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LOAD BEARING OF DIFFERENT FIXED DENTAL PROSTHESIS WITH VARIOUS CANTILEVER LENGTHS

DOCTOR OF PHILOSOPHY THESIS

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DECLARATION

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İlayda Aşkan, DDS

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TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

LIST OF SYMBOLS AND ABBREVIATIONS

BioHPP	Ceramic reinforced high performance polymer
Bis-GMA	Bisphenol-A glycidyl dimethacrylate
CAD/CAM	Computer Aided Design/Computer Aided Manufacture
CFR	Carbon fibre reinforced
CQ	Camphorquinone
GFR	Glass fibre reinforced
GPa	Giga pascal
HA	Hydroxyapatite
HEMA	2-hydroxylethyl methacrylate
H_2SO_4	Sulfuric acid
ICI	Imperial Chemistry Industries
LED	Light Emitting Diodes
MMA	Methyl methacrylate
MDPB	12-methocryloytoxydodecylpyridinium bromide
MEHQ	Hydroquinone monomethylether
MPa	Mega pascal
n-TiO_2	Nano-titanium dioxide
n-FA	Nano-flourineapatite
PAC	Plasma Arc
PAEK	Poly-aryl-ether-ketone
PEEK	Poly-ether-ether-ketone
PEKKTON/PEKK	Poly-ether-ketone-ketone
PEKEKK	Polyaryl-ether-ketone-ether-ketone-ketone
PETIA	Pentaerythritol Thiacrylate
QTH	Quartz Tungsten Halogen
SEM	Scanning Electron Microscopy
SBS	Shear Bond Strength
$SiCl_4$	Silicon chloride
Sr-HA	Strontium-containing hydroxyapatite
TBS	Tensile Bond Strength
TEGDMA	Triethylene glycol dimethacrylate
UDMA	Urethane dimethacrylate

UV	Ultraviolet
W/MK	Watts per kelvin-meter
W/cm^2	Watt/square centimetre
J/Kg.K	Joule/kilogram. Kelvin
K	Kelvin
μm	Micrometer
mm	Milimeter
nm	Nanometer
ºC	Degrees Celsius

ABSTRACT

Aşkan, İ. (2023). Load Bearing of Different Fixed Dental Prosthesis Materials with Various Cantilever Lengths. Yeditepe University, Institute of Health Sciences, Department of Prosthodontic Dentistry, Phd Thesis, Istanbul.

The aim of this in vitro study was to observe structural changes of CAD-CAM polyaryletherketons under load as a cantilever fixed prosthesis framework. Three different types of high performance polymers were milled from CAD-CAM blanks to test, after being cemented on an edentulous scenario [Polyetheretherketon (PEEK) (PE), Polyetherketonketon (PEKK) (PK), Bio High Performance Polymer (Bio-HPP) (BH), Lazer sintered Cr-Co alloy frameworks serve as the control group (CC). Samples have been prepared to create two subgroups by two different cantilevers [14 mm cantilevered samples (PE14, PK14, BH14, CC14) (n=10); 21 mm cantilevered samples (PE21, PK21, BH21, CC21) (n=10)]. Each specimen underwent thermocycle and cemented on the main model with resin reinforced glass ionomer cement. Specimens reciprocated vertical load 2 mm away from the most lateral point of cantilever beam, on Instron Universal Testing Machine 3228, carrying out a load to failure test. Decementation and deformation values have been recorded. The conformity of the parameters to the normal distribution was evaluated by Kolmogorov-Smirnov and Shapiro-Wilks tests. The data was analyzed with two way ANOVA. There was no statistically significant difference between the material groups in terms of decementation and deformation values [(p:0.122 ; $p>0.05$ and (p:0.337 ; $p>0.05$)]. There was a statistically significant difference between the cantilever lenghts in terms of decementation and deformation values [(p:0.005 ; $p<0.05$) and (p:0.001 ; $p<0.05$)]. Within the limitations of this study, load to failure results can be showm respectively as; BH14, PK14, PE14, PK21, BH21, PE21 ($\bar{x}$ deformation values; $\pm$288 N, $\pm$255 N, $\pm$249,5 N, $\pm$150,5 N, $\pm$148 N, $\pm$131 N).

Keywords: Fixed Prosthesis, PEEK, PEKK, Bio-HPP, Cr-Co, Cantilever, CAD/CAM

ÖZET

Aşkan, İ. (2023). Değişik Sabit Protez Materyallerinin Farklı Kantilever Uzunluklarındaki Yük Dayanımı. Yeditepe Üniversitesi, Sağlık Bilimleri Enstitüsü, Protetik Diş Tedavisi Anabilim Dalı, Doktora Tezi, İstanbul.

Bu in vitro çalışmanın amacı, yük altında CAD-CAM poliarileterketonların yapısal değişikliklerini gözlemlemekti. Dişsiz bir senaryo üzerine simante edildikten sonra CAD-CAM bloklarından üç farklı poliarileterketon bazlı sabit protez altyapısı üretildi [Polietereterketon (PEEK) (PE), Polieterketonketon (PEKK) (PK), Biyo Yüksek Performanslı Polimer (Bio-HPP) (BH), Lazer sinterleme ile üretilmiş Cr-Co (CC) alaşımlı altyapılar ise kontrol grubu görevi gördü. Örnekler, iki farklı kantilever boyunda iki alt grup oluşturacak şekilde üretilmiştir [14 mm kantileverli örnekler (PE14, PK14, BH14, CC14) (n = 10); 21 mm kantileverli örnekler (PE21, PK21, BH21, CC21) (n =10)]. Bütün numuneler termosiklustan geçirildi ve ana model üzerine reçine takviyeli cam iyonomer simanla yapıştırıldı. Numuneler, kantileverin en distal noktasından 2 mm uzakta, Instron Universal Test Makinesi 3228 ile yük testine sokuldu. Desimantasyon ve deformasyon değerleri kaydedildi. Parametrelerin normal dağılıma uygunluğu Kolmogorov-Smirnov ve Shapiro-Wilks testleri ile değerlendirildi. Veriler iki yönlü ANOVA ile analiz edildi. Materyal grupları arasında desimantasyon ve deformasyon değerleri açısından istatistiksel olarak anlamlı bir fark yoktu [(p:0.122; p>0.05) ve (p:0.337 ; p>0.05)]. Kantilever uzunlukları arasında desimantasyon ve deformasyon değerleri açısından istatistiksel olarak anlamlı bir fark vardı [(p:0.005; p <0.05) ve (p:0.001 ; p <0.05)]. Bu çalışmanın sınırlamaları dahilinde materyallerin yük dayanımları şu şekilde sıralanabilir; BH14, PK14, PE14, PK21, BH21, PE21 ($\bar{x}$ deformasyon değerleri; $\pm$288 N, $\pm$255 N, $\pm$249,5 N, $\pm$150,5 N, $\pm$148 N, $\pm$131 N).

Anahtar Kelimeler: Sabit Protez, PEEK, PEKK, Bio-HPP, Cr-Co, Kantilever, CAD/CAM

1. INTRODUCTION AND PURPOSE

The number of patients asking for fixed implant supported reconstructions has increased considerably in the past few years (1). In the face of this increasing need, new discoveries are being made every day in the fields of both surgery and dental metallurgy so that physicians can respond to the patient's needs with the most appropriate treatment option (2,3,4). In today's world, where edentulism is still a common disease that requires treatment, prosthetic treatment options with few implants and low costs which allow complex surgical procedures to be skipped, are very popular. In such treatment options, the convenience created by cantilever designs for the patient and the physician, especially in the treatment of completely edentulous jaws with intensely resorbed ridges, is considerable. (5,7,17). Although a wide variety of materials, especially titanium, Cr-Co alloy and zirconia, have been used in the construction of implant-supported fixed prostheses for more than 40 years, yet there are some problems such as brittleness, chipping, stress transmitted to the implant, etc. These problems reveal that further study of the material is still needed (6,7). Many prosthetic devices have been produced in recent years with polyaryletherketones (PAEKs), which have been used in the production of various prosthetics in many areas, especially spine surgery, since the 1980s (8). Polyetheretherketon (PEEK) and it's variant such as Bio-HPP and polyetherketonketon (PEKK) are the main materials from which various removable and fixed prostheses have been produced to the date. (9). These high-performance polymers with low elastic modulus have been recommended by the authors to be used as long-term prostheses due to their shock absorbing properties and low water absorption (10). However, in vitro and in vivo studies supporting this suggestion are still quite insufficient (6). The aim of this study is to compare the fixed prosthetic frameworks on implants produced from PEEK, PEKK and Bio-HPP polymers, a PEEK derivative, in terms of load resistance with the control group infrastructures produced from Chromium-Cobalt (Cr-Co) alloy in an in vitro scenario that mimics the oral environment. Two different cantilever sizes, 14 and 21 mm, were determined for the samples to be tested in this study.

The first null hypothesis was that there would be a statistically significant difference in deformation and decementation values between material groups.

The second null hypothesis was that there would be a statistically significant difference between cantilever lengths in terms of deformation and decementation values.

The third null hypothesis was that the joint effect of material and cantilever length on deformation and decementation values would not create a statistically significant difference.

2. GENERAL INFORMATION

2.1. Implant Supported Cantilever Fixed Prosthesis

The definition of "cantilever fixed prosthesis" is, a fixed complete or partial denture in which the pontic is cantilevered and retained and supported by one or more abutments, according to the Glossary of Prosthetic Terms (23).

Unfavorable local conditions of the residual edentulous ridges may lead to two therapeutic options for a rehabilitation by means of implant supported fixed partial denture (FPD): (i) pre-implant reconstructive and/ or regenerative procedures; and (ii) treatment of a partially edentulous site with cantilever fixed prosthesis. As a simple and economical procedure, edentulous ridges next to implants may be reconstructed by means of cantilevers or pontics (18,20,21,22). Especially, if these ridges need augmentation procedures before or during implant placement, a considerable amount of time and money may be saved (19).

The number of patients requiring a fixed prosthesis with implant retention has increased significantly in recent years (11). These types of treatments can generally be limited for financial or clinical reasons (insufficient bone tissue in the areas where the implant would ideally be placed and/or in the presence of anatomical structures such as maxillary sinus and mental foramen) (12). Techniques such as guided bone regeneration and sinus floor elevation have been developed to treat such patients (13). With the use of such techniques; cost, risk of complications, and increased morbidity may limit the choice of treatment (14). In such cases, cantilever extension prostheses allow more direct rehabilitation of edentulous areas (15). According to the results prepared based on comparative studies, it is reported that implant-supported cantilever fixed partial dentures, which have different biomechanical properties than tooth-supported cantilever restorations, can be used safely, especially in aesthetic areas (13,14,15).

2.1.1. Indications of Implant Supported Cantilever Fixed Prosthesis

Considering that soft tissue aesthetics is one of the most important elements of the aesthetic success of any restorations, one of the major targets in aesthetic zone implant procedures might be the protection of the crestal bone around the implant. The

requirement that the distance between adjacent implants being more than 3 mm reveals possible gap problems in edentulous ridges with more than one missing tooth.

Especially in clinical scenarios where there are lacking canines and lateral incisors or central and lateral incisors in the maxillary anterior region, it is sometimes not possible to meet the above-mentioned implant placement criterias because the interdental, edentulous crestal size is narrow (16). In such cases, It is necessary to consider implant-supported restorations in cantilever design as a treatment alternative for all costs.

Another limitation is the presence of more than one missing tooth in the aesthetic area. The factor that may complicate the treatment process is; limited bone. In cases with height and width, advanced surgical stages, including vertical and horizontal bone augmentation procedures, are required. However, in such cases, the application of conservative methods such as placement of short or narrow implants creates more predictable results than methods using advanced surgical procedures. In such cases of insufficient bone presence, cantilever restorations can be considered as an alternative to treatment plans complicated by advanced surgical methods (7).

Determining implant localization with anatomical limitations the old implantological approach, which envisaged implant positions, was replaced by the selection of implant positions with prosthetic guidance (8). This situation has led to the emergence of a surgical perspective that includes applications that will create the anatomical situation that will allow perfect localization of the implant from a prosthetic point of view (9). Over time, maxillary sinus elevation, guided tissue surgical reconstructive practices such as bone regeneration with autogenous or heterogeneous bone grafts have been developed to reveal the ideal bone condition and their success has been scientifically proven. The aim of these surgical applications is to create a three-dimensional bone structure that will allow placing implants that will support prosthetic restoration in prosthetically appropriate positions (25).

All these advanced surgical techniques bring about long treatment times, high morbidity and complication risks, and are also very costly and therefore cannot be widely used (26). Scientific studies conducted over time have focused on investigating the success of techniques such as short implants, implants placed at an angle, or implant-supported fixed partial or complete dentures with cantilevers, which allow cost-effective treatments to be carried out in order to ensure that implant treatments are ideal (12, 27). In 2011, A study conducted by Esposito et al. based on the results, the authors recommended conservative treatment such as short implants. They reported that these

alternatives provide more predictable results than cases where advanced surgical treatments such as sinus elevation or vertical bone augmentation are applied (28).

In summary, implant-supported prosthetic restorations with cantilever extensions;
| | In some cases where the bone condition is not suitable,
alternative to a number of advanced surgical procedures that would require optimizing
It may be effective in reducing the duration of treatment.
| | More predictable treatment if advanced surgeries are not performed
results will be obtained.
| | It will prevent the increase in costs arising from additional surgeries.
| | Additionally, it will provide a more cost-effective treatment opportunity as it will reduce the number of implants.

2.1.2. Biomechanics of Implant Supported Cantilever Fixed Prosthesis

2.1.2.1. Momentum

The moment of a force is the force that produces rotation or twisting. Moment is expressed with vectors (vectors are defined by intensity and direction), and the intensity of the moment force is the product of the force and the horizontal distance extending from the point where the force is applied to the fulcrum point. These forces can also be called torque forces and have a destructive effect on implant systems. Torque or bending forces are seen in implant systems, especially in cantilever bridges and bars, and can cause failures in the bridge or bar extensions, resulting in fractures, bone resorption, screw loosening, and abutment fractures. In the clinic, three coordinate axes (Occlusoapical, buccolingual, there is a possibility of rotation in a total of 6 directions, including mesiodistal).

