force was applied at a speed of 5 mm per minute from the center point of the support

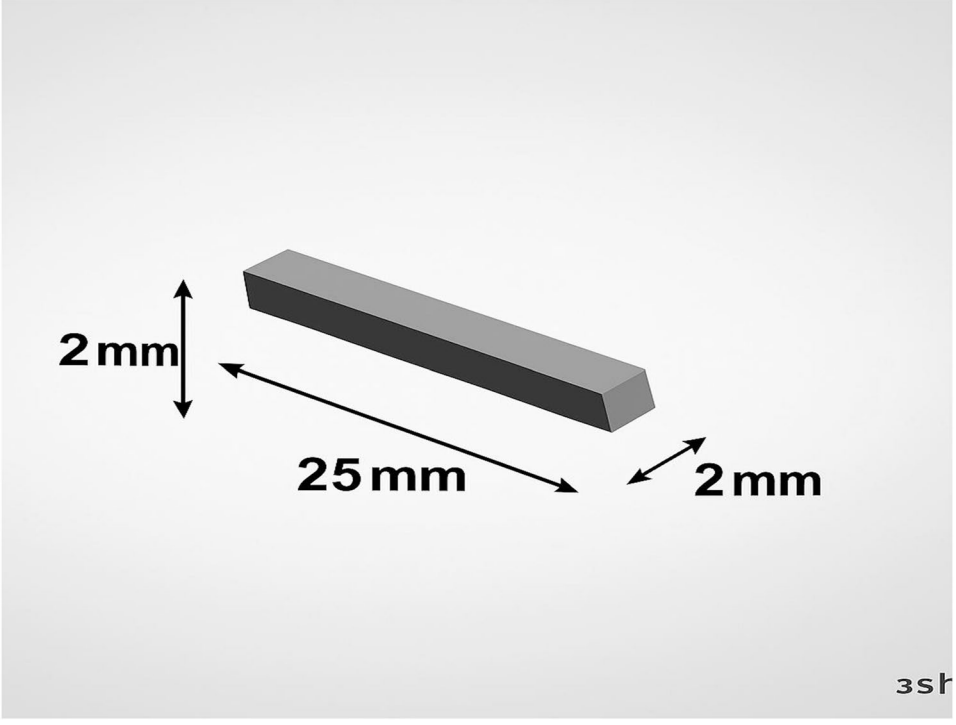


Fig. 1 CAD model of samples and 2D drawing of 3D printed samples

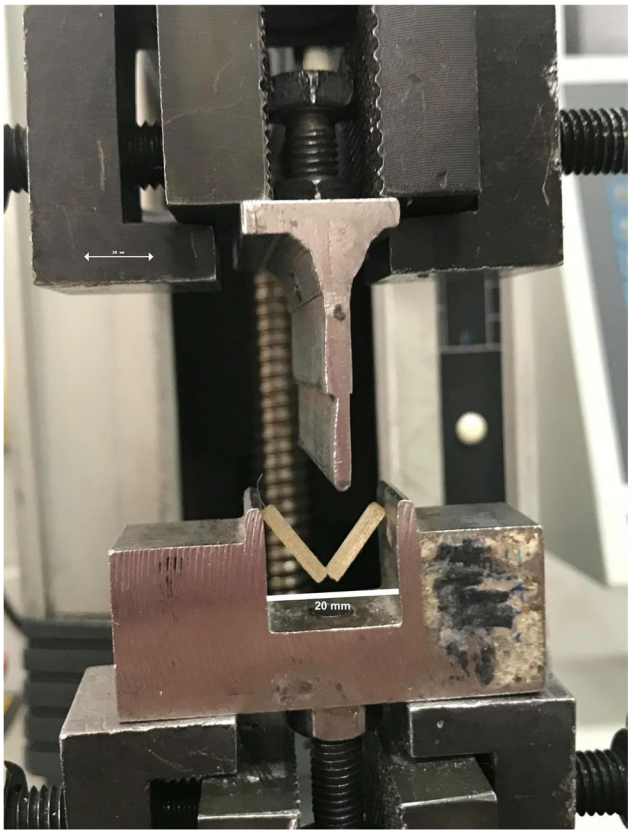


Fig. 2 Three-point bending test for the PPSU-3D group

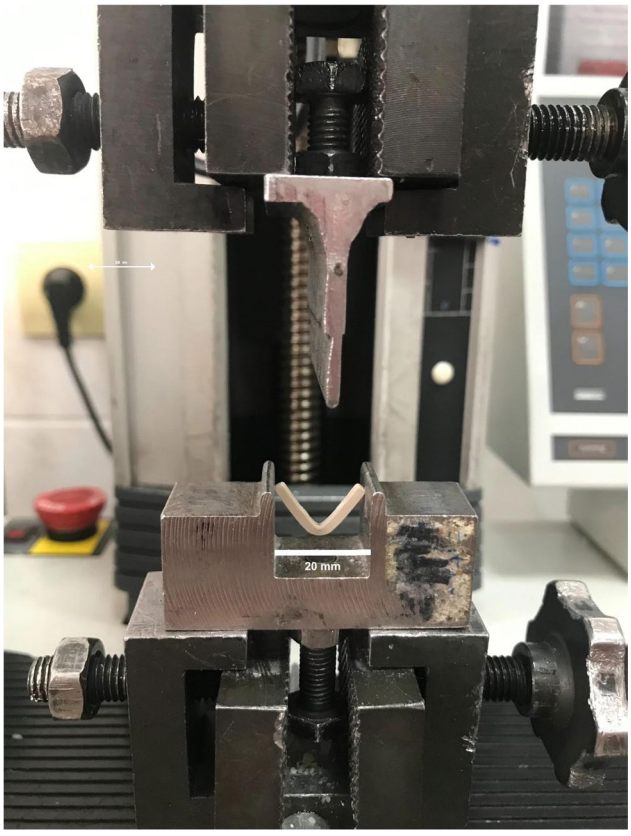


Fig. 3 Three-point bending test of the PEEK-3D group

bars with a circular section steel bar perpendicular to the long axis of the sample (Figs. 2 and 3). The force applied at the moment of deformation was recorded in Newtons (N) by the computer. The data obtained in N were converted to units of megapascal (MPa) by calculating the flexural strength with the formula regulated by the ISO 10477:2020 standard. According to the ISO 178:2003 standard, the flexural properties of rigid and semi-rigid plastics can be determined under defined conditions. This standard allows the calculation of the Young's modulus when appropriate. Additionally, if a test specimen does not break, the flexural strength can be determined by measuring the flexural stress at 3.5% strain. Based on this standard, Young's modulus, flexural stress at 3.5% strain, and flexural strength were calculated using the three-point bending test in this study.

Vickers hardness values measurement

Prior to the Vickers hardness test, the samples were ground and polished to enhance the visibility of the notch edges under a light microscope and to create a smooth and shiny application surface (Metkon Sanding and Polishing Device, Metkon, Bursa, Türkiye). The samples were cleaned with distilled water and ethanol and then promptly dried with compressed air. A force of 300 g was applied to each sample from three different regions via a 136° pyramidal diamond tip for 15 s. (Micro Vickers hardness tester HVS 1000 M, Accud, Vienna, Austria) [25]. The lengths of the notches, formed by the applied force, were measured in microns under a light microscope [26]. Measurements were made for each notch, and the average of the three values obtained was calculated as the Vickers hardness value [26].

Data analysis

Statistical evaluations of the test data in the study were performed via the Statistical Package for the Social Sciences (SPSS) statistical program (IBM SPSS, SPSS Inc.). The Kolmogorov–Smirnov test was used to assess the normality assumption, and the Levene test was used to assess the homogeneity of variance assumption. The Kruskal–Wallis H test was used to examine differences between more than 2 groups. The Mann–Whitney U test was applied to paired groups to determine the source of the difference in groups where there was a difference. ($\alpha = 0.05$)

