1940-nm THULIUM FIBER LASER CERAMIC BRACKET DEBONDING

by

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B.Sc. in Physics Kocaeli University ,2009

Submitted to the Institute of Biomedical Engineering
in partial fulfillment of the requirements
for the degree of
Master of Science
in
Biomedical Science

Boğaziçi University
2012

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DATE OF APPROVAL: 13 June 2012

ACKNOWLEDGMENTS

First of all , it is a pleasure to thank the people who made this thesis possible. I would like to express my gratitude to Assoc . Prof . Murat Gülsoy for making this Master's Thesis available as my supervisor and for his invaluable instructive comments,encouraging guidance every time and useful suggestions on this study . This thesis would not be possible without his kind support and the remarkable patience .

I owe my appreciate to members of my committee, Yrd. Doç. Bora Garipcan and Prof. Dr. B. Cem Şener for allocating their time for critical reviewing of my thesis.

I would also like to thank Ayşe Sena Kabaş Sarp for her invaluable guidance and help on issues related to this thesis . I am greatful to her for sharing her experiences and for her support throughout my MSc study . I wish her the best of luck .

This work was supported by Boğaziçi University Research Fund (Project No : 5891-Murat Gülsoy) . I would like to thank Boğaziçi University for their support .

This study would not have been possible without the assistance of my dear friends in photonic laboratory Nermin Topaloğlu , Ayşegül Şen , Burcu Tunç . It would be very difficult to handle experiments without their supports .

Finally , I am especially indebted to my parents Uğur Demirkan , Kader Demirkan and my grandmother Emine Çetinbaş , and my friend Burak Direk for all their love , encouragement and endless patience during my study .

I would like to dedicate my thesis to my beautiful mother and my father .

ABSTRACT

1940-nm THULIUM FIBER LASER CERAMIC BRACKET DEBONDING

The aim of the study was to determine the proper laser paramaters for 1940-nm Thulium Fiber Laser for ceramic bracket removing. In order to assess the effectiveness of 1940-nm Thulium Fiber Laser in orthodontic ceramic bracket debonding , polycrystalline ceramic brackets were bonded to mandibular bovine teeth with adhesive agent.The samples were divided into 9 different groups due to applied laser power and laser duration, debonding method used . There was a control group that had no laser application . The efficency of the laser was investigated together with the required debonding forces and intrapulpal temperature changes . In this study, keeping intrapulpal temperature changes below the threshold value that is accepted $5,5\,^{\circ}\mathrm{C}$ must be accepted as a must. In most of the lasing groups , the increase in intrapulpal temperature changes were observed almost below the threshold value $5,5\,^{\circ}\mathrm{C}$. The findings revealed that 1940-nm Thulium Fiber Laser irradiation could reduce the needed debonding force or SBS (shear bond strength) values significantly compared to control group . Irradiation of the specimens by 1940-nm Thulium Fiber Laser caused more than 50% reduction in the needed debonding force when compared to the control group. Different application methods : non-scanning and scanning were studied to assess the effects of the distinct configurations. Scanning method was tried to reduce the intrapulpal temperature rise during laser irradiation but in this study side effects of this method were faced. It was revealed that different application methods did not create any remarkable differences . In more than 50% of samples with energies 25 J or more, adhesive remnant hasn't been observed on enamel surfaces for the laser groups.

Keywords: Laser, Debonding, Ceramic Brackets

ÖZET

1940-nm TULYUM FİBER LAZERİ İLE SERAMİK BRAKETLERİN ÇIKARILMASI

Çalışmanın amacı seramik braket çıkarılmasında , 1940-nm Tulyum Fiber Lazer için uygun lazer parametrelerinin belirlenmesidir. 1940-nm Tulyum Fiber Lazerin seramik braket çıkarılmasındaki etkisini değerlendirmek için , polikristal seramik braketler sığır dişleri üzerine dental yapıştırıcı madde ile yapıştırılmıştır. Örnekler her birinde 11 örnek olacak şekilde uygulanan lazer gücü ve lazer süresine ve lazerin uygulama yöntemine göre 9 farklı gruba ayrılmıştır. Gruplar arasında, lazer uygulamasının yapılmadığı kontrol grubu da bulunmaktadır. Lazerin verimi gerekli olan çekme kuvveti ve sıcaklık değişimi açısından değerlendirilmiştir. Bu çalışmada , pulpa sıcaklığını kabul edilmiş olan eşik değerinin altında tutmak şart olarak kabul edilmiştir. Lazer gruplarının çoğunda , pulpadaki sıcaklık artışı eşik değerinin neredeyse altında gözlemlenmiştir . Sonuçlar 1940-nm Tulyum Fiber Lazer ışımasının gerekli olan çekme kuvvetini kontrol grubuyla karşılaştırıldığında önemli boyutta azalttığını açığa çıkarmıştır. Örneklerin 1940-nm Tulyum Fiber Lazer ile ışınmasında , kontrol grubu ile karşılaştırıldığ ında gerekli çekme kuvvetinde 50% den daha fazla azalmaya sebep olmuştur. Farklı methodların etkisini değerlendirmek için tarama uygulanan ve tarama uygulanmayan iki farklı uygulama methodu çalışılmıştır. Tarama methodu pulpadaki sıcaklık artışını azaltmak için denendi, ancak bu çalışmada uygulamanın olumsuz etkileriyle kar şılaşıldı . Farklı methodların uygulanması önemli bir fark yaratmamıştır. 25 J ve daha fazla enerjiye sahip olan laser gruplarında, mine yüzeylerinde yapıştırıcı madde gözlemlenmemiştir .

Anahtar Sözcükler: Lazer , seramik braketlerin çıkarılması , seramik braket .