These rotational forces create microrotations and lead to stress accumulation at the implant-tissue interface at the alveolar ridge and the possibility of bone loss.
There are 3 moment arms in prosthetic restorations;
| | Occlusal height
| | Cantilever length
| | Occlusal width.

2.1.2.2. Cantilever Extension Lenght

High torque forces may occur with vertical forces caused by occlusal contacts in the presence of cantilever extensions. In addition, loads that may come to the cantilever region in the lingual direction also create rotational forces around the implant neck.

Cantilever length is directly proportional to the intensity of the force and is considered as a force arm. As the lever arm lengthens, the load increases and more severe rotational forces occur. Cantilever prostheses supported by splinted implants are subjected to complex loading. While the implant-bone interface adjacent to the cantilever region will receive more compressive load than the second implant, tensile-type forces will be higher in the second implant. In this case, while the first implant carries less risk in terms of bone loss, it is more prone to technical complications in the abutment due to the torsional forces occurring in the abutment area. It has been reported that the tensile-type forces occurring in the second implant may pose a higher risk for bone loss (29).

2.1.2.3. Occlusion and Biomechanics

Occlusal force distributions, the impact mechanisms of these forces on implants and teeth, are the most important factors that determine the behavioral characteristics of teeth and implants under the forces they encounter in the oral environment. The success of implant supported restorations with and without cantilever extensions is also determined by these behavioral characteristics.

The reason why the presence of a cantilever extension has a different effect on restoration success in tooth and implant supported restorations is that teeth and implants exhibit different biomechanical properties under occlusal forces. For this reason, it is important to know increased occlusal forces and their effects. According to some researchers, occlusal overload is seen as one of the most important reasons for peri-implant bone loss and failure of implants or implant prostheses (30, 31, 32-34).

On the contrary, researchers suggest that failures in bone loss or osseointegration occur due to biological complications such as peri-implant infection (38,39). From this point of view, increased occlusal loading does not cause bone loss or osseointegration problems, but these forces may cause mechanical complications in the prosthesis or implant and shorten the life of the implant treatment (40).

Unlike natural teeth, osseointegrated implants are ankylosed to the surrounding bone without a periodontal ligament. In addition, the crestal bone around the implants can act as a fulcrum axis, creating a leverage effect and giving rise to bending-type forces. This situation causes excessive stress accumulation in the crestal bone and the risk of bone loss.

2.1.3. Complication Risks of Implant Supported Cantilever Fixed Prosthesis

While any negativities occurring in the integrity of the implant or abutment is evaluated as a mechanical complication, problems occurring in prosthetic rehabilitation are classified as technical complications (35). Complications such as implant neck fracture, abutment fracture, screw fracture or loosening can be classified as mechanical complications. According to the results of a systematic review published in 2012, the cumulative implant fracture rate for implant-supported cantilever restorations was stated as 0.7%. The most common mechanical complication; It is abutment screw fracture and its incidence rate is 1.6% in 5-year follow-up (36). Restoration infrastructure fracture, loss of retention due to decementation, and veneer fractures are the main technical complications. The most common technical complication is veneer fracture, and the 5-year incidence rate is 5.7% (36).

It is thought that the distribution of chewing forces is unequal in restorations with cantilever extensions. Studies have shown that there is a high force concentration in the implant area, especially in the implant bone gap in the region adjacent to the cantilever extension (41-42). As a result, it is predicted that there may be a higher frequency of complications in cantilever restorations. It has also been assumed that excessive force accumulation will cause some biological complications such as bone loss around the implant (43-44). It has been reported that such extreme forces can create microcracks in the bone and in some areas, forces exceeding the bone remodeling threshold may occur (45).

Romeo et al. reported the survival rate as 97% and the success rate as 98% in the 1-7 year follow-up for cantilever implant-supported fixed prosthetic restorations (46,48). However, Nedir et al. reported complication rates of 29.4% in restorations with cantilever extensions and 7.9% in those without cantilever extensions. According to the study, the cantilever design of the restoration increased the rate of complications (47,48). However, Aglietta et al. according to the results of a systematic review in which they investigated

the long-term survival and complication rates of implant-supported cantilevered fixed restorations, the 5- and 10-year cumulative survival rates are 94.3% and 88.9%, respectively (48). According to these results, cantilever restorations have been reported as reliable treatment alternatives both in terms of long-term survival rates and risk of complications.

Additionally, according to a systematic review conducted in 2012, 5-year the survival rate has been reported as 98.9% (13).

According to the 5-year results of two published clinical studies, cantilever and marginal bone of conventional implant-supported fixed prostheses No difference was shown between the changes in the levels (36, 37).

2.1.4. Long Term Success of Implant Supported Cantilever Fixed Prosthesis

The use of osseointegrated implants in the treatment of partial edentulism is widely accepted as a treatment alternative. Systematic review results reveals very reassuring results for the success and survival rates of implant supported single crown and short fixed dental prostheses (50). In clinical situations where two or more teeth are missing, it is recommended to place two implants to support two single crowns or fixed dental prostheses. However, there may be situations where implants cannot be placed in the ideal three-dimensional position due to limitations in bone condition, inadequacy in the mesio-distal distance of the edentulous area, or anatomical structures. As a solution, implant-supported prostheses with cantilevers can be an alternative.

When it comes to natural teeth, the 5-year survival rate of cantilevered prosthetic restorations is 81.8%, which is significantly lower than the 89.1% survival rate found in conventional tooth-supported fixed restorations (51,52). In implant-supported restorations, cantilever restorations, many studies have evaluated success and survival and without cantilevers give similar results.

According to the results of a systematic review investigating the 5-year survival times of short fixed dental prostheses with implant-supported cantilevers; The survival rate was found to be 94.3% and the estimated biological complication rate was reported as 5.4% (53). The authors reported that the long-term (<5 years) results of fixed restorations with implant-supported cantilever extensions compared to fixed prosthetics with implant-supported abutment terminations. Concluded that it is similar to restorations (55). Cantilever information regarding success and survival rates has been published in

many publications. Although it has been stated that the results of implant-supported restorations with and without extensions are similar, it remains controversial whether there is a difference between the risks of biological and technical complications in the long term.

2.2. Recent History of CAD-CAM Systems in Dentistry

In dentistry, the major developments of dental CAD/CAM systems occurred in the 1980s. There were three pioneers in particular who contributed to the development of the current dental CAD/CAM systems. Dr. Duret was the first in the field of dental CAD/CAM development (54).

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.

The second is Dr. Moermann, the developer of the CEREC® system (56). He attempted to use new technology in a dental office clinically at the chair-side 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.

The third is Dr. Andersson, the developer of the Procera® system (57). 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 (58). 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.

The advantages of using CAD/CAM technology for the fabrication of crowns and FPDs can be summarized as: 1) application of new materials, 2) reduced labor, 3) cost effectiveness and 4) quality control. Materials and their processing technology have been intimately related to the fabrication of dental restorative and prosthetic devices throughout the history of dentistry.

When new materials are introduced as candidates for the material of dental devices, the application of conventional technology is first tested. We sometimes overcome difficulties of processing new materials and succeed in routinely introduce new materials. However, high-strength ceramics that were expected to be the new materials for FPDs frameworks have been difficult to process using conventional dental laboratory technologies. Therefore, this challenged us to apply CAD/CAM processing, particularly at a processing center with large facilities.

Overall, CAD/CAM technology was useful and effective in compensating for changes in dimensions that come with processing chalky material and post-treatment to obtain fit of crowns and FPDs to abutment teeth. Conventional dental laboratory technologies are traditionally labor-intensive. On the other hand, the application of CAD/CAM technology, even within the total processing system, should reduce the labor involved.

Furthermore, systems of outsourcing of some specialized procedures to a processing center using network connections allows for further reduction of labor time. Conventional porcelain dental laboratory processing with powder build-up and baking required a degree of proficiency both in terms of reproducing natural esthetics and shaping because of the extensive sintering shrinkage during baking at high temperatures. Since productivity was ineffective, the cost of conventional porcelain restorations for the patients was also necessarily high. On the other hand, when porcelain crowns are milled from a prefabricated porcelain block using a CAD/CAM machine, for example, the cost of a block is inexpensive because of mass-production. In addition, even with regard to particular esthetic requirements, milled crowns could be completed merely by staining, using a conventional and simple method. This not only reduces labor costs, but provides financial advantages for the owners of dental laboratories and dental offices and, in turn, for patients.

Also production of all-ceramic FPDs using a zirconia framework fabricated by a CAD/CAM process could provide even more financial benefits to owners of dental laboratories because they can invest in small measuring machines and not in large expensive facilities; thus they could concentrate on conventional porcelain processing.

The use of CAD/CAM technology can not only shape restorations by milling, but also allows for quality control of the dental devices by designing optimal shapes based on material characteristics by CAD; thus preventing degradations such as residual strain due to the effects of processing, and ultimately providing reproducible processing. When milling a prefabricated ceramic block, the quality of which has been confirmed beforehand by the manufacturer, there are almost no internal defects in the milled products, whereas conventional powder build-up and baked porcelain products usually contain internal porosity. According to clinical and in vitro studies using finite element and fractographic analyses, the primary causes of failure reported for all-ceramic FPDs differed from those reported for the metal-ceramic FPDs. Fractures of ceramic FPDs tended to occur in the connector areas because of the concentrated stress (59-63).

Therefore, the design of the connector, particularly the dimensions, must be made independently depending on the type of ceramic material used for the framework. CAD better guarantees the durability and reduces the risk of fracture. Processing data can be saved and followed up during the functional period for the device. Even if evidence is required to predict the prognosis of restorations during the functional period, these features detailed here have not been available with the conventional production systems in general use. Therefore, quality control of dental restorative and prosthetic devices using CAD/CAM technology will be a factor with increasing importance in the future with an aging society because such restorative and prosthetic devices will need to function for longer periods as part of the body.

There are no doubts that treatment technologies and materials in dentistry have progressively advanced over the past 50 years, especially in the field of restorative dentistry and prosthodontics. Some opinion leaders in dentistry have stated that dental service in these fields has reached its peak at the moment and that there is no need to develop higher technology in the future. However, we have an objection to this opinion. Restoring a patient's quality of life (QOL) through dental service is becoming more and more important to fostering the health of people in an aging society. We have to offer more comfortable and higher quality dental services to all patients to maintain their oral function and restore their QOL.

Therefore, positive application of new materials and novel technology is essential for dental service in the future. We believe that CAD/CAM technology will contribute greatly to the healthy aging of patients. As already mentioned in this article, there are several directions to apply CAD/CAM systems in dentistry. Among them, introduction of CAD/CAM systems directly into clinics is promising. Patients desire shorter treatment times and early functional recovery for the sake of convenience. Dentists are expected to offer esthetic tooth-colored restorations with one appointment and at reasonable cost for their patients. CAD/CAM technology applied at the chair-side has the potential to deliver this service. Compact, but high-precision, intraoral measurement systems available for use directly in the mouth at the chair-side must be developed.

Additionally, tooth-colored materials that satisfy sophisticated esthetics, with excellent machinability, and the required mechanical characteristics are of course necessary to make this application popular. Besides the demand for esthetic restorations, the demand for comfortable high-quality FPDs and removable dentures is also increasing in the aging society (64).

2.2. Polyaryletherketons (PAEKs) in Dentistry

Polyaryletherketones (PAEKs) are a group of high-performance semicrystalline thermoplastic resins, which family members differ according to their ratio of keto and ether groups (Fig. a). With a higher ratio and sequence of keto groups, the rigidity of the polymer chain and the glass, as well as melting temperature are increasing (95,96).

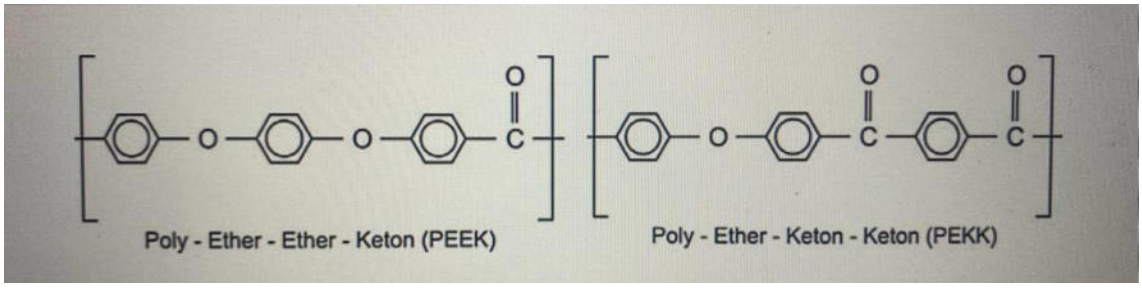


Figure 2.1. Molecular structures of PEEK and PEKK

Following confirmation of its biocompatibility two decades ago, polyaryletherketones (PAEKs) have been increasingly employed as biomaterials for orthopedic, trauma, and spinal implants (97). Commercialized for industry in the 1980s,

PAEK is a relatively new family of high temperature thermoplastic polymers, consisting of an aromatic backbone molecular chain, interconnected by ketone and ether functional groups (98). Two PAEK polymers, used previously for orthopedic and spinal implants, include poly(aryl-ether-ether-ketone) (PEEK) and poly(aryl-ether-ketone-ether-ketone-ketone (PEKEKK) (Fig. b).