Results

The normality assumption was tested with the Kolmogorov–Smirnov test, and the values did not conform to a normal distribution. The assumption of homogeneity of variance was evaluated via Levene's test. The

analysis revealed that the variances were not homogeneous ($P < 0.05$).

Bending strength

The standard deviations and means of the flexural strength values of the test groups are shown in Table 2 (Fig. 4). The Kruskal–Wallis test was used to determine if there was a significant difference between all groups. The results of this analysis revealed a statistically significant difference between the groups ($P = 0.000$) (Table 2). In groups where there was a difference, the paired Mann–Whitney U test was used to determine the source of the difference. A statistically significant difference was found between the PEEK-3D and PPSU-3D groups and between the PEEK-CG and PEEK-3D groups ($P = 0.000$) (Table 3). According to the results of the Mann–Whitney U test performed to compare the flexural strength values of the PPSU-3D and PPSU-INJ groups, no statistically significant difference was found between the groups ($P = 0.705$) (Table 3).

The flexural strength of the experimental samples was qualitatively evaluated to elucidate their failure mechanisms. In the PEEK-CG samples, fractures propagated along regular and parallel lines, indicative of intermaterial delamination (separation between layers). In contrast, fractures in 3D-printed PEEK and PPSU samples generally exhibited properties of brittle fracture, characterized by sharp and irregular fracture surfaces. The PPSU-INJ samples had more homogeneous and smooth fracture surfaces.

Young's modulus values

The means and standard deviations of the Young's modulus values of the test groups are shown in Table 2 (Fig. 5). According to the results of the Kruskal–Wallis test performed between groups to compare the Young's modulus values of all groups, a statistically significant difference was found ($P = 0.000$) (Table 2). To determine the source of the difference in the groups where there was a difference, the Mann–Whitney U test was applied to pairs. According to the results of the Mann–Whitney U test, a statistically significant difference was found between the PEEK-3D and PPSU-3D groups ($P = 0.000$), between the PEEK-CG and PEEK-3D groups ($P = 0.000$), and between the PPSU-3D and PPSU-INJ groups ($P = 0.013$) (Table 3).