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LIST OF ABBREVIATIONS

LASER	Light Amplification by the Stimulated Emission of Radiation
nm	nanometer
°C	Celcius Degree
CW	Continuous Wave
FDA	United States Food and Drug Administration
CO_2	Carbon dioxide
Nd:YAG	Neodymium-doped Yttrium Aluminium Garnet
KrF	Krypton Fluoride
Tm:YAP	Thulium Ytterbium Aluminum Phosphate
GaAlAs	Gallium Aluminum Arsenate
Cl	Chloride
F	Florine
K	Potasyum
Na	Sodium
$Ca_{10}(PO_4)_6(OH)_2$	Hydroxyapatite
μm	micrometer
Bis-GMA	Bisphenol-a and methylmethacrylate
4-META MMA	4-methacryloxyethyl trimellitate anhydride
mm	millimeter
NIR	Near Infra-red
N	Newton
DNA	Deoxyribonucleic Acid
W	Watt
sec	second(s)
ARI	Adhesive Remnant Index
kPa	kilopascal
cm^2	centimetersquare
mm^2	millimetersquare

MPa	Megapascal
cm	centimeter
s	second(s)
kg	kilogram
J	Joules
SEM	Scanning Electron Microscope
A	Ampere
US	United States

1. INTRODUCTION

1.1 Motivation and Objectives

In today's world , people are very sensitive about the health of their teeth. They want to have better apperance for good impression. In order to have beautiful smile and straight sequence of the teeth, orthodontic treatment is an effective way. Orthodontics help people to have beautiful bite and it also provides better resistance to intra-oral diseases. One of the most common problems is tooth irregularity or improper sequence of the teeth. In today's orthodontics , irregularity of tooth is solved by using bracket treatment . In the treatment with orthodontic braces , wires are used. Brackets are also the part of the dental braces and they are bonded to each tooth. One of the types of orthodontic brackets that are used is ceramic bracket. When compared to other types of orthodontic brackets , patients and orthodontists prefer ceramic brackets because of their superior esthetics. Ceramic brackets also provide higher strength [1] . After orthodontic treatment , they need to be removed from the enamel surface . There are different techniques have been suggested to debond ceramic brackets. If conventional method is used for debonding of the ceramic brackets, enamel tear outs, bracket failures and pain can be encountered during conventional ceramic bracket removing techniques [2] . Also conventional debonding method may damage the enamel surface and be time consuming . Damage on the enamel surface can cause poor esthetics and put at risk of long term diseases of the affected tooth [3, 4] . During conventional debonding ,especially ceramic brackets create plenty of problems. Thus, application of the laser in order to debond the brackets may be a new method to overwhelm these drawbacks that may be encountered during debonding.

Laser application in dentistry has been developed since the generation of first ruby laser . In dentistry , lasers have been using for surgery of the soft tissues in the oral cavity [5] , etching of the enamel surface , bleaching , tooth drilling , removing of hard dental tissues and debonding of ceramic brackets . Procedures in laser dentistry do not need sutures and anesthesia . Also , it minimizes bleeding and bacterial infections . In orthodontics , dental lasers help to eliminate the enamel crackes during debonding and bracket wing breakage. All these given procedures and purposes can be accepted as advantages. The use of lasers was introduced into dentistry as different kinds were approved by the United States Food and Drug Administration during the 1980's and 1990's [3] . Since the early 1990's, lasers have been used experimentally for debonding ceramic brackets. In order to debond ceramic brackets,different types of lasers have been used like continously running Nd:YAG (1060 nm), KrF (248 nm), XeCl (308 nm) Tm:YAP (1980 nm) and GaAlAs (808 nm). Thulium fiber laser (1940 nm) is choosen for this study because it has lots of advantages of fiber lasers over other types of dental lasers. Fiber lasers are easier to use because they have extended lifetime and compact size. Fiber lasers show high vibrational stability. Light is already coupled into a flexible fiber it means that the light is already in a fiber allows it to be easily delivered to a movable focusing element. Moreover, fiber lasers can have active mediums several kilometers long, and so can provide very high optical gain. So, fiber lasers have high output power [6]. On the other hand, thermal effect on the pulp on the vital tissue and surface of the enamel is one of the most important problems that are faced during laser-aided debonding,because of the fact that while applying laser energy to the bracket, this energy is converted into heat. If temperature rise exceeds of $5,5\,^{\circ}\mathrm{C}$, thermal pulpal damage is observed. Up to $5,5\,^{\circ}\mathrm{C}$ it is reversible. Excess of the temperature to $11\,^{\circ}\mathrm{C}$ may cause intrapulpal necrosis, it is irreversible . In $45\,^{\circ}\mathrm{C}$ temperature,patient feels the pain [7]. Therefore, bracket removal needs careful attention and appropriate laser parameters. In this study, keeping intrapulpal temperature changes below the threshold value must be accepted as a must. That is why $5,5\,^{\circ}\mathrm{C}$ was accepted as a threshold temperature in this study and intrapulpal temperature changes during ceramic orthodontic bracket removal were observed. Motivation of the proposed study is to reduce the enamel damage that can be faced during bracket debonding procedure. Previously untested 1940 nm Thulium fiber laser is used in this study,thus from this study we aim the following items below:

1)Eliminating the enamel tear outs , bracket failures and the pain that are encountered during conventional ceramic bracket removal techniques [2] .

2) Determination of the suitable laser parameters for debonding ceramic brackets,keeping that intrapulpal temperature changes below the threshold value.

1.2 Scope of The Thesis

In the proposed study , previously untested 1940-nm Thulium Fiber Laser is used in order to remove orthodontic polycrystalline ceramic brackets from the tooth surface. This laser is preferred because of the benefits of fiber lasers over other types of dental lasers. Fiber laser are easier to use and they have extended lifetime and compact size. In addition , they can have active regions several kilometers long and thus can provide very high optical gain. 1940-nm Thulium Fiber Laser is choosen for this study because the power of this laser can be controlled and adjusted easily . Two different application methods are tested in this study : non-scanning and scanning application methods . By trying the scanning applied laser irradiation method , our aim is to reduce the intrapulpal temperature change increase during laser-aided bracket debonding . During the experiments, intrapulpal temperature changes are controlled and recorded at real time by using K-Type thermocouple. $5,5\,°C$ is accepted as the threshold value for difference in intrapulpal temperature in order to prevent irreversible effects of laser irradiation . Application of laser irradiation , measurement of both needed debonding forces to remove ceramic brackets by universal testing machine and intrapulpal temperature changes during laser irradiation are studied at the same time in the proposed experiment . In this study , application of both debonding force and laser irradiation to the samples are studied at the same time with the aim of minimizing unnecesseary heat transfer from the ceramic bracket to the tooth. Bracket base assessment is done to observe the effects of the 1940-nm Thulium Fiber Laser irradiation on the enamel surface of the samples.