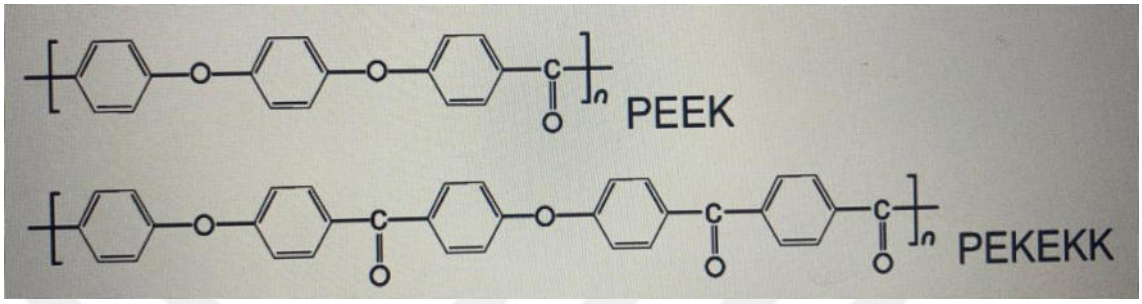


Figure 2.2. Structural differences between PEEK and PEKEKK

The chemical structure of polyaromatic ketones confers stability at high temperatures (exceeding 300 1C), resistance to chemical and radiation damage, compatibility with many reinforcing agents (such as glass and carbon fibers), and greater strength (on a per mass basis) than many metals, making it highly attractive in industrial applications, such as aircraft and turbine blades, for example (96,99). Historically, the availability of polyaromatic polymers arrived at a time when there was growing interest in the development of ‘‘isoelastic’’ hip stems and fracture fixation plates, with stiffnesses comparable to bone. Although neat (unfilled) polyaromatic polymers can exhibit an elastic modulus ranging from 3 to 4 GPa, the modulus can be tailored to closely match cortical bone (18 GPa) or titanium alloy (110 GPa) by preparing carbon-fiber-reinforced (CFR) composites with varying fiber length and orientation (100). In the 1990s, researchers characterized the biocompatibility and in vivo stability of various PAEK materials, along with other ‘‘high performance’’ engineering polymers, such as polysulphone (PS) and polybutylene terephthalate (PBT) (101). However, use of these polymers in implants was abandoned for reasons that are not well documented in the literature. Other polyaromatic ketone polymers, such as PEKEKK, were discontinued by their industrial supplier and thus ceased to be available for biomaterial applications. By the late 1990s, PEEK had emerged as the leading high-performance thermoplastic candidate for replacing metal implant components, especially in orthopedics and trauma (102,103,104). Not only was the material resistant to simulated in vivo degradation,

including damage caused by lipid exposure, but starting in April 1998 PEEK was offered commercially as a biomaterial for implants (Invibio, Ltd., Thornton-Cleveleys, United Kingdom). Facilitated by a stable supply, research on PEEK biomaterials flourished and is expected to continue to advance in the future (106).

2.2.1. Polyetheretherketon (PEEK)

Numerous studies documenting the successful clinical performance of PAEK polymers in orthopedic and spine patients continue to emerge in the literature (107–112). Recent research has also investigated the biotribology of PEEK composites as bearing materials and flexible implants used for joint arthroplasty (113–116). Due to interest in further improving implant fixation, PEEK biomaterials research has also focused on compatibility of the polymer with bioactive materials, including hydroxyapatite (HA), either as a composite filler, or as a surface coating (117-122). As a result of ongoing biomaterials research, PEEK and related composites can be engineered today with a wide range of physical, mechanical, and surface properties, depending upon their implant application. The versatility of PEEK biomaterials necessarily translates into increased complexity, both for implant designers, as well as for researchers seeking to explore new modifications of PEEK for novel implant applications. In recent years, advances in the processing and biomaterials applications of PEEK have been progressing steadily. However, much of the previous research on PEEK implants has been fragmented in the materials science, composites, biomaterials, and application-specific literature. Consequently, the primary goal of this review is to synthesize the disparate repositories of data to provide a comprehensive, state-of-the-art assessment of PEEK and PEEK composites as a family of biomaterials.

PEEK became the dominant high-performance thermoplastic contender for replacing metal implant components, particularly in orthopedics and trauma, by the late 1990s. In 1992, PEEK began to be utilized in dental applications, initially as cosmetic abutments and later as implants. Since then, several compositional alterations have been made to change and enhance the operating features of the implant (123). In April 1998, PEEK was introduced commercially as an implant biomaterial (Invibio, Ltd., Thornton-Cleveleys, United Kingdom). PEEK biomaterials research blossomed because to a steady supply and is predicted to continue to progress in the future.

As a thermoplastic that can withstand high temperatures, PEEK may be processed using a number of conventional commercial procedures, such as injection molding, extrusion, and compression molding.

Injection molding is an appealing production method that is well-suited for the mass production of PEEK implant components.

The production of long stock shapes, such as rods, sheets, and monofilament fibers, can be accomplished through a manufacturing process known as extrusion. Extrusion typically begins with the raw material consisting of PEEK pellets or granules as the starting material.

The production line for stock forms, such as plates or thick sheets, is referred to as compression molding. In most cases, compression molding is utilized for low-volume production, prototyping and evaluation work, as well as for the manufacturing of industrial components with extremely thick sections (124).

Because of its superior chemical, mechanical, and thermal properties, PEEK is now utilized in a variety of applications within the medical industry. These properties are expressed by the material's high strength in conjunction with its adequate milling and grinding properties.

With some of its chemical adjustments and commercial compositions, PEEK material has been managed to bring to the desired state for use in medical and surgical applications. This is due to the fact that PEEK material has demonstrated advancements in the dental industry. The use of PEEK in spinal surgery has been improved thanks to the material's advantageous biomechanical properties, long lifespan, and biocompatibility. In spite of its widespread application, there have been no documented cases of allergy or hypersensitivity to PEEK (125).

Craniofacial deformities are another area of medicine in which PEEK is preferred over other materials. PEEK has been shown to be effective in treating craniofacial deformities, despite the fact that no material has yet been identified as being entirely suitable for this purpose. It is regarded to be a viable alternative to other graft materials due to the fact that it is chemically inert, it can be decontaminated by steam or gamma rays, it has an elasticity that is similar to that of cortical bone tissue, it is biocompatible, it is strong, and it is radiolucent. All of these qualities make it an ideal graft material. According to a number of studies, carbon fiber reinforced PEEK is an excellent choice as a material for the fixation of femurs and tibias (126). Due to the fact that PEEK possesses superior attributes, it has recently begun to be utilized in the field of dentistry, following

its application in the industrial and medical topics. Because the tensile properties of PEEK are similar to those of bone, enamel, and dentin, this material is well-suited for use in dental restorative applications. PEEK is reliably used for implants, abutments, framework material for removable and fixed prosthesis, endocrowns, and post-core systems due to its high degree of biocompatibility and the properties listed above (127).

Because it contains inorganic chemicals, PEEK can be utilized in fixed prostheses as it possesses suitable mechanical properties for such applications. The fact that PEEK material can be repaired more easily than ceramics, that it does not wear down within the mouth, and that there is no deterioration seen in the material characteristics during processing increases the possibility that it can be used as crown material. In addition to this, despite having a low elasticity modulus and hardness, because of its high resistance to wear, this material is capable of competing favorably with metallic alloys (128).

Because of its high level of tensile strength, PEEK is an excellent choice for the framework of a fixed dental prosthesis. The biocompatibility of materials made of PEEK is significantly higher than that of ceramics made from metal. Because of its lower weight, PEEK could serve as an acceptable substitute for chromecobalt ceramics. It is resistant to corrosion even when brought into close proximity with other metals in the mouth. PEEK is insoluble in water and has a low level of chemical reactivity with other materials; as a result, it might be an option worth considering for patients who are allergic to metals or who have a heightened sensitivity to the taste of metals (129).

Because PEEK is a white-gray color, it needs to be veneered with composites in order to achieve the desired aesthetic. Due to the fact that PEEK is chemically inert, a variety of approaches have been investigated and tried in order to achieve a strong bond with veneering materials (130).

2.2.2. Polyetherketonketon (PEKK)

There are no methacrylates present in the thermoplastic high-performance polyether-ketone-ketone material. It was first presented to the public by Bonner in 1962, and ever since that time it has been utilized for a variety of purposes in both the commercial and the military spheres. PEKK, which has properties that make it suitable for use in dental and medical applications, has become an increasingly popular biomaterial in recent years. It is useful for a variety of dental procedures, including implant dentistry, prosthetics, and restorative dentistry. When it comes to the

development of cranial and orthopedic implants, PEKK shows a lot of promise as a material. Their extensive biomedical applications can be attributed to their superior mechanical strength as well as the presence of a second ketone group, which makes it possible to make more surface modifications to the substance's surface (131).

The PAEK is a form of a linear aromatic polyether ketone, and it is represented by polyethylene with an extremely high molecular weight. Both PEEK and PEKK have aromatic rings in their structures, but the proportions of ether- to keto-groups in these rings are different (132). The presence of a second ketone group in PEKK contributes to an increase in both polarity and backbone rigidity. This, in turn, causes an elevation in both the glass transition temperature and the melting temperature. PEKK can behave in either an amorphous or crystalline manner, and this allows for a variety of products to be made from it. Additionally, PEKK possesses robust polymer chains thanks to the presence of an additional ketone group. This contributes to the material's improved physicomechanical properties, such as its compressive strength.

PEKK is a liner thermoplastic polymer that is made up of a benzene ring that has ether or ketone-groups attached to it in a consecutive order. Producing PEKK involves combining diphenyl ether, iso- and terephthaloyl chlorides, aluminum chloride, and nitrobenzene in the appropriate proportions (130).

The properties of pekk material can be examined in two parts as physical and mechanical properties and biological properties.

The physical and mechanical properties of PEKK, such as melting point and compressive strength, are among the best in the industry. The mechanical properties of PEKK are superior to those of PEEK in terms of flexure, tensile, and compressive strength (133).

When compared to unreinforced PEEK, the compressive strength of a product manufactured by PEKK (Pekkton@ ivory Cendres + Métaux, SA, Switzerland) is approximately eighty percent higher (134). The incorporation of titanium dioxide into PEKK results in an increase in both the material's hardness and its resistance to wear (135).

PEKK's ability to absorb shock due to its suitable strength (65 MPa) and fracture resistance characteristics raises the possibility that it could be used as a restorative material (136). When compared to dentin, the PEKK possesses a compression strength that is comparable despite having a lower modulus of elasticity (137). In a manner comparable to that of PEEK, the elastic modulus of PEKK is similar to that of bone.

Therefore, PEKK has the potential to be utilized as a biomaterial for dental implants due to its superior mechanical properties and improved stress distribution (138).

Table 2.1. Comparison of mechanical properties of PEEK and PEKK

	PEKK	PEEK
Tensile strength (MPa)	115	100.69
Elastic modulus (GPa)	5.1	3.5
Flexural strength (MPa)	200	163.88
Compressive strength (MPa)	246	118-169
Melting temperature (º C)	363-386	334-350
Hardness	252 MPa	26-29 VHN
Water absorption ($\mu g/mm^3$)	8.7	0.1-0.5
Density (g/cm^3)	FEFF1.3	1.3

PEKK has been introduced as a potentially more beneficial alternative material to titanium for use in long-term orthopedic applications because of its high level of biocompatibility. In addition to this, PEEK is being put to extensive use in the dental industry as a biomaterial for prosthetics and implants. It provides restorations that don't use any metal, which is beneficial for those who have metal allergies (136).

According to the findings of the studies, the ability of PEKK's surface to undergo chemical modification is enhanced by the presence of another ketone group. This results in a complex surface topography, a greater surface area, and a microrough surface, all of which will have an effect on the behavior of the cells and the osteointegration on the surface of PEKK (138).

In terms of its antibacterial properties, the research indicates that PEKK displays less bacterial adhesion on its surface when compared to the PEEK used in the orthopedic industry (139).

PEKK has recently found use in a variety of different areas of dentistry due to its superior mechanical properties, including its fracture resistance, shock-absorbing capabilities, and better stress distribution. The PEKK is considered an alternative to both ceramics and metals because it does not contain any metals in its restorations, which contributes to its excellent biocompatibility. PEKK is utilized in the manufacturing of dental implants, abutments, crowns, post-core systems, and endocrowns. Additionally, it is a framework element for removable prostheses.

The PEKK material has a high strength to weight ratio, low elastic modulus, acceptable wear resistance, and a low density. It has the potential to be used as a restorative material in fixed prosthodontics, which is one of its potential applications (140).

Recent years have seen the introduction of CAD/CAM technology in the process of fabricating PEKK prosthetic restorations (143,144). Pekkton[@] ivory PEKK is used for monolithic and bi-layered material with an indirect composite veneer (141).

In restorative and prosthetic dentistry, it is absolutely necessary for PEKK to bond to the various restorative materials. In order to bond PEKK using a variety of different adhesives systems, different surface treatment techniques of PEKK have been developed. According to the findings of the studies that compared the bond strength of PEKK to dental resin composite using a variety of surface treatment methods, mechanical surface treatment performed significantly better than chemical surface treatment (142).

Another crucial aspect of prosthetic dentistry is ensuring that the dental restoration is a good fit for the patient. Plaque buildup, recurrent caries, and periodontal damage can all result from inadequate marginal fit, which can ultimately lead to the restoration failing. In a study on marginal fit, compared with zirconia, they found that the marginal fit fell within the acceptable range. This was discovered through observation. On the other hand, the PEKK showed a lower stress distribution around its loading areas, and the PEKK coping was found to have a higher level of fitness when compared to the zirconia coping (140).

In an ideal situation, the tooth and enamel wear that is caused by dental restorations should not be any worse than the natural wear and tear that occurs on teeth over time. The selection of appropriate restorative materials, the goal of which is to have a degree of hardness that is almost similar to that of enamel, is essential for minimizing or impeding the detrimental and irreversible ramifications of tooth or enamel wear. According to the findings of the studies, PEKK led to the greatest amount of material wear while causing the least amount of adverse tooth wear (141).

PEKK has the benefit of having sufficient strength, being lightweight, being resistant to wear, and having an elastic modulus that is comparable to that of dentin. Implant abutments (142) framework material for implant prosthesis (140–144), prosthetic crown materials (134,145), and implant biomaterial are all possible applications for PEKK (146).