Flexural stress at 3.5% strain values

The means and standard deviations of the flexural stress at 3.5% strain values for the test groups are presented in Table 2. A Kruskal–Wallis test was conducted to compare the flexural stress at 3.5% strain values across all groups, revealing a statistically significant difference ($P = 0.000$), as shown in Table 2. To identify the sources of these differences, the Mann–Whitney U test was performed

Table 2 Descriptive statistical test values and kruskal–wallis H test results of the bending strength values, flexural stress at 3.5% strain values, young's modulus values and Vickers hardness values for all the groups

	Bending Strength (MPa)				Young's Modulus (GPa)				Flexural Stress at 3.5% Strain (MPa)				Vickers Hardness (kg/mm²)			
	N	Mean ± SD	Kruskal–Wallis H	df	*P* value	Mean ± SD	Kruskal–Wallis H	df	*P* value	Mean ± SD	Kruskal–Wallis H	df	*P* value	Mean ± SD	Kruskal–Wallis H	df	*P* value
PEEK-CG	10	299.69 ± 12.26	32.96	3	0.000	8.92 ± 0.28	33.546	3	0.000	6.97 ± 0.65	27.79	3	0.000	40.02 ± 10.38	21.98	3	0.000
PEEK-3D	10	220.75 ± 12.07				7.05 ± 0.94				3.94 ± 0.42				21.48 ± 2.69			
PPSU-3D	10	180.80 ± 6.08				4.99 ± 0.38				3.11 ± 0.41				38.40 ± 2.69			
PPSU-INJ	10	183.83 ± 6.85				5.53 ± 0.47				3.53 ± 0.47				39.96 ± 3.09			

N number of samples, *SD* standard deviation, *RA* rank average, *df* degrees of freedom, *MPa* megapascal, *GPA* gigapascal, *kg* kilogram, *mm* millimeter

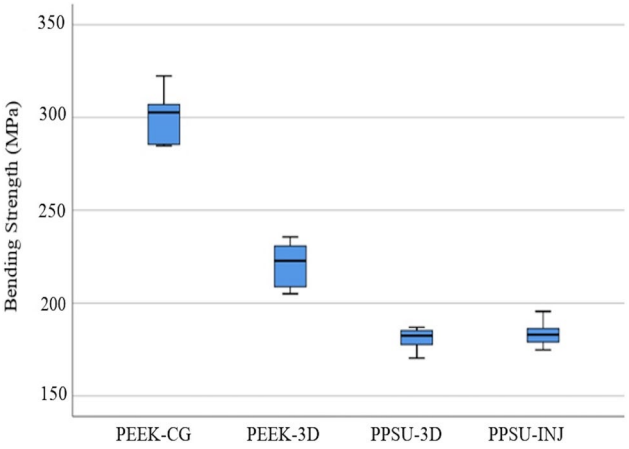


Fig. 4 Box plot of bending strengths for PEEK and PPSU across manufacturing methods

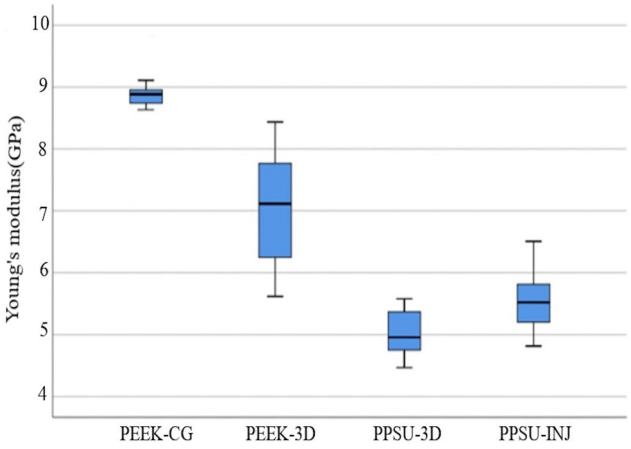


Fig. 5 Box plot of Young's modulus for PEEK and PPSU across manufacturing methods

pairwise. According to the results of the Mann–Whitney U test, there was a statistically significant difference between the PEEK-3D and PPSU-3D groups ($P = 0.002$) and between the PEEK-CG and PEEK-3D groups ($P = 0.000$). However, no significant difference was found between the PPSU-3D and PPSU-INJ groups ($P = 0.053$) (Table 3).

Hardness values

The means and standard deviations of the Vickers hardness values of the test groups are shown in Table 2 (Fig. 6). According to the results of the Kruskal–Wallis test performed to compare the Vickers hardness values of all the groups, a statistically significant difference was found ($P = 0.000$) (Table 2). The paired Mann–Whitney U test was used to determine the source of the difference in the groups where there was a difference. According to the results of the Mann–Whitney U test, a statistically

Table 3 Mann–Whitney U test results of the bending strength (MPa), flexural stress at 3.5% strain (MPa), young's modulus (GPa) and Vickers hardness (kg/mm^2) values of the two groups