1.3 Outline

Part 2 gives general knowledge about dentistry , orthodontics ,dental lasers and laser-hard tissue (enamel) interactions. In addition , methods that are used for bonding and removing of the orthodontic brackets are explained. Part 3 gives particular information about materials that are used in the experiment and also methods in the proposed study are given . Part 4 contains findings and the results of the proposed study . In Part 5 , discussion of the present study is given. Part 6 gives the conclusion and further works of the given study.

2. BACKGROUND

2.1 Dentistry

Dentistry diagnoses and treats problems with a patient's teeth, gums, and other regions of the mouth. Dentistry is widely considered necessary for complete overall health. It is a branch of medicine that contains diagnosis, prevention, and treatment of any disease of teeth, oral cavity, and associated structures. The oral cavity represents the first part of the digestive tube. The oral cavity is anteriorly surrounded by lips, the cheeks laterally, the floor of the mouth inferiorly, the oropharynx posteriorly, and the palate superiorly [8, 4] . The oral cavity is properly bounded by the alveolar arches, teeth and gums, and palate and tongue. The oral cavity is oval shaped and it consists of two portions, the vestibule and the oral cavity proper (lingual) .The bony base of the cavity is shown by the maxillary and mandibular bones (Figure 2.1) .

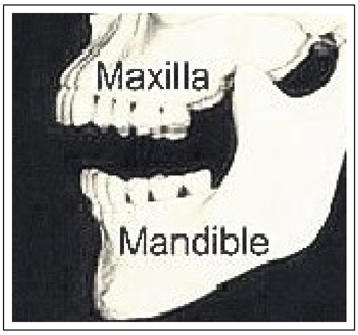

Figure 2.1 The Bony Base Of The Cavity.

In the oral cavity , the upper jaw is named as maxilla and the lower jaw is called mandible. Maxillary teeth are the teeth of the upper arch since the roots of the maxillary teeth are embedded within the alveolar process of the maxilla. The lower arch ones are called mandibular teeth, because the roots of them are embedded within the alveolar process of the mandible. The teeth are classified as incisors, canines, premolars, and molars. The eight incisors are used to cut food by their edges. The four canines ("eye-teeth") supports in cutting. The eight premolars ("bicuspids") assist in crushing food. The deciduous molars are replaced by them. The twelve molars crush and grind food (Figure 2.2) .

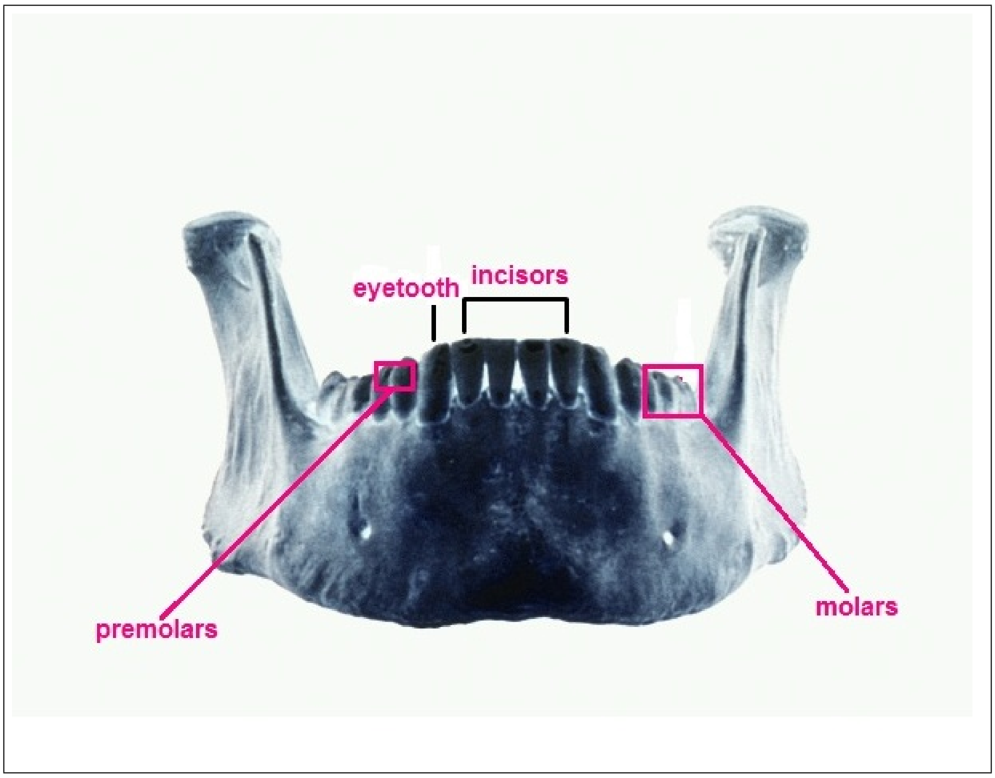

Figure 2.2 Frontal view of mandibula and the location of molars,premolars,incisors and eyetooth on the mandibula.

The imaginary plane which is accepted in the center dividing the dental archright from left is called Median sagittal plane. Median line is the imaginary line on that plane that bisects the dental arch at the center. Mesial means towards to anddistal means away from the center (median) line of the dental arch. Mesial surface is toward and distal surface is away from the midline. Facial means toward the cheeks orlips. Labial is facial surface of anterior teeth (toward the lips) and buccal means facialsurface of anterior teeth (toward the cheeks). Lingual is toward the tongue. Occlusal is the biting surface; that surface that articulates with an antagonist tooth in an opposing arch. Incisal is cutting edge of anterior teeth and apical is toward the apex, the tip of the root (Figure 2.3) [9, 10].