PEKK is a material that does not contain any metals and offers an alternative to titanium implants (147). The advantage of using PEKK abutments as the framework for an implant-supported prosthesis is that they are adjustable, compatible with a wide variety of veneering materials, and can be used as a framework (148). A potential material for providing long-term retention in implant prostheses is a combination of the PEKK attachment system and titanium (149).

When subjected to compressive stress, the PEKK framework exhibits lower levels of stress to the implant and tissue than it does when subjected to tensile stress. Because of this, the PEKK framework ought to have some restrictions placed upon it in certain areas so that it can maintain its resilience. There is a wide range of applications for PEKK in oral implantology; however, these applications should be made for appropriate reasons, and additional research is required to investigate the chemical modulation of PEKK in order to improve implant contact (150).

Clasps made of metal in removable dental prostheses have the drawback of being unaesthetic. Additionally, some patients may experience oral galvanism and allergic reactions if they are used (151). Thermoplastic materials have provided a partial solution to these types of problems (152). In recent years, PEKK has become popular for use in removable partial dentures as dental clasps and frameworks through the application of digital technology (153).

PEKK clasps have the capability of providing retention for an extended period of time. In addition, PEKK can be utilized as an insert material for removable partial dentures (154). When compared to the nylon inserts, the PEKK insert demonstrated significantly lower levels of retention change and abrasion (155).

Research concluded that the amount of stress evenly distributed in the framework material of three and four unit small bridges had a significant impact on the structure's integrity. Because of the choice of material, the surrounding tissues were not affected in any way by the strain that was applied to the surrounding area. This demonstrated that PEKK, a polymer, has the potential to serve as an alternative to metal framework (156).

Given its adequate mechanical strength, appropriate processing, and shock-absorbing ability, the PEKK biomaterial has attracted interest in post-core systems. In comparison to metal and fiberglass post-core systems, the biomechanical performance of PEKK is significantly higher. Because of its lower elastic modulus and higher flexural strength, the PEKK demonstrated superior fracture resistance in comparison to metal and fiberglass post-core systems (157,158).

When compared to traditional post-core materials, the PEKK post-core exhibited a more beneficial stress dispersion at the intraradicular surface. This suggests that there is a lower risk of root fracture with this post-core material. PEKK was able to transfer greater stresses to the materials that were in contact with it. As a result, the likelihood of crown and cement debonding failure would be higher at the interface level than it would be with rigid post-core systems, which would be more likely to result in root fracture (158).

The PEKK post-core provides an advantageous stress distribution, which lessens the likelihood of a root fracture occurring. Despite this, there is a possibility that debonding and crown failure will occur more frequently in PEKK post-core because of its flexibility. The use of PEKK material acts as a stress breaker and inhibits the forces that are transferred to the restoration and the tooth root. PEKKTON can be utilized in place of endocrowns for teeth that have had endodontic treatment. In addition to this, PEKK posts are appropriate for the fabrication of posts (159).

2.2.3. Bio High Performance Polymer (Bio-HPP)

Bio-HPP (High Performance Polymer) is a high-tech thermoplastic polymer based on PEEK. It was created and optimized for dental use. It contains ceramic microparticles for better polishing of the restorations. These ceramic fillers have a size of about 0.3-0.5 microns and occupy 20% of the total volume of BioHPP (fig 2.3.) (160). Because of their micro size, homogeneity is achieved in the macrostructure of the polymer. The high degree of polishability of the material results in a lack of plaque retention and colour stability over time. BioHPP is as close as possible to the bone, because of its coefficient of elasticity (around 4 GPa). This is very important in implant treatment in cases when twisting forces may occur. The chewing pressure is transmitted as gently as possible, and the risk of fracture is reduced, as a result of the BioHPP modulus of elasticity close to that of the spongiose bone (161).

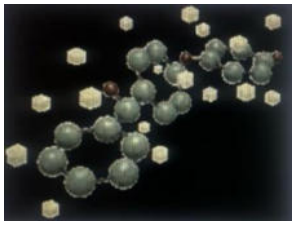


Figure 2.3. Ceramic fillers are shown in beige color in Bio-HPP compound

BioHPP (High Performance Polymer) is based on polyether-ether-ketone (PEEK) polymer and was introduced as dental material for manufacturing the superstructure dentures on dental implants by the Bredent factory. Their strength is due to the special ceramic filler (with the grain size of 0.3 to 0.5 µm), which optimised the mechanical properties. Due to this very small grain size, constant homogeneity can be produced. This homogeneity is an important prerequisite for these outstanding material properties and forms the basis for consistent quality. In figure 1 is presented the structural formula of a PEEK molecule, in which the white cloud is an indicator of the ceramic filler which accounts for excellent mechanical material properties, especially for the use in dental techniques (162).

BioHPP, approved as a Class II a medical device, is a semi-crystalline and pigmented thermoplastic. Its base material is PEEK and it contains about 20% ceramic filler. With a modulus of elasticity of around 4 GPa, BioHPP is about as elastic as bone, which helps mitigate any stress that might develop and reduces stress shielding. This also means bone-related torsion can also be balanced out to some extent, which is important with larger implant work. In addition, BioHPP is also particularly suitable for patients with allergies because of its very low water solubility of < 0.3 µg/mm3 and its low reactivity to other materials. In figure 3 is presented the elasticity comparison for bone and framework materials used in dentistry, in logarithmic representation (fig 2.4.) (163).

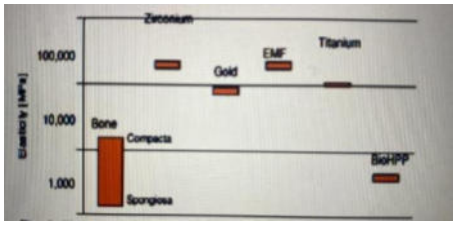


Figure 2.4. Elasticity compariason for bone/framework materials

The combination of its mechanical properties its biocompatibility makes PEEK/ BioHPP material very attractive to medical and dental applications (164). The material BioHPP is a pigmented, semi-crystalline thermoplastic. The base material is polyetheretherketone (PEEK), which was developed as a veneer-compatible framework material. The good material properties are not impaired during processing (165,166). The E-modulus of BioHPP lies in the range of 4000 MPa, which very strongly resembles the elasticity of human bone (e.g. in the mandible), so that the chewing forces are therefore

cushioned. The maximum fracture resistance indicates the force (in Newtons) at which the sample fails. Values of up to 1200N were reached during the tests which, in comparison to a maximum chewing force of 500N for a human bite, represent an adequate safety margin. The bond strength of BioHPP framework is of over 25 MPa. Gum irritation is ruled out due to the surface quality of the material and its low rough depth of 0.018 µm RA (Jena Uni). Other characteristics of BioHPP polymeric dental material are: flexural strength is >150 MPa, water absorption=6.5 µg/ mm3 , water solubility 25 MPa, thickness=1.3-1.5 cm3 , hardness (HV)=110 HV 5/20, thermocycling 10,000 cycles 5°C/55°C in accordance with DIN EN ISO 10477, E-modulus=4,000 MPa (167,168). After the researches of Vosshans et al, BioHPP as a framework material have a lot of advantages like: preparation of restorations with a low specific weight, elasticity similar to that of bone, sock-absorbing effect, metal-free restorations, low material fatigue, no viscoplastic fractures, high biocompatibility, low plaque accretion, no corrosion (169). So, this polymer used for dental restorations, satisfy various requirements: osseointegration of implants, stress-free primary framework, fixedness of prostheses, convenient insertion/removal of movable prostheses for patients, good hygiene, plaque resistance, colour stability, low weight. In the material BioHPP the authors have found the alternative we have been looking for to be used on a day-to-day basis in both practices and laboratories. After the study of Wiesli and Ozcan, HPP polyetheretherketone consisting of a single monomer and featuring a low Young modulus may be advantageous (170). PEEK seems to lead to less osteolyses and healing problems and no scattering in radiation was observed. Some animal studies showed direct contact between PEEK and the bone with high biocompatibility and no evidence for cytotoxicity, mutagenicity, carcinogenicity, and immunogenicity to the present day.

After the researches of Kistler et al, BioHPP is extremely resistant to abrasion, have highly dense structures and also have excellent anti-discolouration properties (171). Unlike traditional plastics, stress tests involving coffee, tea, tobacco, methylene blue, and red wine recorded average levels of discolouration when compared with ceramic materials. In cases of extreme stress involving damage, the plastic (unlike restorations with ceramic veneers), can simply be repaired. There is no need for time consuming drying of the framework and it is also possible to carry out repairs in the mouth if necessary using light curing systems.

BioHPP developed by Bredent GmbH has been developed especially for dental applications. This polymeric material has been tested in the Universities of Jena and

Regensburg, and the tests have been shown that the elasticity of the material resembles strongly to the elasticity of human bone, property which makes it a very interesting material for the restoration of dental implants. Implants are osseointegrated without the formation of a periodontal ligament compared with a natural tooth. Conventional materials such as metal and zirconium have very little elasticity, which can lead to fractures or TMJ problems (165). Denture framework are conventionally made from Chrome-Cobalt. BioHPP can be a viable alternative to Chrome-Cobalt as it is lighter, does not cause any galvanic elements (corrosion) when in contact with other metals in the mouth. BioHPP which can be pressed seems to offer slightly less bulky structures but long term in vivo tests are not available yet (166,167). Considering mechanical and physical properties similar to bone, PEEK can be used in many areas of dentistry. Improving the bioactivity of PEEK dental implants without compromising their mechanical properties is a major challenge. Further modifications and improving the material properties may increase its applications in clinical dentistry (171).

The numerous disadvantages of classic thermo- polymerisable acrylic dentures are well known: poor resistance to deformation and wear, poor long-term performance and stability, poor tolerance, usual presence of residual monomer which induces allergies in a high percentage of the patients, porosity which helps the development of microorganisms and deposits (172). Multidisciplinary studies present the benefits for the knowledge of biomaterials' properties, in reducing the failures of their using, and for their optimization of biomechanical performances of the dentures and their lifetime, with an evident influence on senior's oral health (173). The system solution based on BioHPP as a highly cross linked polymer, the "for 2 press" compression-moulding process tailored to it, and their use within a monolithic or veneered application represent a viable alternative to metallic or even full ceramic restorations in terms of either implant-based prosthetics or restorative therapy. Since it is white in colour, it is ideal for aesthetic restoration purposes.

The material is associated with a high level of stability, very good polishing qualities, and a low affinity for plaque. With bigger implant projects, however, bone-related torsion can still be balanced out by the elasticity of the material, which is similar to that of bone. Its insolubility in water and low reactivity with other materials mean the material PEEK is also very suitable for patients with allergies. This means BioHPP favours new approaches to treatment based on familiar manufacturing methods. There is no need for time consuming and costly training for new systems. The added value

associated with any restorations is kept within the laboratory, not least because the high investment costs for things like milling units do not apply (174).

2.4. Chrome-Cobalt (Cr-Co) Alloys and Direct Metal Lazer Synthering

Metal-supported restorations have been used in dentistry for many years, and their success has been demonstrated by long-term clinical studies (65-68). It is more common for metal-supported restorations to fail due to biological complications rather than technical complications (69-72).

Cobalt-chromium alloys are biocompatible materials, and metal-ceramic restorations are widely used as metal infrastructure due to the high costs of noble alloys and the risk of allergic reactions in chromium-nickel alloys (73-75). Generally, cobalt-chromium alloy contains 65% Co, 26% Cr, 9% Ni. Cobalt and nickel are hard and durable metals. The function of chromium is to further harden the alloy by solution hardening and to make it resistant to corrosion through its passivation effect. Chromium exposed on the surface oxidizes rapidly and forms a thin passive oxide layer. This oxide layer protects the inner layers against corrosion. It belongs to the extra hard (Type IV) alloy group.

Traditionally, Co-Cr restorations are produced using waste-wax and casting techniques. The most important disadvantages of this technique are distortion in modeling, metal shrinkage during casting, formation of irregular edges, as well as the fact that the processes are technically sensitive and take a long time (76).

Today, CAD/CAM milling and direct metal laser sintering techniques have begun to be used to prevent the problems that occur during the production of Co-Cr restorations by casting. These procedures shorten laboratory times and make it possible to produce restorations with standard parameters (76). In the CAD/CAM milling method, restorations are obtained from prefabricated Co-Cr blocks by milling method. In the laser sintering method, restorations are obtained by adding Co-Cr powders by spraying under high temperatures (76).

Technological advances constantly contribute to the development of new production techniques in daily dental practice. In the last decade, digital technologies have been used in prosthetic dentistry to produce metallic infrastructures (77-81). Digital methods used to produce metal restorations; They can be examined in two main classes: subtractive and additive methods. With the subtraction method, computer-aided production is carried out by milling the produced metal blocks, while in the additive

method, production is made by combining metal powders by melting them under high temperatures with laser core heat. This method can be called laser sintering, and in this technique, production is done with CAD/CAM technology (82). Laser sintering technique, which has a not very old history, is a very popular research topic today. There are many studies in the literature on the use of Co-Cr metal alloys with laser sintering (80, 83-92). In addition, a small number of studies have been found in the literature regarding its use with Ti alloys (93). These studies on Cr-Co alloys are generally; It is related to the marginal adaptations (80, 85-88, 94), connection strengths with dental porcelains (79, 89, 93), internal porosity (78), micro irregularities (85) and electromechanical properties of the restorations produced with this technique. It causes large microstructural differences between the production process in the casting technique and the laser sintering production process. It has been reported in the literature that restorations produced by laser sintering have higher hardness degrees and more homogeneous microstructures than cast restorations (82).

2.5. Thermocyle

The biggest disadvantage of laboratory studies in the field of dentistry is that they cannot fully reflect the clinical environment, that is, the oral environment. The oral environment contains all the factors necessary for fatigue failure of prosthetic restorations. For this purpose, long-term clinical studies are needed for realistic data on the longevity and durability characteristics of dental restorations. However, due to the difficulty of providing ideal conditions and standardization in clinical studies and laboratory studies, devices have been developed that can mimic the oral environment in the laboratory environment and allow the restorations to be exposed to fatigue equivalent to the desired clinical period (94).