Property		*N*	RA	SD	Mann–Whitney U	Z	Relative Difference (%)	*p* value
Bending strength	PEEK-CG	10	15.5	12.26	0.000	−3.780	35.79	0.000
	PEEK-3D	10	5.5	12.07				
	PPSU-3D	10	10	6.08	45.00	−0.378	−1.65	0.705
	PPSU-INJ	10	11	6.85				
	PEEK-3D	10	15.5	12.07	0.000	−3.780	22.12	0.000
	PPSU-3D	10	5.5	6.08				
Flexural Stress at 3.5% Strain	PEEK-CG	10	6.9	0.65	100.0	−3.779	77.16	0.000
	PEEK-3D	10	3.9	0.42				
	PPSU-3D	10	3.5	0.47	76.0	−1.965	−11.90	0.053
	PPSU-INJ	10	3.1	0.41				
	PEEK-3D	10	3.9	0.42	90.0	−3.023	26.69	0.002
	PPSU-3D	10	3.1	0.41				
Young's modulus	PEEK-CG	10	15.5	0.28	0.000	−3.780	26.52	0.000
	PEEK-3D	10	5.5	0.94				
	PPSU-3D	10	7.2	0.38	17.000	−2.495	−9.76	0.013
	PPSU-INJ	10	13.8	0.47				
	PEEK-3D	10	15.5	0.94	0.000	−3.780	41.28	0.000
	PPSU-3D	10	5.5	0.38				
Vickers hardness	PEEK-CG	10	15.3	10.384	2.000	−3.628	86.43	0.000
	PEEK-3D	10	5.7	2.690				
	PPSU-3D	10	9.1	2.694	36.000	−1.058	−3.90	0.290
	PPSU-INJ	10	11.9	3.096				
	PEEK-3D	10	15.5	2.690	0.000	−3.780	−44.06	0.000
	PPSU-3D	10	5.5	2.694				

N number of samples, *RA* rank average, *SD* standard deviation

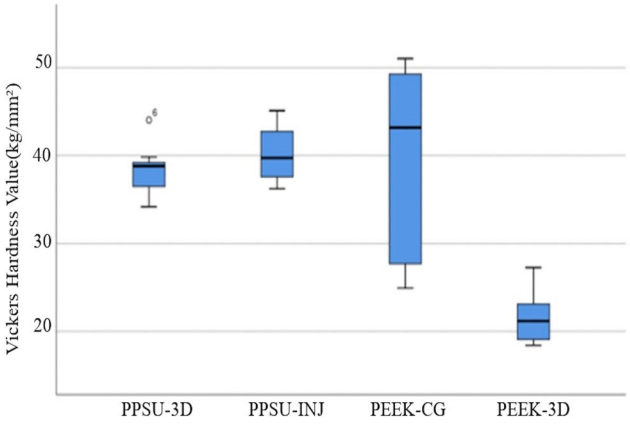


Fig. 6 Box plot of the Vickers hardness values for PEEK and PPSU materials across manufacturing methods

significant difference was found between the PEEK-3D and PPSU-3D groups ($P = 0.000$) and between the PEEK-CG and PEEK-3D groups ($P = 0.000$) (Table 3). According to the results of the Mann−Whitney test performed to compare the Vickers hardness values of the PPSU-3D and PPSU-INJ groups, no statistically significant difference was found between the groups ($P = 0.290$) (Table 3).

Discussion

This study aimed to evaluate the effect of the manufacturing method on the mechanical properties of PEEK and PPSU polymeric materials and to evaluate the effect of the polymeric material composition on the tested parameters in the additive manufacturing process. The first null hypothesis, which stated that the material composition used in the additive manufacturing method would not affect the tested mechanical properties, was rejected. The second null hypothesis that the manufacturing method used would not affect the flexural strength, Young's modulus, or hardness values for PEEK was rejected, whereas the third null hypothesis for the PPSU polymer was partially rejected. Although PEEK-CG and PEEK-3D are produced from the same core material, PEEK-CG exhibited superior flexural strength, indicating that the manufacturing process plays a significant role.

The additive manufacturing of high-performance polymers, particularly PEEK, has recently received significant attention for medical implant applications [5]. The bioactivity and biomechanical properties of personalized PEEK implants produced via additive methods are critical to their overall performance [5]. The biomechanical properties of high-performance polymers produced by additive CAD-CAM techniques have been a focus of recent research. The melting point of a material also plays a critical role in additive manufacturing. PEEK, widely known for its successful application in implant manufacturing, has a melting temperature of 343 °C, which must be achieved during the extrusion process. PPSU, a material with a lower melting temperature that is more cost-effective and practical for printing, has been identified as a suitable alternative to PEEK [12]. Among additive manufacturing methods, FFF is currently the most widely utilized for PEEK because of its low cost and efficient material usage [19]. Owing to their advantages, PEEK-3D and PPSU-3D groups were fabricated via the FFF method in this study.

Schönhoff et al. [13] evaluated the tensile bond strength of PPSU and PEEK materials fabricated by FFF with the composite resin used to fabricate the superstructure. Their findings revealed that the tensile bond strength between PPSU and the superstructure composite resin was comparable to that of PEEK. These results indicate that PPSU, which has mechanical properties similar to those of PEEK, is suitable for dental treatments.