A tooth is constituted of four dental tissues: Enamel, dentin and cementum, which are hard (calcified) and pulp, that is soft (noncalcified) . The visible part of the tooth is called the crown. It is made of enamel. Enamel is mostly made of calcium phos-

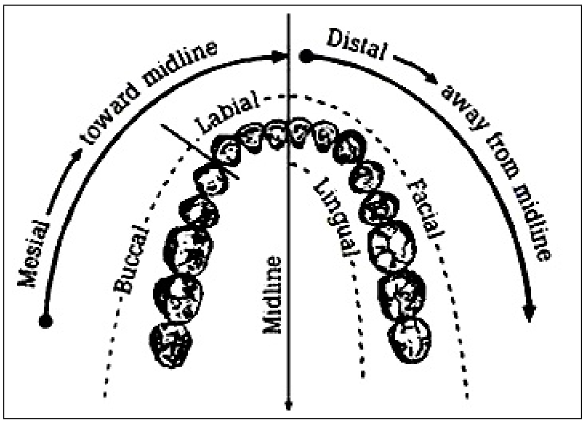

Figure 2.3 Midline, Mesial, Distal, Lingual, Labial, Facial, and Buccal terms are indicated [9] .

phate,a rockhard mineral. Enamel the hardest and most highly mineralized substance in the human body. Dentin is a calcified tissue and a layer underlying the enamel in human body. It is the largest part of the tooth. When compared to the enamel, it has softer structure in tooth. It contains 70 % hydroxyapatite ($Ca_{10}(PO_4)_6(OH)_2$), 20% organic matter and 10 % water. Dentin is also more sensitive to cold and hot. Organic and inorganic components show different amounts in dentin when compared to the enamel. In the composition of enamel , it stores 95 % hydroxyapatite ($Ca_{10}(PO_4)_6(OH)_2$) , 4 % water and 1% organic matter. The chemical represantation of hydroxyapatite is given by the chemical formula ($Ca_{10}(PO_4)_6(OH)_2$) . Cementum is hard connective tissue and bony material. It covers the tooth root and gives attachment to the periodontal ligament. Peridontal ligament is the tissue that supports hold the teeth firmly against the jaw. Root is covered by cementum and the part of the tooth embedded in the alveolar process. The end of root tip is apex and apical foramen is the opening at the root tip. Pulp is softer compared to the other parts of the teeth.It is located inner structure of teeth. Pulp contains blood vessels and nerves. In addition to that ,it includes connective tissues. Connective tissues help interchange between pulp and dentin. When a person has a toothache,the pulp is what hurts. Tooth structure composition is not homogenous (Figure 2.4) [10] .

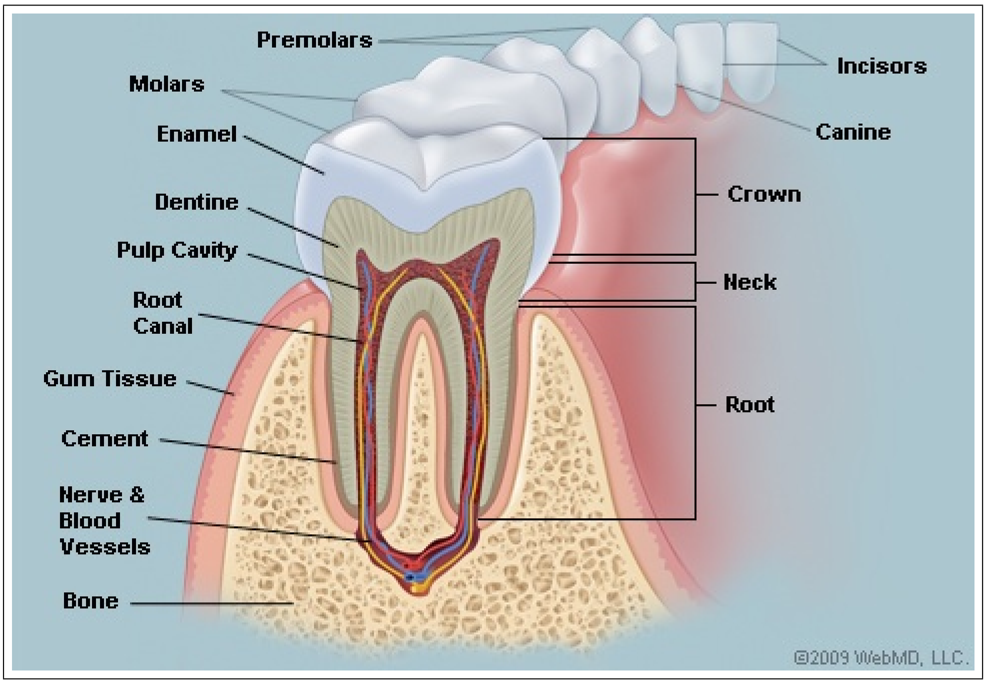

Figure 2.4 Tooth Anatomy [10].

2.2 Orthodontics

Orthodontics itself is a word that comes from Greek and it means to correct bad bite properly. In order to have attractive smile and straight sequence of the teeth,orthodontics treatment is an effective way. Orthodontic treatment can focus on dental displacement only, or can deal with the control and facial growth modification. In today's orthodontics, dental displacement is solved by bracket treatment (Figure 2.5) .

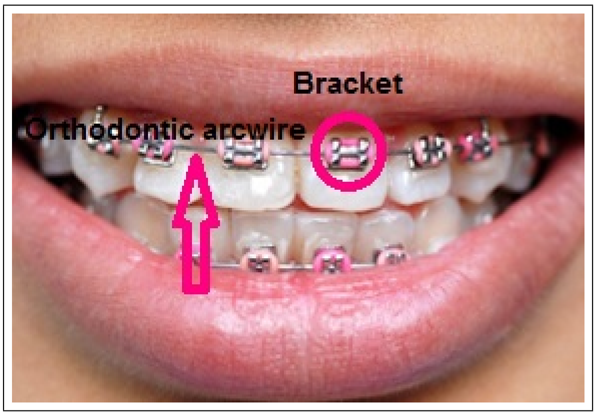

Figure 2.5 Bracket Treatment of Tooth Irregularity.