3. MATERIALS AND METHOD

Preparations for our study were made at Yeditepe University, Faculty of Dentistry, Hard Tissue Laboratory. The infrastructure models to be tested were engraved in Istanbul, Şişli Optimal Dental Prosthesis Laboratory; The entire experiment was carried out at Yeditepe University, Faculty of Mechanical Engineering, Materials and Metallurgy Laboratory.

All equipment and materials used in this research and manufacturer information are shown in table 3.1.

Table 3.1. Materials used and manufacturers

Gereç ve Materyaller	Üretici Firma
5.1 - 10 mm Bone Level Implant	Institut Straumann, Basel, Switzerland
Anatomik Titanyum Dayanaklar	Institut Straumann, Basel, Switzerland
Impression Posts	Institut Straumann, Basel, Switzerland
Pattern Resin	GC, America Inc., Alsip, United States
Acrylic Resin	Technovit 4000, Heraus Kulzer GmbH, Wehrheim, Germany
Cement	Fast Setting White Cement, Küçükçekmece, Turkey
Static Load Device	Instron 3382, Instron, Massachusetts, United States
CAM	Dentifa Nio, Umraniye, Turkey
Metal Laser Synthering	SISMA S.p.A., Vicenza, Italy
PEEK Blank 16 mm	CopraPeek Rose, Whitepeaks Dental Solutions GmbH & Co. KG, Essen, Germany
PEKK Blank 16 mm	Pekkton, Cendres+Métaux, Sweden
Bio-HPP Blank 16 mm	BreCAM, Bredent, Germany
Cr-Co Alloy	ADORBOND CC Plus Pulver, Germany
GC FujiCEM Evolve	GC Co., Tokyo, Japan
Acrylic Stick	GVNART, Kadikoy, Turkey
Optical Scanner	inEos X5, Dentsply Sirona, Erlangen, Germany
Thermocycle Device	Salubris Technica, Dentester, Turkey

3.1. Main Model Preparation

In this study, it was aimed to prepare a main model that mimics the clinical scenario with fully or partially long edentulous areas (Figure 6). In order to achieve this, a tray had to be created that could keep the components in the oral environment balanced and stable against the load. Moreover, this table had to be compatible with the table of the load device on which the experiment would be performed and be able to transmit the load arm to the most distal point of the model.

After examining the physical properties of the device, taking into account the heights of the intraoral components and the models to be cemented on them, the ideal width and length values of the main model table were determined and noted. To make the oral environment similarity more real, the implants to which the materials to which the load would be applied had to be embedded in a bone-like material.

Since previous studies showed that load tests using an acrylic resin that mimics alveolar bone density were successful, it was decided to embed the implants in this study in the same acrylic resin. Fast-setting white cement was used for the production of the subplate. Cement was poured into a rectangular prism-shaped mold with dimensions of 15 cm x 13 cm x 3 cm. Before the cement set, an old cylindrical cream box with a diameter of 6 cm and a depth of 3 cm was placed in the center of the block and the box was removed after hardening was completed (Fig 3.5.). Thus, the oral cavity where the acrylic resin and components would be placed was preserved on the surface of the cement table (Figure 3.1.). After the cement set, acrylic resin (Technovit 4000, Heraeus Kulzer GmbH, Wehrheim, Germany), which mimics alveolar bone density, was applied to this space.

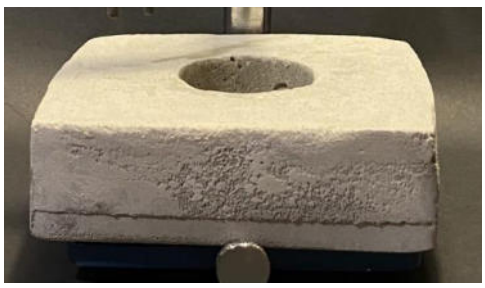


Figure 3.1. (Left) Main model preparation from fast setting cement

Figure 3.2. (Right) Acrylic resin

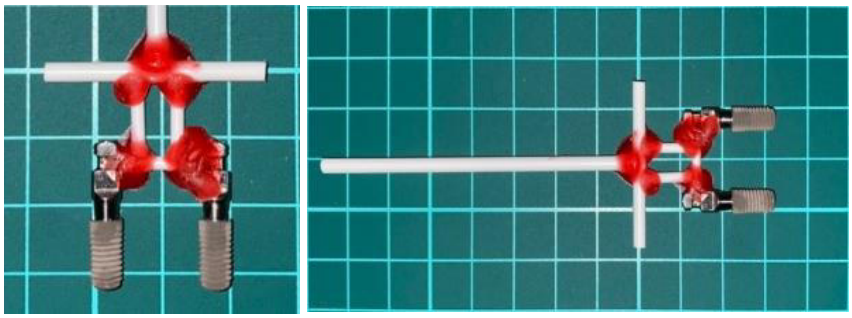


Figure 3.3. Straumann bone level implants

Two bone level implants (Institut Straumann, Basel, Switzerland) with a diameter of 5.1 mm and a length of 10 mm were fixed to a plastic rod with a distance of 15 mm between their centers, with a pattern of resin from impression posts screwed into them (Figure 3,3). A green architectural drawing background was used to ensure that the fixed parts were arranged neatly and the components were aligned in parallel. After the implants fixed to the fixed acrylic rod were placed in the parallelometer, the resin in the cement block was buried at the bone level (Figure 3,4).

Anatomical titanium abutments (Institut Straumann, Basel, Switzerland) with a diameter of 4.8 mm, compatible with the implants, were selected (Figure 3.5.). Abutment screws are made of Titanium-Aluminum-Niobium alloy (Ti6AL7Nb). This type of abutment can only be used as a support for cemented type crown and bridge restorations. In this study, no etching process was performed on the abutments. The fracture, fatigue and bending durability of these abutments and the success of the implant-abutment connection type have been demonstrated in many studies.

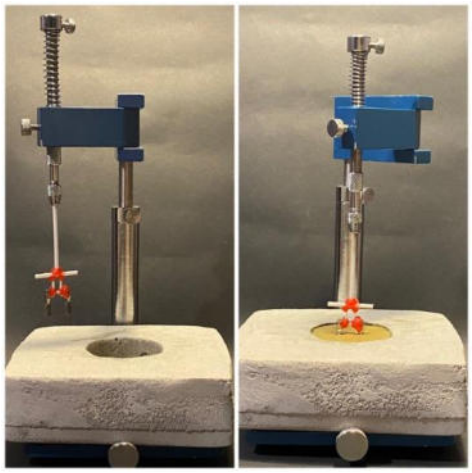


Figure 3.4. Implants setted with paralelometer

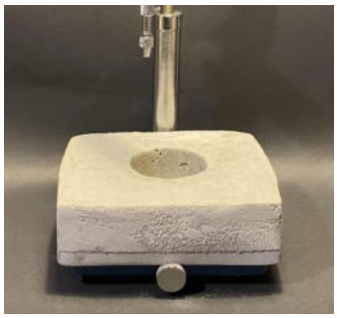
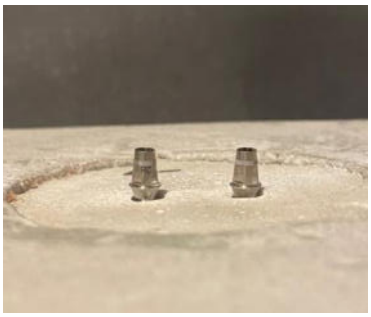
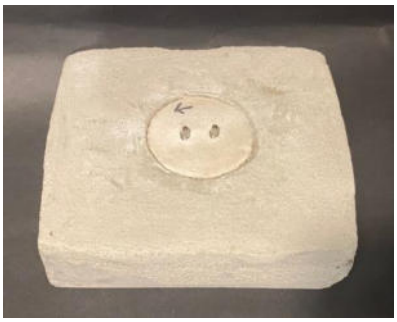




Figure 3.5. (Left) The cavity preserved in main model cement for acrylic resin and implant components

Figure 3.6. (Middle) Acrylic resin dries after 10 minutes

Figure 3.7. (Right) Final view of the main model before scanning

3.1. Transferring the Main Model to Digital Media

The abutments on the concrete block were scanned with an optical scanner (inEos X5, Sirona, Erlangen, Germany) and then transferred to the computer. The components on the concrete block were sprayed with matte powder so that they could be viewed in the optical scanner without glare.

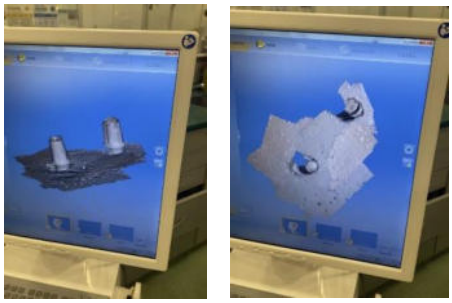


Figure 3.8. Scanning trial with standard intraoral scanner, fails because of the glare

Figuıre 3.9. (Left) Scanning with external lab scanner and mattyfying spray succeeded

Figure 3.10. (Right) Abutments on digital media

3.2. Preparation of the Samples

In the study, substructures with lengths of 31.5 mm and 38.5 mm and cantilever lengths of 14 or 21 mm were produced to be cemented on two implants with a center distance of 15 mm. The height of the infrastructure models is 5 mm and the width is 7 mm. In determining the cantilever lengths, the average mesiodistal premolar width of 7 mm was taken as a basis, and the 2 or 3-membered cantilever scenario was imitated. The parameters used in previous studies were effective in determining the height as 5 mm. (175,182).

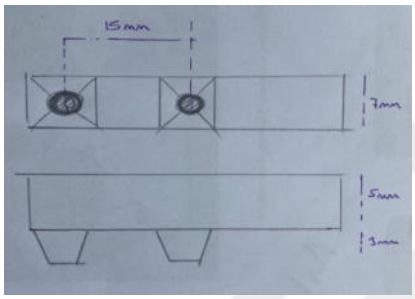


Figure 3.11. Schema of the specimen

Substructure samples were milled from PEEK (CopraPeek Rose, Whitepeaks Dental Solutions GmbH & Co. KG, Essen, Germany), PEKK (Pekkton, Cendres+Métaux, Switzerland), Bio-HPP (BreCAM BiO-HPP, Bredent, Germany) blocks; The Cr-Co alloy forming the control group was made by laser sintering method. From each material group, substructures to be cemented on the abutments were produced (n = 20, N = 80), creating a 14 mm (n = 10) and 21 mm (n = 10) long cantilever. These values were obtained from previous studies (Fig 3.12.) (182).

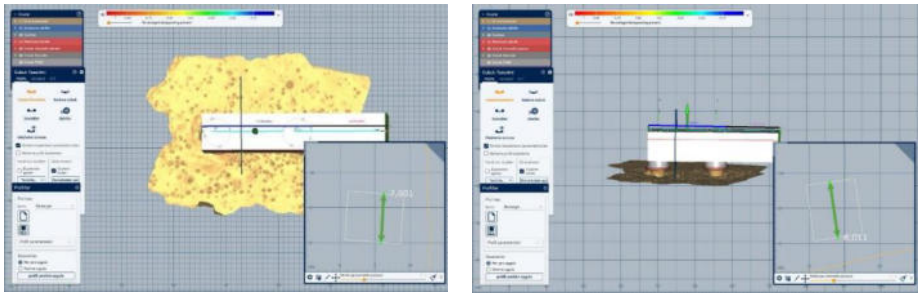


Figure 3.12. Making the design on Exocad design programme


Figure 3.13. (Left) PEKK blank is getting milled

Figure 3.14. (Right) PEEK blank is getting milled



Figure 3.15.(Left) CopraPeek Rose, Whitepeaks Dental Solutions GmbH & Co. KG, Essen, Germany

Figure 3.16. (Middle) BreCAM BiO-HPP, Bredent, Germany

Figure 3.17. (Right) Pekkton, Cendres+Métaux, Switzerland

Due to the sample sizes determined in this study, two blanks from each material group were used. After every polymeric material is milled, the burr was changed.

Blanks were placed into CAM device for the milling process. Minimum twice blanks were used for every other polymer.

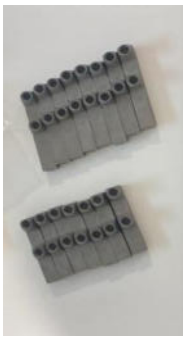


Figure 3.18.- 3.20. (Left) Metal lazer synthering, SISMA S.p.A., Vicenza, (Middle) ADORBOND CC Plus Pulver, Germany, (Right) End products of the machine

Same cad data was used to produce Cr-Co specimens finally. After the data was loaded on metal lazer synthering device, ADORBOND CC Plus Pulver was added (Fig 3.18.). The process started by laser light welding the pulver. All of the specimens came out from the same pulver bottle. Polymers were ready for the testing phase of the study however Cr-Co samples hand polished first with lab burrs to eliminate unleveled surface after the models were taken out from the synthering device. After the polishing and removal of the the unleveled surfaces on Cr-Co samples, all of the specimens made groups in cloth bags and placed in thermocycling device alltogether. It was setted on 20.000 cycles at 5°C and 55°C. Each cycle was completed at 20 seconds.

Figure 3.21. (Left) Thermocycling Device (Salubris Technica, Dentester, Turkey)
Figure 3.22. (Right) Specimens ready for the device

3.3. Cementation Process and Load to Failure Test

The substructure samples were bonded to the main model using the traditional cementation method for ease of repeated application and cleaning. In this study, resin-reinforced glass ionomer cement was preferred due to its ease of use and cleaning with a syringe (Fig 3.23.).

Figure 3.23. GC Fujicem Evolve, GC Co., Ltd., Japan
Figure 3.24. Cleaning out the residuel cement with a probe

The substructure samples were cemented to the abutments according to the manufacturer's instructions, and the models were placed in the testing device after 5 minutes after the cement set according to the manufacturer's instructions.

After each infrastructure sample was bonded and tested with the same protocol, it was dismantled from the main model, and the process was repeated after the main model supports were cleaned from the residual cement with probe and a toothbrush (Fig 3.24.).