Schönhoff et al. [12] evaluated the flexural strength of samples via a three-point bending test after applying thermal cycling. The samples included those fabricated from PPSU filaments via the FFF method (Fil-A Gehr PPSU, Gehr Plastics Inc., Mannheim, Germany), samples cut from PPSU bulk material manufactured via injection molding (Radel R-5000 NT, Solvay, France), and PEEK samples fabricated via the subtractive CAD-CAM method (Juvora Dental Disc, Juvora Ltd., United Kingdom). The resistance and hardness values were compared. The PPSU samples produced via the FFF method were fabricated with a nozzle diameter of 0.4 mm, a nozzle temperature of 410 °C, a print bed temperature of 110 °C, a layer thickness of 0.15 mm, and a print speed of 400 mm/min. Schönhoff et al. [12] reported a maximum flexural strength of 169.4 MPa for samples cut from PPSU bulk material manufactured via injection molding, 94.1 MPa for PPSU samples produced via the FFF method, and 222.9 MPa for PEEK samples fabricated via the subtractive CAD-CAM method. It was concluded that PPSU samples cut from bulk material after thermal cycling and PEEK samples fabricated via the subtractive CAD-CAM method presented the highest flexural strength values, whereas PPSU samples fabricated via the FFF method presented the lowest values.

In line with the findings of Schönhoff et al. [12], PPSU samples fabricated by the FFF method in this study had lower values than those cut from injection-molded bulk material. For PPSU-3D, certain steps in the 3D manufacturing process may weaken the material. This observation suggests that the additive manufacturing parameters used for the additive manufacturing of PPSU samples require further optimization. The observed differences in the mechanical properties may be attributed to variations in the printing parameters. Future research is needed to develop optimized printing parameters that will allow PPSU samples produced by FFF to exhibit mechanical

properties comparable to those produced by injection molding.

The 3D printing parameters greatly affect the microstructure of a material [15, 22, 27]. Yang et al. [27] reported that the nozzle temperature strongly affects the crystallinity and mechanical properties of PEEK and that the nozzle temperature affects the crystal melting process, the crystallization process, the interface between the printing lines, and the degradation phenomenon of polymer materials.

Zanzanjiam et al. [15] reported that low ambient temperature causes rapid cooling of PEEK chains, resulting in defective crystallization and significant bending failure. Higher temperatures can help slow the crystallization phenomenon and reduce internal stresses [28]. In fact, at high ambient temperatures, the polymer chains have enough time and energy to crystallize, so PEEK can experience a more perfect crystallization process and thus achieve a higher degree of crystallinity [15]. These results indicate that the crystallinity and hardness/toughness of PEEK can be adjusted by selecting an optimum ambient temperature [15].

The voids and pores inherent to the buildup of the material in FFF are still issues [29]. These voids and pores are largely unavoidable and can amount to up to 8% in volume even for 100% filled PEEK prints [29]. By adjusting these parameters, it is possible to produce denser PEEK prints and further refine their mechanical specifications [29]. The density of PEEK also increases with increasing nozzle temperature [29]. This is due to the improved fluidity of PEEK, which facilitates the filling of voids and pores [30]. Higher nozzle temperatures reduce the number and size of voids. The results of some studies [27, 30, 31] have shown that PEEK has higher crystallinity when printed at higher nozzle temperatures.

Wang et al. [22] reported that the most effective parameter among the three printing parameters (printing speed, nozzle temperature, and nozzle diameter) for determining the mechanical properties of PEEK samples produced via the FFF method was the nozzle diameter, followed by the printing speed and nozzle temperature. To obtain samples with the best flexural strength, the combination of a 0.4 mm nozzle diameter, a 430 °C nozzle temperature, and a 5 mm/s printing speed are the most suitable values. In the study by Wang et al. [22], a maximum flexural strength of 193.33 ± 7.04 MPa was obtained using a 0.4 mm nozzle with a nozzle temperature of 430 °C and a printing speed of 10 mm/s. In this study, a mean flexural strength of 220.75 ± 12.07 MPa (maximum 235.606 ± 12.07 MPa) was obtained by using a 0.4 mm nozzle with a nozzle temperature of 440 °C and a printing speed of 10 mm/s. The difference between the values may be due to the use of a higher nozzle temperature in this study.

It has been observed that producing at lower speeds allows the extruded material to remain exposed to the heat of the die for longer, sufficiently melting the material and resulting in lower viscosity [13]. For PEEK samples produced by the additive method at 440 °C, an increase in extrusion speed was found to result in higher Young's modulus values, which may be related to the high viscosity of the PEEK material. Wang et al. [22] obtained a maximum Young's modulus of 2.193 ± 0.058 GPa at a nozzle temperature of 420 °C, a nozzle diameter of 0.6 mm, and a printing speed of 10 mm/s. In this study, a mean Young's modulus value of 7.05 ± 0.94 GPa was obtained with a nozzle temperature of 440 °C, a nozzle diameter of 0.4 mm, and a printing speed of 10 mm/s. According to Wang et al. [22], the higher values found in this study can be attributed to an increase in the Young's modulus of high-viscosity PEEK due to the high printing speed at high nozzle temperatures.

In accordance with ISO 1567:2001, the required Young's modulus for denture base materials is specified as 2000 MPa. In this study, the Young's modulus values obtained for all groups exceeded the minimum threshold defined by the ISO standard. These findings suggest that PEEK and PPSU materials possess sufficient mechanical properties to be considered suitable for use as denture base materials. The results obtained in this study are supported by previous studies [7, 32] in the literature that have examined the use of PEEK as a denture base material.