An uneven bite condition is known as a malocclusion. Malocculusion contains teeth that are crooked or crowded . An improper bite can interfere with speaking and chewing and can lead to problems with the jaws. Orthodontic treatment is put into practice because of aesthetic reasons with regards to having better appearance for good impression. Orthodontic procedures are most commonly done on children. In recent years, orthodontic procedures have been much more preferable by adults. However, teeth movement occurs slower in the orthodontic treatment of adults . Correct placement of the teeth can create a good-looking smile, but more importantly, orthodontic treatment results in a healthier mouth. Different kinds of orthodontic appliances may be used for different and various aims in an orthodontic treatment.

There are several different types of appliances used in orthodontics:

1. Dental braces are the most common type of orthodontic appliances. They are formed by sets of brackets bonded to the front of the each tooth.They are often used to correct improper bite such as underbites, overbites, cross bites and open bites, or crooked teeth. The kind and also continous pressure of orthodontic braces slowly moves and correctly repositions the teeth. Teeth move through the use of force. Dental braces require to controlled periodically by the orthodontist. They can be made of metal (gold, stainless steel, silver), plastic or ceramic material.

2. An orthodontic arch wire is the wire that attaches to dental braces.It is like the engine that moves and guides the patient's teeth. An orthodontic arch wire is needed for applying force in correcting irregularities in the placement of the teeth with braces. Orthodontic arch wires come in different sizes and have different material alloys. "Size" of an orthodontic arch wire, refers the cross-section or thickness of the wire. There are four main types of material alloys for orthodontics arch wires : Stainless steel, nickel-titanium, beta-titanium and molybdenum.

3.Brackets are also the part of the dental braces and they are bonded to each tooth.

2.3 Bonding Orthodontic Brackets

The procedure of bonding orthodontic brackets on enamel has changed significantly in the last 30 years. This is due to the introduction of materials and methods that allow effective attaching of the orthodontic brackets directly to the enamel. Process of attaching orthodontic brackets on enamel surface is based on adhesion between two different materials. Adhesion can be defined as the debonding force between filling material and tooth structure when they are came into intimate contact. In an attempt to provide bonding or adhesion, adhesive is used that is the material to which it is applied is called the adherend .The bonding adhesives used to glue orthodontic bracket to enamel have improved tremendously over the years. The performance of all dental materials, whether ceramic, polymeric or metallic is based on their atomic structure [2] . Before bonding process to enamel, orthodontists must be sure that the enamel surface is clean and also dry, or else no attaching will be performed. A dry and clean region is very important because the materials used for bonding need a clean enamel surface.This certifies that the bonding material has the best possible chance of creating a complete attaching to the enamel. The presence on the surface of anything could be considered as a contaminant itself is weakly bonded to the solid and will prevent the adhesion of adhesive to substrate [11] . Adhesion may be divided into two mechanisms: mechanical and chemical. Chemical adhesion contains attaching or bonding at atomic or molecular level. Mechanical one is depended on retention by penetration of one phase into the surface of the other. In many cases, it is also possible to observe both chemical and mechanical attaching together. Penetration of the bonding material into microscopic or submicroscopic irregularities (i.e as pores and crevices) in the surface of the substrate by acid-etching method may be observed in mechanical adhesion. Bonding with composites has been done by etching tooth surface with phosphoric acid [12] . Acid etching principle is to simply clear microscopic amounts of enamel leaving pores and crevices. Characteristically, etching is achieved using phosphoric acid (34-37 %) . Capillary penetration into surface irregularities inspires attaching of resins to etched enamel. These projections of polymer into the enamel have been named as resin tags. Resin tags may penetrate 10 to 20 μm into the enamel porosities [13]. This micromechanical attaching mechanism has been commonly used in dentistry because of absence of truly adhesive cements or restorative materials [14]. A more recent example of mechanical debonding is that of resin restorative materials. The acid produces minute pores and other irregularities in the surface of enamel into which the resin subsequently flows when it is placed into the preparation. The greatest problems associated with bonding to enamel surfaces are the in adequate removal of etching debris and contamination by water or saliva [15] . According to their chemical features, dental adhesive materials that are used for orthodontic bracket adhesion may be distinguished into two types.They are both polymers and also categorized as acrylic or diacrylate resins. The acrylic resins are derivatives of ethylene and contain a vinyl group in their structural formula [16] .Chemical name of acrylic resin is "polymethylmethacrylate" . It is transparent and transmits light in the ultraviole range to a wavelength of 250 nm [9]. Depolymerization occurs between $125\pm$ °C and $200\pm$ °C . Approximately at $450\pm$ °C , almost 90% of the polymer depolymerizes to form the monomer [15] . Most diacrylate resins are based on the acrylic modified epoxy resin. One of the first methacrylates used in dentistry was Bis-GMA. Bis-GMA resin is described as the reaction product of bisphenol. It is used as a bond implant material and as the resin component of dental sealants. There is an important difference between first type resin and second type resin [17] . Some studies are done in order to compare the debonding adhesives. For example, in 1995 Mimura et al. [18] studied on the comparison of two bonding materials for laser debonding. The selected bonding agent in this study were 4-META MMA (4-methacryloxyethyl trimellitate anhydride) resin and Bis-GMA adhesive resin. In this study, it is observed that debonding force for MMA resin was sufficently at a lowest power of energy than required for Bis-GMA resin groups. On the enamel surface in MMA resin group, more adhesive remained compared to Bis-GMA samples. As a conclusion, they concluded that debonding MMA resin with a laser is safer than debonding Bis-GMA resin with a laser. Moreover,in Rueggeberg and Lockwood's study was on ten commercial brands of orthodontic materials representing three modes of delivery systems: Two paste, no mix and power liquid types [19] . Stainless steel orthodontics brackets were bonded on bovine teeth. During heat application to the brackets, each temperature at debonding were saved. They concluded that a higher temperature was observed for two-paste systems compared the no mix systems. Moreover, the power liquid types needed the lowest temperature. In orthodontic treatments, in order to align tooth irregularity,orthodontic brackets are used. They are very small and also used to attach an arch wire.It has two wings,a base and channel (thinnest part) for locating an archwire (Figure-2.7) . Orthodontic brackets are divided into three types: ceramic based brackets,plastic based brackets and metal based brackets (Figure-2.6) . Of these, most orthodontists prefer using metal brackets for routine treatments

Figure 2.6 (a) Physical appearance of orthodontic brackets.