The load arm was adjusted to contact the imaginary tubercle tip level of the premolar tooth on the cantilever bar of the substructure samples. For each sample with a length of 14 and 21 mm, the load contact level is the point that corresponds to 3.5 mm from the posterior to the anterior of the cantilever bar.

3.4. Measuring Methods

The main model was placed on the table of the universal testing device and before each sample, it was made sure that the substructures were parallel to the table with a digital spirit level.

During the experiment, for each sample, a graphical screenshot, which was the projection of the path followed by the load arm, was recorded on the computer screen of universal testing device. In the recorded graphs, four situations were extracted:

-The amount of load applied until the first decementation of the material [named as decementation value (N) in this study].

-The amount of load applied until the second decementation or fracture [named as deformation value (N) in this study].

-The amount of downward extension of the load arm until the first decementation of polymeric materials (mm).

-The amount of load applied to control group until abutment deformation (N).

Decementation values (N) and deformation values (N) were the common ground of six group of materials [PE14, PE21, BH14, BH21, PK14, PK21; (n=10)], hence were taken for statistical analysis. The downward extension values (mm) and deformation load (N) of abutment under alloy frameworks (CC14) will later be discussed in the discussion section.

3.6. Statistical Analysis

While evaluating the findings obtained in the study, IBM SPSS Statistics 22 program was used for statistical analysis. The suitability of the parameters for normal distribution was evaluated with Kolmogorov-Smirnov and Shapiro Wilks tests and it was determined that the parameters were suitable for normal distribution. While evaluating the study data, two-way ANOVA test (2-way ANOVA) was used to evaluate the effect of material and cantilever length on deformation and decementation parameters. Significance was evaluated at $p<0.05$ level.

4. RESULTS

Polymeric materials flexed or fractured to failure during the test. This leaded the furthest abutment from the cantilever extension to decement first and it continued with the second abutment's decementation with all of the PEEK and Bio-HPP samples. Total of PEKK samples however, fractured from the same collar area following the first decementation.

Table 4.1. Descriptive data of decementation values (N)

Decementation Values (N)	n	$\bar{x}$	min	max	$\pm$	median	skewness	kurtosis
PE14	10	147,5	20,0	330,0	102,4	122,5	0,5	-0,9
PE21	10	70,5	30,0	125,0	35,2	65,0	0,4	-1,2
PK14	10	120,5	20,0	260,0	74,6	122,5	0,5	-0,2
PK21	10	99,0	30,0	200,0	61,8	85,0	0,6	-0,0
BH14	10	180,0	80,0	290,0	67,9	175,0	0,2	-1,0
BH21	10	119,5	20,0	205,0	63,6	117,5	-0,2	-1,2

Table 4.2. Descriptive data of deformation values (N)

Deformation Values (N)	n	$\bar{x}$	min	max	$\pm$	median	skewness	kurtosis
PE14	10	249,5	110,0	350,0	68,8	245,0	-0,5	0,9
PE21	10	131,0	100,0	175,0	26,6	122,5	0,7	-0,9
PK14	10	255,0	20,0	260,0	74,6	122,5	-0,4	-0,2
PK21	10	150,5	30,0	200,0	61,8	85,0	1,4	-0,0
BH14	10	288,0	195,0	550,0	97,2	257,5	2,6	7,4
BH21	10	148,0	90,0	205,0	37,7	145,0	0,2	-0,8

4.1. Deformation of PEEK Samples

-PEEK frameworks with 14 mm cantilever started to bend clockwise from the connection point of distal abutment and cantilever beam until the average vertical load of 147,5 N causing the cantilever beam to bend downwards between 1,3 mm – 1,7 mm.
At average vertical load of 147,5 N the furthest abutment to the cantilever beam decemented. From this point, cantilever beam started to flex on the single abutment and the second decementation completed until total dislocation of the samples at the average vertical load of 249,5 N.

-PEEK frameworks with 21 mm cantilever started to bend clockwise from the connection point of distal abutment and cantilever beam until the average vertical load of 70,5 N causing the cantilever beam to bend downwards between 0,5 mm – 3,0 mm. At average vertical load of 70,5 N the furthest abutment to the cantilever beam decemented. From this point, cantilever beam started to flex on the single abutment and the second decementation completed until total dislocation of the samples at the average vertical load of 131 N.

Figure 4.1. – 4.3. (Left) PE1401 Framework reciprocating 330 N, load arm extension 1,2 mm, not decemented yet, (Middle) PE1401 Framework decemented, reciprocating 110 N, (Right) PE1401 Framework dislocated after reciprocating 200 N

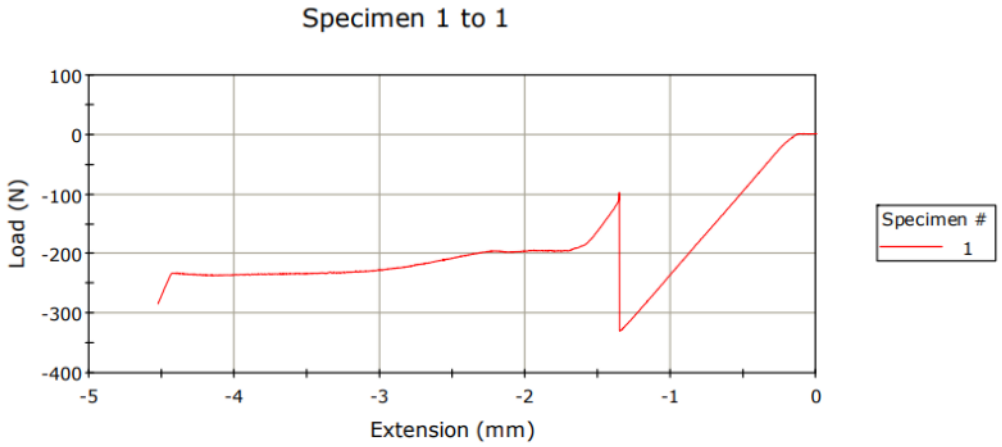


Figure 4.4. Load to Failure (N) test graphic of PE1401 Framework

Fig 4.5. – 4.7. (Left) PE2101 framework reciprocating 175 N, load arm extension 2,51 mm, not decemented yet, (Middle) PE2101 framework decemented, reciprocating 150 N, (Right) PE2101 framework dislocated after reciprocating 200 N

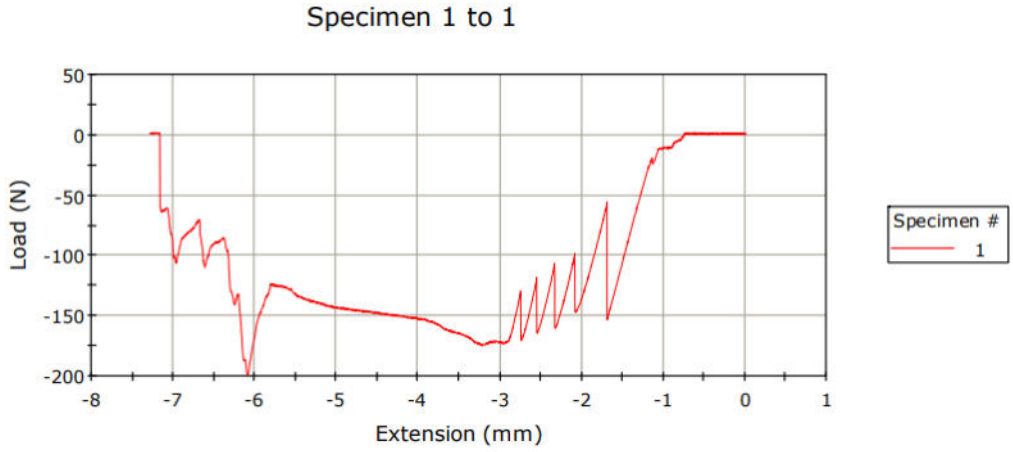


Figure 4.8. Load to Failure (N) test graphic of PE2101 framework

4.2. Deformation of PEKK Samples

-PEKK frameworks with 14 mm cantilever started to bend clockwise from the connection point of distal abutment and cantilever beam until the average vertical load of 120,5 N. As the load continued to apply, samples flexed on the single abutment, causing the cantilever beam to bend downwards between 0,3 mm – 1,0 mm and eventually fractured from the collar area at the average vertical vertical load of 255 N.

-PEKK frameworks with 21 mm cantilever started to decement instantly with the load applied. The furthest abutment to the cantilever beam decemented first at the average vertical load of 99 N. As the load continued to apply, samples flexed on the single abutment, causing the cantilever beam to bend downwards between 0,18 mm – 1,95 mm and eventually fractured from the collar area at the average vertical load of 150 N.

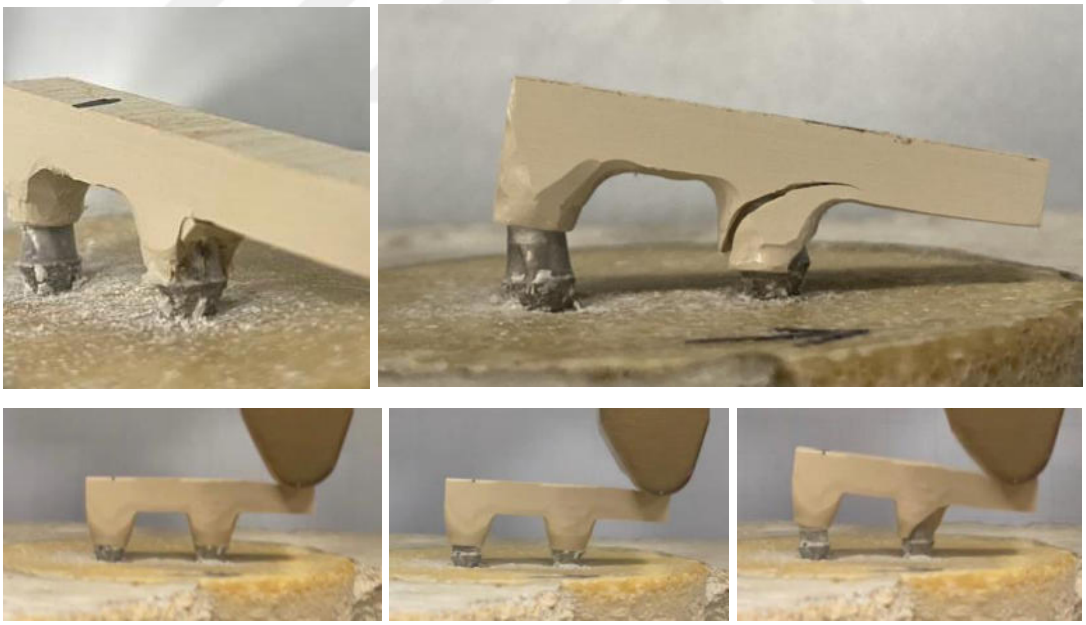


Fig 4.9. – 4.13. (Above, Left) Fracture on the PK2101 framework, (Above, Right) Diagonal fracture on the PK1409 framework, (Below, Left) PK1401 framework reciprocating 83 N, load arm extension 0,38 mm, not decemented yet, (Below, Middle) PK1401 framework decemented, reciprocating 67 N, (Below, Right) PK1401 framework fractured after reciprocating 239 N

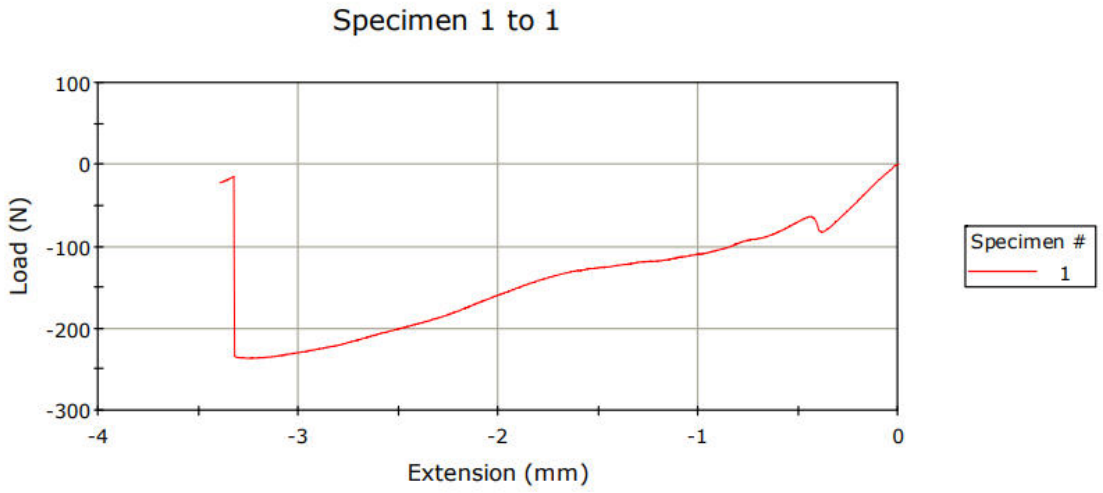


Figure 4.14. Load to Failure (N) test graphic of PK2101 framework

4.3. Deformation of Bio-HPP Samples

-Bio-HPP frameworks with 14 mm cantilever started to bend clockwise from the connection point of distal abutment and cantilever beam until the average vertical load of 180 N causing the cantilever beam to bend downwards between 0,9 mm – 1,4 mm.
At average vertical load of 180 N the furthest abutment to the cantilever beam decemented. From this point, cantilever beam started to flex on the single abutment and the second decementation completed until total dislocation of the samples at the average vertical load of 288 N.

-Bio-HPP frameworks with 21 mm cantilever started to bend clockwise from the connection point of distal abutment and cantilever beam until the average vertical load of 119,5 N causing the cantilever beam to bend downwards between 0,3 mm – 1,7 mm.
At average vertical load of 119,5 N the furthest abutment to the cantilever beam decemented. From this point, cantilever beam started to flex on the single abutment and the second decementation completed until total dislocation of the samples at the average vertical load of 148 N.