In the study by Kassem et al. [26], the Vickers microhardness of PEEK samples fabricated from CAD-CAM blocks via the subtractive method was evaluated. The mean Vickers microhardness of PEEK was reported to be 31.55 ± 2.67 kg/mm². In this study, the mean Vickers hardness value for the PEEK-CG group was 40.02 ± 10.38 kg/mm². The differences between these values may be attributed to variations in the blocks, as they belong to different brands. No studies in the literature have examined the Vickers hardness values of PPSU materials.

Mayinger et al. [23] investigated the chemical and mechanical properties of PPSU on the basis of its composition and production methods. This research provides the first demonstration of successful doping of PPSU with silver-coated zeolite fillers. The results indicated that 3D-printed objects presented the highest distribution of silver ions. Furthermore, it was confirmed that silver-coated zeolite-filled PPSU restorations could be effectively coated and bonded to achieve aesthetic results. The study emphasized that the combination of additive manufacturing with an antimicrobial thermoplastic presents significant opportunities in prosthetic dentistry. This material was identified as having potential applications in

removable dentures (such as clasps), temporary or permanent fixed dentures, and implant abutments.

PEEK and PPSU, attractive polymers for dental prosthetics, offer high strength and chemical resistance [7, 9]. However, these polymers are limited in their ability to mimic natural tooth color when used without a veneer material and may have difficulty with color stability over long-term use [33, 34]. This poses a significant obstacle in aesthetic dentistry applications. The inert surface of PEEK makes bonding with composite resins and abutment teeth difficult, limiting its clinical application [34].

As in this study, previous studies [21, 27, 35] clearly demonstrated that precise temperature and pressure settings are essential during 3D production of PEEK and PPSU. Small deviations in production parameters can adversely affect the mechanical properties of the material, potentially leading to inadequate capacity to withstand load-bearing stresses. Consequently, the lifespan of a material may be reduced. The challenges associated with the utilization of PEEK and PPSU materials in dental applications primarily stem from the elevated costs of material processing and the requirement for specialized equipment in additive manufacturing.

While numerous studies have explored the fabrication of PEEK materials via the FFF method, a consensus on the most appropriate parameters for additive manufacturing of PEEK has yet to be established. Consequently, the additive manufacturing parameters employed in this study provide valuable data to help determine optimal parameters for producing small PEEK samples, such as the personalized dental implants previously investigated.

This study's limitations include not investigating the biocompatibility of PPSU—a material newly introduced for dental treatments—and focusing only on a few mechanical properties. PPSU is known for its high water resistance, and future research is needed to evaluate the changes in surface properties after thermal cycling. Thermal cycling, which simulates fluctuations in intraoral temperature and prolonged exposure to saliva, is essential for understanding the material's performance and durability in oral environments. Future research should prioritize these aspects to enhance the understanding of PPSU suitability for dental applications. Additionally, future studies should explore other mechanical properties and the biocompatibility of PPSU.

On the basis of the present findings, the choice of material and manufacturing process plays a significant role in determining mechanical properties. While this study demonstrates the potential for PPSU to achieve successful long-term clinical use in both removable and fixed dental prostheses, it remains uncertain whether optimization of printing parameters in additive manufacturing can yield properties comparable to those of conventionally fabricated samples. Future in vivo and in vitro studies are needed to provide a deeper understanding of the clinical effects of PPSU in dental treatments.

Conclusions

The results of the three-point bending strength and Vickers hardness tests revealed that PEEK samples fabricated through the subtractive CAD-CAM method had superior mechanical properties compared with those made via the additive CAD-CAM method. Similarly, PPSU samples from injection-molded, ready-made plates presented higher values than those produced by the additive CAD-CAM method. The assessment of three-point bending strength and Vickers hardness indicated that the performance of the samples made through additive processes was generally lower than that of the samples produced via traditional methods for both materials. These results emphasize the need to refine printing parameters in the additive manufacturing process to improve output quality. Given that PPSU demonstrates mechanical properties sufficient for dental applications and comparable to those of PEEK, PPSU appears to be a suitable alternative for both removable and fixed dentures.

Abbreviations

PPSU	Polyphenyl sulfone
PEEK	Polyetheretherketone
PAEK	Polyaryl ether ketone
CAD	Computer-aided Design
CAM	Computer-aided manufacturing
FFF	Fused filament fabrication
3D	Three-dimensional
PEEK-CG	PEEK by the subtractive CAD-CAM method
PEEK-3D	PEEK by 3D printing filaments
PPSU-3D	PPSU with 3D printing filaments
PPSU-INJ	PPSU cut from bulk material
MPa	Megapascal
GPa	Gigapascal
kg	Kilogram
mm	Millimeter
N	Number of samples
SD	Standard deviation
RA	Rank average
df	Degrees of freedom

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FG contributed to conceptualization, methodology, writing of the original draft, review and editing, and data collection. TK contributed to writing of original draft, methodology, supervision, and funding acquisition. All authors read and approved the final manuscript.

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Data availability

Datasets analyzed during the current study are available from the corresponding author on reasonable request. Data is provided within the manuscript or supplementary information files.

Declarations

Ethics approval and consent to participate
Not applicable.

Consent for publication
Not applicable.