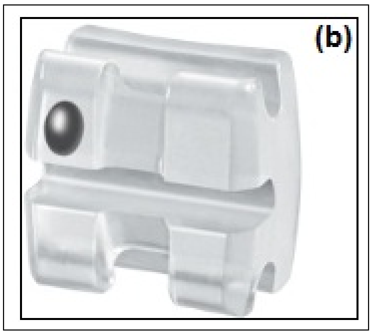

Figure 2.7 (b)Bracket Surface.

Metal brackets was introduced in the early 1970's. A few years later plastic brackets were used because of their esthetic apperance compared to the metal counterparts. In the mid-1980s,ceramic brackets were introduced into orthodontics. Ceramic brackets are more preferable because of the superior esthetics when compared to the metal brackets. Metal brackets rely on mechanical retention for bonding, and mesh gauze is the conventional method of providing this retention. Also, photoetched recessions are available. The area of the bracket base is not an important factor due to bond strength with mesh-backed brackets. The usage of less discernible, small metal bases helps beware irritation of gingival. The base should be designed to follow tissue contour along gingival margin . The base must not be smaller than the bracket wings, however, because of strength reasons and the danger of demineralization around the periphery (Figure-2.8) [9]. The brackets of mandibular molar and premolar must be kept away from occlusion, or the orthodontic brackets may easily come loose. Metal brackets' corrosion can be a problem and black and green spots have appeared with attached stainless steel brackets. Crevice corrosion of the metal arising in areas of poor bonding may result from the type of stainless steel alloy used [19, 20] . On the other hand, other factors such as galvanic action, bracket base construction, particular oral environment and thermal recyling of orthodontic brackets can be contributing factors. Because, the corrosion susceptibility of stainless steel interest is growing in the use of more corrosion-resistant and biocompatible bracket metals such as titanium [9] .

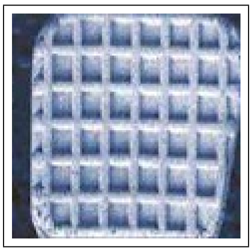

Figure 2.8 Mesh Base Design of Opti-MIM RÂ° bracket base [9] .

Plastic brackets (polycarbonate brackets) are made of polycarbonate and acrylic. They are mainly preffered for esthetic reasons. In 1965, according to Newman's report, they were non-resistant, being easily fractured or distorted. These accessories are not chemically resistant when in contact with solvents and, under high temperatures, allow migration of monomers away from the original products. Their lack of strength results in attachment problems,wing breakage and permanent deformation or creep. Polycarbonate bracket slots distorted with time under a stable physiologic stress rendering them insufficiently strong to withstand longer treatment times or transmit torque reported significantly higher torque losses and lower torquing moments with polycarbonate brackets compared to metal brackets [21] .