Figure 4.15. – 4.17. (Left) BH1401 framework reciprocating 210 N, load arm extension 1,22 mm, not decemented yet, (Middle) BH1401 framework decemented, reciprocating 200 N, (Right) BH1401 framework dislocated after reciprocating 292,5 N

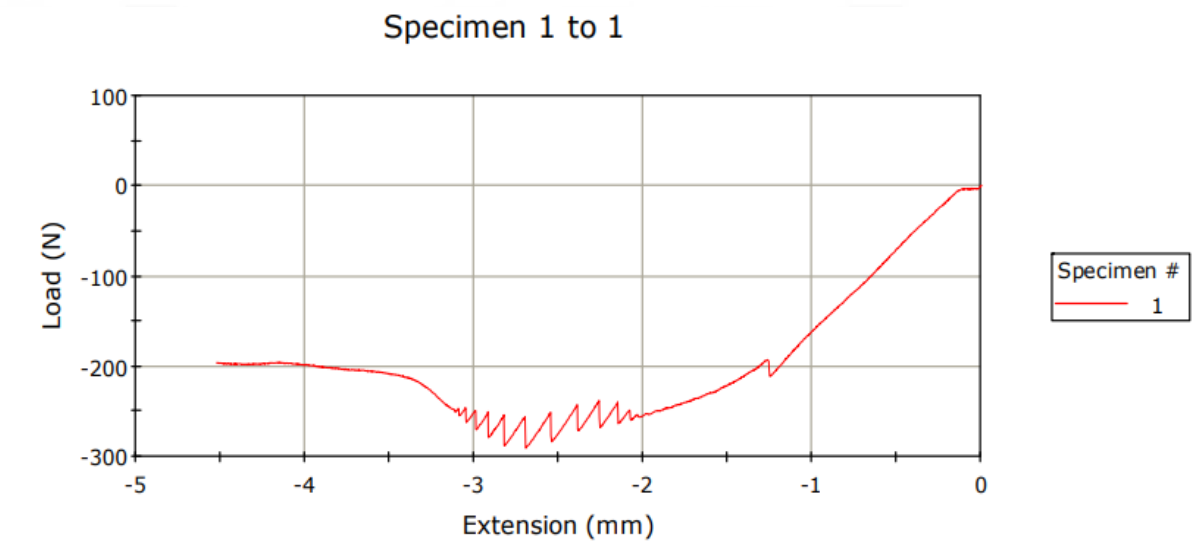


Figure 4.18. Load to Failure (N) test graphic of BH1401 framework

4.4. Deformation of Cr-Co Samples (Control Group)

While testing Cr-Co frameworks, the abutments of the main model was visibly damaged after the third sample. Despite the damage, we continued the test with the fourth and last sample because Cr-Co group was the last testing group, so the main model will not be used again. After the third specimen, main model broke into pieces and abutments bended and no more sample were able to fit onto the model anymore. Eventually only four specimens' test measurements were valid and they were not suitable to run a statystical analysis, so we left Cr-Co datas out from the

results and statystical analysis. The tested specimens were CC1401, CC1402 and CC1403 ,

(n=3).

Figure 4.19. (Left) CC1403 specimen's load to failure test (N)

Figure 4.20. (Right) Abutments were visibly damaged after the first testing

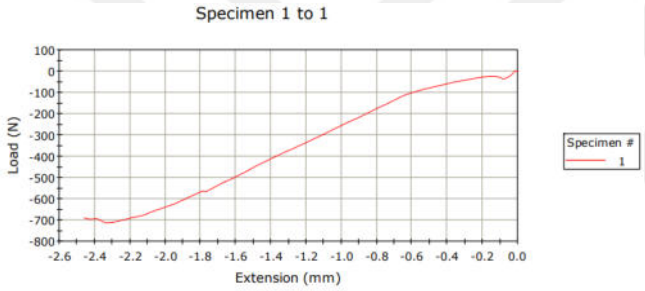


Figure 4.21. Load to Failure (N) test graphic of CC1401

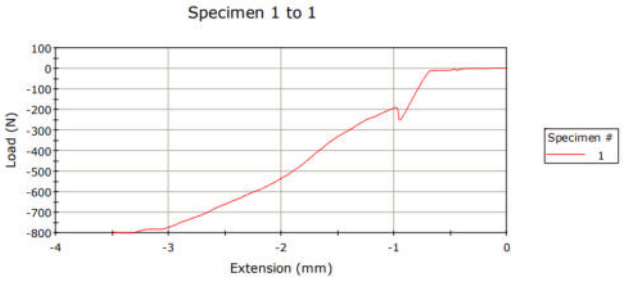


Figure 4.22. Load to Failure (N) test graphic of CC1402

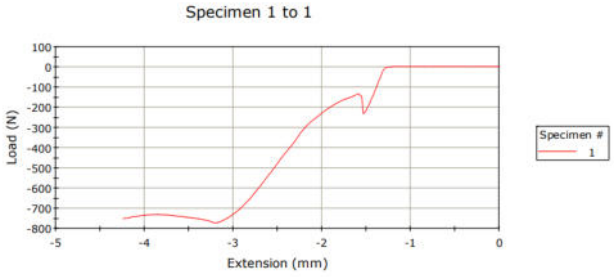


Figure 4.23. Load to Failure (N) test graphic of CC1403

-It was observed that CC1401 sample didn't decement totally and reciprocated more than 700 N of vertical load. But after the first testing it was visible that the distal abutment was deformated. Despite the deformation, we could cement two more specimens onto the main model.

-By the second sample, total decementation of the distal abutment started to occur between the vertical load of 200-300 N.

4.5. Statistical Comperative Results

Table 4.3. Evaluation of the joint effect of material and cantilever length on decementation values (N)

Decementation Values (N)	**Type III Sum of Squares**	**df**	**Mean Square**	**F**	**p**
Material	21740,83	2	10870,42	2,191	0,122
Cantilever Lenght	42135	1	42135	8,494	0,005*
Material * Cantilever Lenght	8122,5	2	4061,25	0,819	0,446

Two-way ANOVA Test *$p<0.05$

There was no statistically significant difference between material groups in terms of decementation values (p:0.122; p>0.05).
There was a statistically significant difference in decementation values between cantilevers (p:0.005; p<0.05).
The joint effect of material and cantilever length on decementation values is not statistically significant (p:0.446; p>0.05). Material and cantilever length together do not affect decementation values.

Table 4.4. Evaluation of the joint effect of material and cantilever length on decementation values (N)

	PEEK	**PEKK**	**BİOHPP**	
Decementation Values (N)	**Ort±SS**	**Ort±SS**	**Ort±SS**	**p**
14mm	147,5±102,45	120,5±74,55	180±67,91	0,292
21mm	70,5±35,15	99±61,77	119,5±63,62	0,155
p	0,046*	0,492	0,055	

Two-way ANOVA Test *$p<0.05$

There was no statistically significant difference between the 14 mm cantilever material groups in terms of decementation values (p:0.292; p>0.05).

There was no statistically significant difference between the 21 mm cantilever material groups in terms of decementation values (p:0.155; p>0.05).

When PEKK material is used; There was no statistically significant difference between the decimentation values of 14mm and 21mm cantilever (p:0.492; p>0.05).

When BIOHPP material is used; There was no statistically significant difference between the decimentation values of 14mm and 21mm cantilever (p:0.055; p>0.05).

Table 4.5. Evaluation of the joint effect of material and cantilever length on deformation values (N)

Deformation Values (N)	**Type III Sum of Squares**	**df**	**Mean Square**	**F**	**p**
Material	7725,833	2	3862,917	1,111	0,337
Cantilever Lenght	219615	1	219615	63,146	0,001*
Material * Cantilever Lenght	3197,5	2	1598,75	0,46	0,634

Two-way ANOVA Test *$p<0.05$

There was no statistically significant difference in deformation values between material groups (p:0.337; p>0.05).
There was a statistically significant difference in deformation values between cantilever lengths (p:0.001; p<0.05).
The joint effect of material and cantilever length on deformation values is not statistically significant (p:0.634; p>0.05).
Material and cantilever length together do not affect the deformation values.

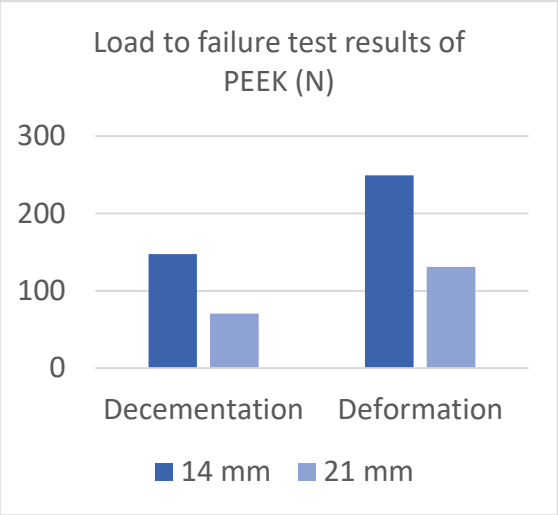
Table 4.6. Evaluation of the joint effect of material and cantilever length on deformation values (N)

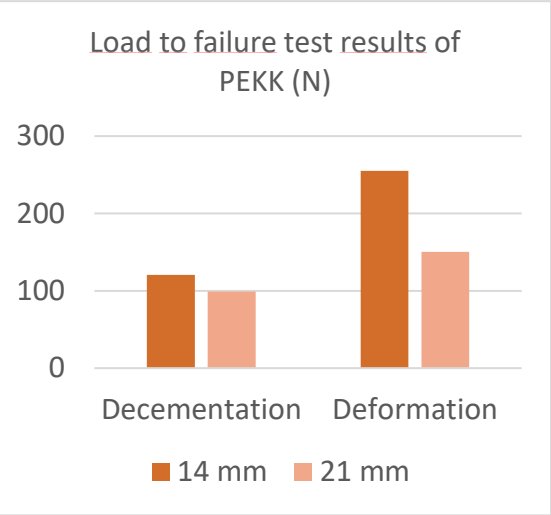
Deformation Values (N)	**PEEK** **Ort±SS**	**PEKK** **Ort±SS**	**BİOHPP** **Ort±SS**	**p**
14mm	249,5±68,82	255,0±54,57	288,0±97,16	0,478
21mm	131,0±26,65	150,5±39,75	148,0±37,73	0,415
p	0,001*	0,001*	0,001*	

Two-way ANOVA Test **p<0.05*

There was no statistically significant difference in deformation values between the 14 mm cantilever material groups (p:0.478; p>0.05).
There was no statistically significant difference in deformation values between 21 mm cantilever material groups (p:0.415; p>0.05).

When PEEK material is used; the average deformation with the 14mm cantilever were statistically significantly higher than that in the 21mm cantilever (p:0.001; p<0.05).
When PEKK material is used; The average deformation in the 14mm cantilever was statistically significantly higher than that in the 21mm cantilever (p:0.001; p<0.05).
When Bio-HPP material is used; The average deformation in the 14mm cantilever was statistically significantly higher than that in the 21mm cantilever (p:0.001; p<0.05).





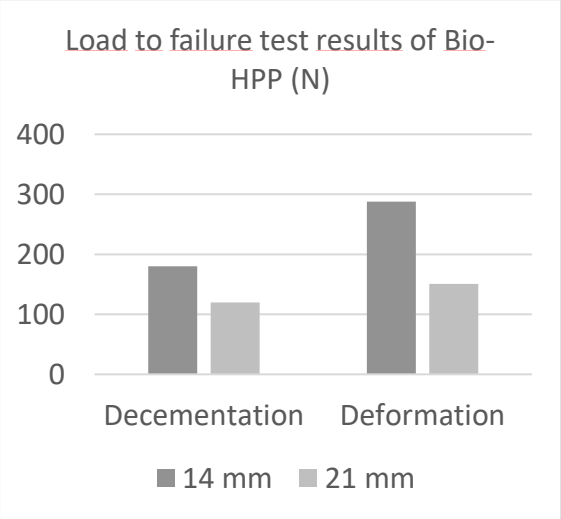
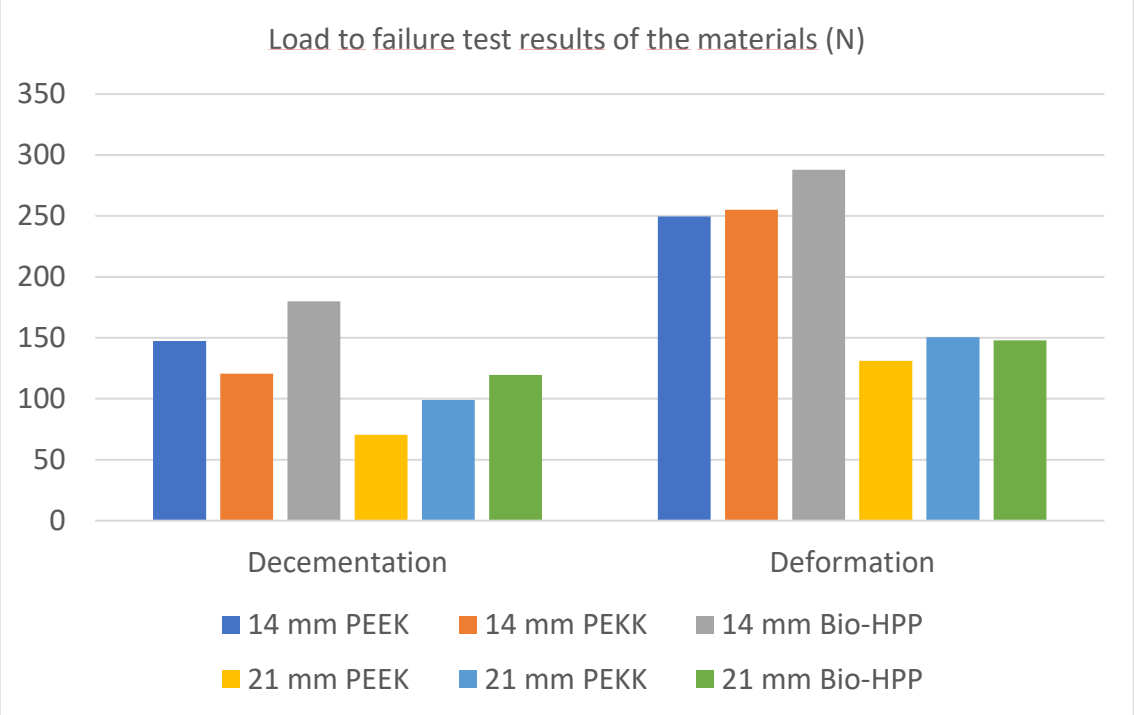


Fig 4.24. – 4.27. (Above right) Load to failure test results of PEEK (N), (Above left) Load to failure test results of PEKK (N), (Middle) Load to failure test results of Bio-HPP (N), (Below) Load to failure test results of the materials (N)



5. DISCUSSION

The treatment option of cantilever fixed prosthesis stands in a wide range of fixed prosthesis types from total edentulism and all-on-x concepts to 2-units anterior aesthetic region bridges, therefore the main goal of this study was to investigate through a large common field which the information was mostly scarce around newly introduced polymers and their derivatives (175). Computer-assisted design and computer-assisted manufacturing (CAD-CAM) high performance polymers (HPPs) are recently marketed for the fabrication of dental prostheses. Commonly used HPPs are made of polyaryletherketone (PAEK) materials known as polyetheretherketone (PEEK), (PEKK) and PEEK based materials. HPPs are used in prosthetic dentistry as framework materials for removable partial dentures, implant-supported fixed prostheses (ISFPs), (3-unit fixed partial denture (FPD), endocrown and resin-bonded FPD. It has been reported in previous studies that HPP materials have favorable mechanical and biological properties but most of the data comes from the manufacturers and more studies should have been on this area for truthfull support of these materials.