Competing interests
The authors declare no competing interests.

Author details
[1]Faculty of Dentistry, Department of Prosthodontics, Gazi University, Emek mahallesi, Bişkek caddesi 1. Sokak No:6, Ankara 06490, Turkey

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References

1. Simsiriwong J, Shrestha R, Shamsaei N, Lugo M, Moser RD. Effects of microstructural inclusions on fatigue life of polyether ether ketone (PEEK). J Mech Behav Biomed Mater. 2015;51:388–97. https://doi.org/10.1016/j.jmbbm.2015.07.020.
2. Najeeb S, Zafar MS, Khurshid Z, Siddiqui F. Applications of polyetheretherketone (PEEK) in oral implantology and prosthodontics. J Prosthodont Res. 2016;60:12–9. https://doi.org/10.1016/j.jpor.2015.10.001.
3. Stawarczyk B, Thrun H, Eichberger M, Roos M, Edelhoff D, Schweiger J, et al. Effect of different surface pretreatments and adhesives on the load-bearing capacity of veneered 3-unit PEEK FDPs. J Prosthet Dent. 2015;114:666–73. https://doi.org/10.1016/j.prosdent.2015.06.006.
4. Schwitalla AD, Spintig T, Kallage I, Müller WD. Flexural behavior of PEEK materials for dental application. Dent Mater J. 2015;31:1377–84. https://doi.org/10.1016/J.DENTAL.2015.08.151.
5. Al Maruf DSA, Ren J, Cheng K, Xin H, Lewin W, Pickering E, et al. Evaluation of osseointegration of plasma treated polyaryletherketone maxillofacial implants. Sci Rep. 2025;15:1895. https://doi.org/10.1038/s41598-024-80335-z.
6. Diken Türksayar AA, Petersmann S, Spintzyk S. The effect of thermomechanical aging on the fracture resistance of additively and subtractively manufactured polyetheretherketone abutments. J Dent. 2024;149:105225. https://doi.org/10.1016/j.jdent.2024.105225.
7. Khurshid Z, Nedumgottil BM, Ali RMM, Bencharit S, Najeeb S. Insufficient evidence to ascertain the long-term survival of PEEK dental prostheses: A systematic review of clinical studies. Polymers. 2022;14:2441. https://doi.org/10.3390/polym14122441.
8. Najeeb S, Khurshid Z, Matinlinna JP, Siddiqui F, Nassani MZ, Baroudi K. Nanomodified PEEK dental implants: bioactive composites and surface modification-A review. Int J Dent. 2015;2015:381759. https://doi.org/10.1155/2015/381759.
9. Villefort RF, Tribst JPM, Dal Piva AMO, Binda NC, Ferreira CEA, et al. Stress distribution on different bar materials in implant-retained palatal obturator. PLoS ONE. 2020;1510:e0241589. https://doi.org/10.1371/journal.pone.0241589.
10. Direk A, Tekin S, Khurshid Z. Fracture strength of CAD-CAM milled polyetheretherketone (PEEK) post-cores vs conventional post-cores; an in vitro study. PeerJ. 2024;12:e18012. https://doi.org/10.7717/peerj.18012.
11. Dorf T, Perkowska K, Janiszewska M, Ferrer I, Ciurana J. Effect of the main process parameters on the mechanical strength of polyphenylsulfone (PPSU) in ultrasonic micromolding process. Ultrason Sonochem. 2018;46:46–58. https://doi.org/10.1016/J.ULTSONCH.2018.03.024.
12. Schönhoff LM, Mayinger F, Eichberger M, Reznikova E, Stawarczyk B. 3D printing of dental restorations: mechanical properties of thermoplastic polymer materials. J Mech Behav Biomed Mater. 2021;119:104544. https://doi.org/10.1016/j.jmbbm.2021.104544.
13. Schönhoff LM, Mayinger F, Eichberger M, Lösch A, Chem D, Reznikova E, et al. Three-dimensionally printed and milled polyphenylene sulfone materials in dentistry: tensile bond strength to veneering composite resin and surface properties after different pretreatments. J Prosthet Dent. 2022;128:93–9. https://doi.org/10.1016/j.prosdent.2020.12.042.
14. Tammam R. Comparison in vitro between three-dimensionally printed, milled CAD-CAM and manually fabricated interim crown materials. Al-Azhar Assiut Dent J. 2021;4:141–50. https://doi.org/10.21608/aadj.2021.208206.
15. Zanjanijam AR, Major I, Lyons JG, Lafont U, Devine DM. Fused filament fabrication of PEEK: A review of process-structure-property relationships. Polymers. 2020;12:1665. https://doi.org/10.3390/POLYM12081665.
16. Kurtz SM. Synthesis and processing of PEEK for surgical implants, in: PEEK biomaterials handbook. 2nd ed. Amsterdam: Elsevier; 2012.
17. Yavuz E, Yilmaz S. Diş Hekimliğinde Yeni ve Hızla Ilerleyen üretim teknolojisi: 3 Boyutlu Yazıcılar (New and rapidly advancing production technology in dentistry:3D printers). Akdeniz Med J. 2021;12:197–205. https://doi.org/10.53394/akd.958759.