Ceramic brackets were introduced to orthodontics to meet the increasing demand for esthetic appearance and compensate for the lack of the strength and rigidity of the original plastic brackets in the mid 1980's. In the mid 1980's,the first ceramic brackets were made of monocrystalline [3] sapphire and polycrystalline ceramic materials become widely used. The first ceramic brackets were granulated from single crystals of sapphire (monocrystalline) using diamond tools and monocrystalline ones include a single crystal of aluminum oxide. They were closely followed by polycrystalline sapphire (alumina) ceramic brackets, which are manufactured and sintered using special binders to thermally fuse the particles together [21] . Unlike polycarbonate brackets, they resist staining and slot distortion and are chemically inert to fluids that are likely to be ingested. On the other hand, because of their inert aluminium oxide composition, they are not able to bond chemically with acrylic and diacrylate bonding adhesive materials. Ceramic materials are very rigid and brittle. Ceramic brackets provide higher strength,more resistance to wear and deformation,better color stability and, preferred for cosmetic reasons. Alumina is a typical member of modern ceramic brackets, formed when aluminum is added to steel to remove oxygen dissolved in the steel. It can be used as a single crystal material or as a polycrystalline material during production of the ceramic brackets. Ceramic brackets are machined from polycrystalline or single-crystalline (monocrystalline) aluminumoxide. In today's orthodontics, all currently available ceramic brackets mainly include aluminium oxide. The manufacturing process of ceramic brackets is a crucial aspect and plays an important role in the clinical performance. The production process of the single crystal brackets is more complicated compared to the production of polycrystalline ceramic brackets. In the manufacturing process of polycrystalline ceramic brackets,it is initiated with blending the particles with a binder. Then,this mixture is molded into a shape from which the critical parts of the brackets can be cut. The molded part is then fired at a temperature that permits the binder to be burnt out and the aluminum oxide particles to fuse but not melt. This process is called "sintering". It is relatively inexpensive and because of this property it is very popular manufacturing method. Unfortunately , this process causes structural imperfections at grain boundaries and the incorporation of trace amounts of inpurities.These slight imperfections and impurities may serve as foci for propogation of cracks under applied load or stress. So, all in all , bracket fracture can be observed. However , polycrystalline brackets are more readily available at present.Monocrystalline ceramic brackets are manufactured from aluminum oxide . In their manufacturing process , the oxide particles are melted and then slowly cooled by allowing proper crystallization.This manufacturing procedure reduces the stress-inducing impurities and imperfections found in the polycrystalline ceramic brackets [22] . Optical clarity is the most clear difference between polycrystalline and monocrystalline ceramic brackets. Single-crystal ceramic brackets are more diaphanous. Luckily, both of them set against staining and discoloration [23]. Ceramic brackets are famous for their hardness and their resistance to degradation at high temperature and to chemical degradation. Physical properties of ceramic brackets that are crucial to the orthodontics contains tensile strength, hardness and fracture toughness or brittleness [23]. The physical properties of them are a result of their atomic bonding. A very important physical property of ceramic brackets is the extremely high hardness of aluminium oxide. This property provides an important benefit to both single-crystal and polycrystalline ceramic brackets over stainless steel brackets. According to the study of Swartz et al.,ceramic brackets are nine times harder than stainless steel brackets or enamel [24]. Moreover,in the study of Viazis et al.,it was concluded that abrasion of enamel from ceramic brackets may occur rapidly, if contacts between teeth and ceramic brackets exist [25]. Tensile strength is another significant property of ceramic brackets. In monocrystalline alumina, the tensile strength is much more higher than in polycrystalline alumina, that is in turn significantly more than stainless steel. This property dependens on the condition of the ceramic bracket's surface . A shallow scratch on the surface of a ceramic bracket decreases the load needed for fracture. The elongation for ceramic at failure is less than 1% in contrast with approximately 20% of stainless steel, thus making ceramic brackets more brittle. In other word metal brackets deforms 20 In orthodontics,ceramic brackets have highly localized, directional atomic bonds. This oxidized atomic lattice does not allow shifting of bonds and redistribution of stress.If the interatomic bonds break and material failure occurs,it means that stresses reach critical levels.This fact is called "brittle failure". In ceramic brackets, fracture toughness is 20 to 40 times less than metallic ones, by making it much easier to fracture a ceramic bracket than metallic ones [26, 27] . Among all ceramic brackets, polycrystalline alumine shows higher fracture toughness than monocrystalline alumina. Monocrystalline brackets are not fractured easily [28, 29]. During debracketing,the brittle nature of ceramic brackets has resulted in a higher incidence of bracket failure [16]-[30]. Unlike metals ,ceramic compounds, are also susceptible to crack propagation caused by minute imperfections or material impurities. The fracture toughness of the ceramic is higher than that of enamel. Moreover, ceramic brackets bonded to rigid, brittle enamel have little ability to absorb stress [2]. During ceramic bracket debonding, enamel fracture is related to the high bond strength of ceramic brackets and associated with sudden impact loading. There are two significant problems that are stemmed from the combination of very hard and brittle properties and high bond strength. One of them is bracket failure during debonding and the second one is enamel failure which may occur during function but mostly during debonding. Ceramic brackets are radiolucent and if they are inhaled, they would not be visible on the radiograph. Ceramic brackets are esthetic, strong, and resistant to chemical degradation. However, the atomic structure that explains these advantages also accounts for the most obvious fault of ceramics, namely their brittleness and low fracture toughness. Due to their benefits, ceramic brackets also show some significant drawbacks. Ceramic is the third hardest material known to humans. Therefore,ceramic brackets in contact with the opposing teeth may lead wear of the softer enamel [22] , because of their inert aluminium oxide composition, they cannot bond chemically with both acrylic and diacrylate bonding adhesive agents. As a result, the early ceramic brackets used a silane-coupling agent to act as a chemical mediator between the ceramic bracket base and the adhesive material. This chemical retention resulted in extremely strong bonds that caused the enamel/adhesive interface to be stressed during debonding of ceramic brackets, risking irreversible enamel damage in the form of crack and delamination that often needed dental restorations. As a result, the challenge was to develop a bond between the ceramic bracket base and the enamel that clinically has satisfying strength to accomplish treatment but can be broken at debond without any damage to the enamel surface. The majority of the currently available ceramic brackets rely solely on mechanical retention, using standard light or chemically cured adhesives, without the need for additional special adhesive materials. Numerous mechanical base designs are now available ranging from microcrystalline, mechanical ball, dovetail, dimpled chemo/mechanical, silane coated buttons and polymeric bases with many manufacturers claiming consistent bond strengths and debonding characteristics comparable to that of stainless steel mesh. There are lots of studies that have been evaluated the bond strength of ceramic brackets with different retention mechanisms and concluded that mechanically retained ceramic brackets have sufficent bond strength and seems to cause less enamel fracture of failure during debonding when compared to the chemically retained varieties[26, 31] . By the selection of adhesive material, different kinds of enamel conditioning and different durations in etching process, bond strength can also be modified. The mean bond strength of metal reinforced brackets is lower than conventional ceramic brackets and also comparable with stainless steel brackets. Omana et al. showed that mean shear bond strength of the polycrystalline ceramic brackets is significantly greater than that obtained when stainless steel brackets are used. Single crystal ceramic brackets produce the lowest mean shear bond strength values. Gwinnet reported that the mean values for the different bracket types are not statistically significant, but this conflicts with the results of many other studies. When compared to stainless steel brackets,the frictional properties of polycrystalline ceramic brackets are worst with any archwire combination whether bearing against stainless steel, nickel-titanium, cobalt-chromium or beta titanium archwires [24, 32]. The low fracture toughness (the ability of a material to resist fracture) of ceramic brackets causes to a higher incidence of bracket breakages or failure than with stainless steel brackets. Tie wings of the brackets can easily be broken of fracture because of the high torsional debonding forces in ceramic brackets. Base surface characteristics of ceramic brackets contains undercuts or grooves that supplies a mechanical interlock to the adhesive material. There are two types of ceramic bracket bases available. Firstly, ceramic brackets may have a flat base, covered with a silane layer with recesses for mechanical anchoring. Secondly, bracket base having a smooth surface rely on a chemical coating to enhance bond strength. A silane coupling agent is used as a chemical mediator between the adhesive material and the bracket base because of the inert composition of the aluminium oxide ceramic brackets. The manufacturers of such brackets have reported that they achieve higher bond strength when compared with mechanical retention. In our study, the brackets that are used have a base type that supplies a mechanical retention as well as a chemical coating was used on base to enhance the bond strength. Bonding process of orthodontic brackets has been used as a clinical method since 1970. In the bonding process,enamel surface changing or alteration that is created by acid etching is a crucial procedure. This procedure was developed by Buonocore in 1955. The steps that must be followed by clinicians are given below:

1. Cleaning,
2. Etching,
3. Sealing,
4. Bonding.

During specimen preparation, before bonding soft tissue debris and coronal pulps must be removed. The bonding surfaces of enamel must be polished with a non-fluoridated pumice paste to remove plaque and the organic pellicle that normally covers the teeth surface. Then, the teeth is conditioned with a 37% phosporic acid for 15 to 30 s, followed by thorough washing adn drying. A routine etching removes from 3 to 10 μm of surface enamel [17] . After that process, by using orthodontic composite adhesive material orthodontic brackets are bonded by one operator on the labial surfaces of incisors. After all etched enamel surfaces are coated, bracket placement should be started immediately (Figure 2.9) [17] . Excess adhesive must be removed before storing the prepared specimen in pure water at 37 °C for 48 hours in order to minimize the likelihood of bracket fracture.