The most common formulations of the PAEKs are: Unfilled, pure 100% PEEK (eg Coprapeek, White Peaks Dental Systems GmbH & Co. KG., Essen, Germany), 80% PEEK with 20% nanoceramic filler (eg, BreCAM.BioHPP, Bredent GmbH, Senden, Germany) and 80% PEKK with 20% filler including titanium dioxide (eg, Pekkton Ivory, Cendres and Mettaux, SA, Biel/Bienne, Switzerland) (175).

The manufacturer claimed shock-absorbing behavior of PAEKs may make them favorable for immediate loading or long-term restorations. PEEK has low elastic modulus (4 GPa) closer to bone when compared to different framework materials. This may result in a shock-absorbing effect that may make PEEK a suitable material for implant prostheses in load bearing areas (175).

Low elastic modulus may be a chance while using these materials as dental implants but using them as superstructure comes along new requirements like veneering complications, and aesthetic suitability. Many load to failure tests, including veneered, unveneered, monolithic and infrastructure, have been conducted with high performance polymer derivatives belonging to the PAEK family, especially PEEK (180,181). While the load to failure values obtained from these tests support the use of polymers as temporary prostheses, resembling a much better alternative compared to conventional

temporary materials, it is underlined that they may cause complications if used as fixed prostheses (175,176).

In load to failure tests, if the test material animates a fixed prosthesis, in in vitro tests vertical directional forces are evaluated according to chewing forces measured in clenching and bruxism behaviors (175). Gibbs et al. reported a mean maximum clenching force of 462 N (range, 98 to 1031 N) in adults with tooth loss. In participants without tooth loss, the mean clenching force was 720 N and ranged between 244 and 1243 N (176). Braun et al reported the average clenching strength as 738 N, which ranged between 342 and 1280 N (179). Chong et al. measured the average occlusal forces in 20 healthy elderly and young adults and reported that the average occlusal force of elderly adults was 420.5 N and 541.4 N in young adults (28). According to these in vitro studies of different cohorts of patients and in vitro experiments, results caused by deformation is a common prosthetic complication and material selection is important to minimize the complications, especially the likelihood of fracture when parafunctional habits are considered.

According to manufacturer's instructions, 5 minutes of setting time was enough for the cementation process to complete. However, while cementing the specimens to the main model, we tried to wait for 15 minutes after the second specimen and the decementation values instantly got higher. So cementation time of the resin reinforced cement used during tests can be 15 minutes to be minimum.

One of the manufacturer's assertions was Bio-HPP having superior mechanical properties than PEEK because of the %20 ceramic filler particles, although there were no significant difference statistically between PEEK and Bio-HPP frameworks, during the experiments it was observed that Bio-HPP displayed higher decementation and deformation resistance over PEEK.

The cement of choice of this study was not the first choice of adhesion material of PAEKs but due to Cr-Co group and repeated action ease, was used. In the future studies, resin bonded adhesion may evoke more realistic results by eliminating low adhesion of conventional cementation.

In this study, "The amount of load arm extension (mm)" data were calculated from the graps recorded. However, after the experimenting phase is over we discovered that, the device could also collect time/extension data seperately as well as the graphs. In the future studies this feature can be used to evaluate accuracy on flexural behaviours of the materials in in vitro situations. The amount of load arm extension (mm) datas had high

variation, hence didn't join the statistical analysis. But it was observed that highest amount of load arm extension (3,0 mm) was seen with 21 mm cantilever PEEK samples.

During the experiments, it was observed that the slowest deformation occurred with Cr-Co control group and fastest deformation occurred with PEKK samples. With Cr-Co samples, it was observed that vertical loads above 700 N cause deformation on titanium abutments (Fig 4.20.,Fig 4.21., Fig 4.22.). Fast setting cement broke into pieces before the experimenting is over during Cr-Co phase. In the next studies another choice of material to fix the components to the testing table might beneficial.

Shackleton et al. reported that the survival rate was significantly better for prosthesis with cantilever lengths of 15 mm or less, compared with cantilevers longer than 15 mm. Other authors state that a length of 10–20 mm is acceptable, depending on the bone quality in which the implants are anchored (183,184,185). It has been understood from this study that, with the materials used, the 14 mm cantilever extensions are more acceptable for implant supported prosthesis framework than 21 mm.

Previous studies have compared zirconium, titanium and PEKK materials and they showed that when compressive stresses are dominant PEKK's shock absorbing behavious are limited (186). Within the limitations of our study, we also confirmed that even with 14 mm cantilever lenght, PEEK, PEKK or Bio-HPP (mean maximum deformation values respectively; 249,5 N, 255 N, 288 N) are displaying risky when used as implant supported prosthesis framework material

Cosme et al. reported the mean maximum bite force values and standard deviations for the young dentate adults (Age, range, 20-38) as 859 ± 304 N for nonbruxers and 806 ± 282 N for bruxers and (187). According to the results of the most current study, PEEK, PEKK and Bio-HPP materials could withstand mean loads of 1698.61 N, 738.19 N and 1272.25 N respectively (10). PEKK material with Ti base had load-to-failure values (738.19 N) closest to the mean maximum clenching forces reported in the previous studies (Braun et al,32 738 N, range, 342-1280 N, Gibbs et al,33 720 N, range, 244-1243 N) (10). PEKK showed more advantadgeous results over PEEK and Bio-HPP in the previous studies. Another study reported that fracture loads of 3-unit reinforced PEEK FDPs without a veneer and fabricated by different methods ranged between 1738 N and 2354 N (197). The most recent study showed that BioHPP frameworks fractured at a mean load of 1518 ±134 N. These outcomes conflicts our results in the study, the reason by standing too high to what we found, this may come from the varying testing environment, cementation process and test design.

In our study, Bio-HPP [80% PEEK with 20% nanoceramic filler (0.3-0.5 μm)] showed higher load-to-failure values among the other HPP groups with 14 mm cantilever however had lower load-to-failure values than the maximum clenching and bite forces reported. Within the limitations of this study it can be said that frameworks from Bio-HPP can be the most favorable material to use (BH14 ± 288 N, BH21 ± 148 N). While making frameworks out of this material, cantilever lenght should not be chosen as 21 mm. It is seen that 14 mm cantilever lentgh can be more resistant to decementation and deformation.

The second most suitable material was determined to be PEKK in this study (PK14 ± 255 N, PK21 ± 150,5 N). All of the PEKK samples fractured during the test. Although PEKK appeared on the dental market as a fiber reinforced version of the PEEK material, it has gained a brittle feature due to the fillers in its structure (80% PEKK with 20% filler including titanium dioxide) (10). No matter how the intraoral vertical chewing forces are depicted in this study, it is obvious that these forces will not progress as aggressively in the patient's mouth as in the experiment. However, lower deformation values must taken into consideration while choosing this material for framework manufacture. If PEKK material is going to be chosen for fixed dental prosthesis framework, desired cantilever lenght should be preferably 14 mm than 21 mm.

Among these three polymeric materials PEEK is the only material group that was studied the most at the past. Prechtel et al. suggested the use of PEEK as FDP material (188). Bathala et al. reviewed the use of PEEK as material and suggested the use 5 prosthetic situations that included 15% of use in fixed dental prosthodontics (189). Sinha et al. reported a successful use of PEEK FDP over 6 months (190). In our study PEEK displayed the least favorable load-to-failure results among Bio-HPP and PEKK (PE14 ± 249,5 N, PE21 ± 131 N). Cantilever lenght should be 14 mm over 21 mm.

The grayish or whitish color and the low translucency of PEEK, Bio-HPP and PEKK still limit their use as a monolithic, anatomic contour dental restoration material. Thus, additional veneering is required to obtain satisfactory esthetics (190). In general, the mechanical retention of the veneering composite resin depends on its viscosity and therefore on the weight percentage of the filler content. An increase in particle content is associated with an increase in viscosity. The chemical composition and the low surface energy of PAEKs may lead to difficulties for bonding to resin-based materials (191). There are still several studies being conducted in the area of veneering of PAEKs yet so far the most favorable choice is methyl methacrylate (MMA) based adhesive systems as

well as surface treatments (192). It is found that etching the PEEK surface prior to conditioning with MMA based primers and veneering increases surface free energy and surface roughness, as well as the tensile bond strength after water storage for 60 days (193). Nevertheless, PEKK and PEEK are different in their chemical structures, and currently there is very little evidence about the effects of different surface treatments on the bonding strength and durability of veneering materials to PEKK.

With Bio-HPP, after surface conditioning and MMA primer, material selection for veneering is also composite resins with varying filler percentadges like PEEK and PEKK. The recent studies are concentrating on PEKK about veneering protocols because of its superior mechanical properties. In our study, PEKK also demonstrated higher load to failure results than PEEK.

At the most current studies, while veneering PEKK, "the indirect laboratory composite resins" were the optimum choice of material showing much higher shear and tensile bond strenght over lithium disilicate ceramics (198).

If more studies happen to be conducted in this area, it may seen that low elastic modulus of PAEKs may resemble a disadvantadge for chipping and fracturing of the veneers because of the flexion of these polymeric materials however the evidence is still scarce on this subject. In our study, we could only observe the physical changes of the framework materials, however, it can be guessed that with these amount of flexibility, chipping of the veneering material is going to be inevitable after various veneering protocols. In the further studies including veneering process to the experiment might be beneficial to enlighten this area.

When choosing an implant-supported fixed prosthetic infrastructure material, if we were to choose a polymer-structured material with enough shock absorption ability to reduce the risk of peri-implantitis, our material choice would be Bio-HPP in the light of this study with 14 mm cantilever lenght. However the data is still scarce in this area and more studies should be conducted on mechanical and physical properties of these materials as infrastructures.

It was recommended to condition the surface with microabrasives before cementation, then prepare the surface with an MMA-based primer and bond it with an adhesive system. However, if the monolithic use of these materials, which can be produced in a very short time with CAD-CAM, can be made widespread by changing the filler ratios or adding various color codes, many risks of complications due to veneers can be eliminated. Because it seems like prosthetic complications have been reported to be

relatively often, with the most common being fracture of the veneering material (194,195,196).

6. CONCLUSIONS

In this in vitro study, four different cantilever fixed prosthesis material tested with vertical load to failure test. There were some limitations of this study. With the test method it was hard to mimic the intraoral area. There are various polymer CAD-CAM materials available in the field, and in this study, four of them got selected because they were the most used and preferred around previous studies that reviewed. In future studies, researches should be carried out to find a method that create a better adhesion between main model and framework specimens, hence the decementation resistance may incline. The number of studies comparing PEEK and Bio-HPP materials is few and more studies should be conducted on the material choice of fixed prosthesis. In order to measure the resilience of material, there should be more studies with fixed prosthesis scenarios on environments as close as possible to natural oral cavity.

In the current study, the conclusions can be summarised as follows:

1. Bio-HPP is the first choice of material for fixed dental prosthesis frameworks according to this study among polyaryletherketons like PEKK and PEEK.
2. The highest load to failure values were recorded from BH14 group.
3. According to load to failure test results, materials' mean maximum deformation values are respectively; BH14, PK14, PE14, PK21, BH21, PE21.
4. When 14 mm cantilever lenght is sufficant, the most favorable material will be Bio-HPP.
5. When 21 mm cantilever lenght is needed, the most favorable material will be PEKK.
6. PEKK is the most brittle polymer group among tested groups of PEEK and Bio-HPP.
7. Bio-HPP has higher rates of adhesion to titanium abutments than PEEK when cemented with resin reinforced glass ionomer cement.
8. Vertical loads of more than 700 N can cause deformation on titanium abutments if the framework material selection is Cr-Co.

The first null hypothesis was that there would be a statistically significant difference in deformation and decementation values between material groups. The first null hypothesis was wrong.

The second null hypothesis was that there would be a statistically significant difference between cantilever lengths in terms of deformation and decementation values. The second null hypothesis was right.

The third null hypothesis was that the joint effect of material and cantilever length on deformation and decementation values would not create a statistically significant difference. The third null hypothesis was right.

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8. CURRICULUM VITAE

Personal Information

Name	İlayda	**Surname**	Aşkan

Education

Degree	**Department**	**The Name of the Institution Graduated From**	**Graduation Year**
PhD	Prosthodontics	Yeditepe University	2023
University	Dentistry	Marmara University	2016
High School		Burak Bora Anatolian High School	2010