18. Sulaiman TA. Materials in digital dentistry-a review. J Esthet Restor Dent. 2020;32:171–81. https://doi.org/10.1111/jerd.12566.
19. Turner N, Strong B, Gold RA. A review of melt extrusion additive manufacturing processes: I. Process design and modeling. Rapid Prototyp J. 2014;20:192–204. https://doi.org/10.1108/RPJ-01-2013-0012.
20. Rahman KM, Letcher T, Reese R. Mechanical properties of additively manufactured PEEK components using fused filament fabrication. Proceedings of the ASME 2015 International Mechanical Engineering Congress and Exposition. Volume 2A: Advanced Manufacturing. Houston, Texas, USA. 2015. https://doi.org/10.1115/IMECE2015-52209
21. Dua R, Rashad Z, Spears J, Dunn G, Maxwell M. Applications of 3D-printed PEEK via fused filament fabrication: A systematic review. Polymers. 2021;13(22). https://doi.org/10.3390/polym13224046.
22. Wang Y, Müller WD, Rumjahn A, Schmidt F, Schwitalla AD. Mechanical properties of fused filament fabricated PEEK for biomedical applications depending on additive manufacturing parameters. J Mech Behav Biomed Mater. 2021;1:115. https://doi.org/10.1016/j.jmbbm.2020.104250.
23. Mayinger F, Lösch A, Reznikova E, Wilhelm C, Stawarczyk B. Influence of silver coated zeolite fillers on the chemical and mechanical properties of 3D-printed polyphenylene sulfone restorations. J Mech Behav Biomed Mater. 2024;160:106756. https://doi.org/10.1016/j.jmbbm.2024.106756.
24. Shrivastava SP, Dable R, Raj AN, Mutneja P, Srivastava SB, Haque M, et al. Comparison of mechanical properties of PEEK and PMMA: an in vitro study. J Contemp Dent Pract. 2021;22:179–83. https://doi.org/10.5005/jp-journals10024-3077.
25. Da Cruz MB, Marques JF, Peñarrieta-Juanito GM, Costa M, Souza JCM, Magini RS, et al. Bioactive-enhanced polyetheretherketone dental implant materials: mechanical characterization and cellular responses. J Oral Implantol. 2021;47:9–17. https://doi.org/10.1563/aaid-joi-d-19-00172.
26. Kassem YM, Alshimy AM, El-Shabrawy SM. Mechanical evaluation of polyetheretherketone versus zirconia. Alexandria Dent J. 2019;44:61–6.
27. Yang C, Tian X, Li D, Cao Y, Zhao F, Shi C. Influence of thermal processing conditions in 3D printing on the crystallinity and mechanical properties of PEEK material. J Mater Process Technol. 2017;248:1–7. https://doi.org/10.1016/J.JMATPROTEC.2017.04.027.
28. Wu WZ, Geng P, Zhao J, Zhang Y, Rosen DW, Zhang HB. Manufacture and thermal deformation analysis of semicrystalline polymer polyether ether ketone by 3D printing. Mater Res Innovations. 2014;18(5–12):5–16. https://doi.org/10.1179/1432891714Z.000000000898.
29. Rendas P, Figueiredo L, Machado C, Mourão A, Vidal C, Soares B. Mechanical performance and bioactivation of 3D-printed PEEK for high-performance implant manufacture: A review. Prog Biomater. 2023;12:89–111. https://doi.org/10.1007/s40204-022-00214-6.
30. Wang P, Zou B, Xiao H, Ding S, Huang C. Effects of printing parameters of fused deposition modeling on mechanical properties, surface quality, and microstructure of PEEK. J Mater Process Technol. 2019;271:62–74. https://doi.org/10.1016/j.jmatprotec.2019.03.016.
31. Zhao Y, Zhao K, Li Y, Chen F. Mechanical characterization of biocompatible PEEK by FDM. J Manuf Process. 2020;56:28–42. https://doi.org/10.1016/J.JMAPRO.2020.04.063.
32. Muhsin SA, Hatton PV, Johnson A, Sereno N, Wood DJ. Determination of polyetheretherketone (PEEK) mechanical properties as a denture material. Saudi Dent J. 2019;31:382–91. https://doi.org/10.1016/j.sdentj.2019.03.005.
33. Saraswathy A, Latha N, Ayyadanveettil P, Thavakkara V. Comparative evaluation of surface roughness and color stability of polyetheretherketone with conventional interim prosthetic materials: an in vitro study. J Contemp Dent Pract. 2024;25:930–5. https://doi.org/10.1016/j.jmapro.2020.04.063.
34. Luo C, Liu Y, Peng B, Chen M, Liu Z, Li Z, Kuang H, Gong B, Li Z, Sun H. PEEK for oral applications: recent advances in mechanical and adhesive properties. Polym (Basel). 2023;15:386. https://doi.org/10.3390/polym15020386.

35. Tasopoulos T, Chatziemmanouil D, Kouveliotis G, Karaiskou G, Wang J, Zoidis P. PEEK maxillary obturator prosthesis fabrication using intraoral scanning, 3d printing, and CAD/CAM. Int J Prosthodont. 2020;33:333–340 https://doi.org/10.11607/ijp.6575. PMID: 32320188.

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