Figure 2.9 Direct bracket bonding [17] .

2.4 Debonding Orthodontic Brackets

The purposes of the orthodontic bracket debonding are to remove the attachments and all the adhesive material from the teeth and restore the enamel surface as closely as possible to its pretreatment condition without any irreversible damages.

In order to accomplish these aims correctly, a correct method is of fundamental importance. Debonding of the orthodontic brackets may damage to the enamel and be time consuming if it is not achieved carelessly [20] . There are different techniques have been suggested to debond orthodontic ceramic brackets: Mechanical debonding by using special pliers (Figure 2.10) (Figure 2.11) , ultrasonic debonding , debonding by special kinds of burses, electro thermal debonding and laser aided debonding [2, 20, 33, 34, 35].

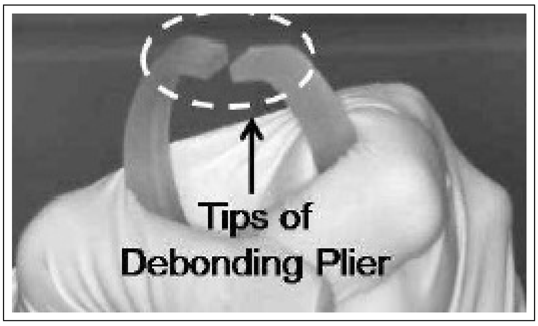

Figure 2.10 Tips of Debonding Pliers [9] .

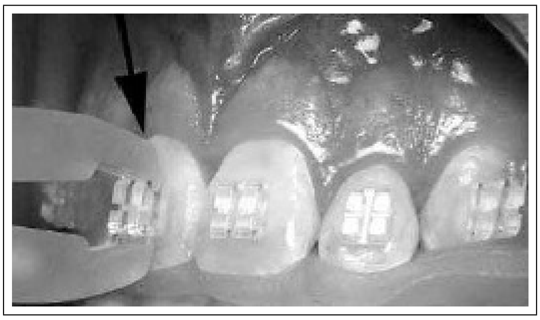

Figure 2.11 Conventional Debonding Method [9].

Effectiveness of various debonding methods change accordance to their advantages and disadvantages. The use of the lasers in debonding procedure can minimize risks and make debonding more efficent. The earliest type and currently one of the most popular mechanical debonding methods used for orthodontic brackets includes application of the blades of a debonding plier near the enamel surface but within the adhesive material. This method is quick and simple. However, it increases required force to debond orthodontic brackets and the risk of the enamal damage. In the study of Thomas and Prassana [36], effects of debonding metal and ceramic brackets from enamel surface by mechanical methods were compared. Four groups of brackets were used in this study. Enamel damage was seen significantly more in the groups with ceramic brackets than debonding metal brackets. In addition , it was concluded that ceramic brackets using mechanical retention appear to cause enamel damage less often those using chemical retention. In ultrasonic debonding method, specially designed tips are applied at the bracket-adhesive interface to erode the adhesive layer between the enamel surface and bracket base, the resulting force magnitudes required with the ultrasonic approach are significanlty lower than those needed for the conventional techinique of orthodontic bracket debonding. However, this method has an disadvantage, debonding time using in this technique is 30 to 60 seconds for each bracket compared with 1 to 5 seconds for other bracket removal method. Time consuming and excessive wear of ultrasonic tips can be accepted as drawbacks. In electrothermal debonding,instruments are rechargeable .The used instruments in this method transfer the heat through the bracket by softening the adhesive agent and permitting the bond failure between the bracket base and adhesive material . Required debonding force, risk of the enamel damage,pulpal damage,soft tissue burns and patient discomfort reduce when compared to other types of methods. On the other hand, water spray coolant needed to minimize the detrimental heating effect on pulp is one of the disadvantages of this technique. In other words , high temperature produced at the heated tip is the major disadvantages of this method [36] . All in all , usage of dental laser in orthodontic bracket debonding is one of the hot topics in orthodontics. Firstly , following chapter will give necessary information about laser and hard dental tissue interaction. Secondly, studies in laser-aided orthodontic bracket debonding up to now will be explained.

2.5 Laser and Dental Hard Tissue Interactions

Lasers are devices that produce highly directional, monochromatic, and intense beams of light. They are the most commonly used light source for biophotonics. It is an optical device that produces an intense monochromatic beam of coherent light. Since the first demonstration of laser action in 1960, lasers have enriched all aspects of life. Optic laser technology began with the invention of ruby lasers. During the 1980s and early 1990s, the use of lasers was introduced into dentistry and ophthalmology as various types were approved by the United States Food and Drug Administration. Since the early 1990s, lasers have been used experimentally for removing orthodontic ceramic brackets. Lasers can be distinguished into different categories using different considerations. In classification of the laser systems, wavelength, active material used,power of laser operation must be evaluated. Some of these classifications are given below in the table (Figure 2.12) .

Parameters	Wavelength	Mode of Operation	Active Medium Material	Power
	UV	Pulse Mode	Gas	High Power
	Visible	Continous Wave Mode (CW)	Liquid	Low Power
	IR		Solid	
			Electronic	

Figure 2.12 Classification of Laser Parameters .

A tissue is a self-supporting bulk medium. Therefore,biological tissues act like any bulk medium in which light propagation produces absorption, scattering, refraction, and reflection. These are fundamentals of light matter interactions,when medical lasers contact with the biological tissue,same photophysical process occur. These four possible processes are given in (Figure 2.13) .

In optics of biomedical, absorption of photons is the most important event. In biological tissues, water molecules or macromolecules (e.g proteins and pigments) are the agents that mainly cause absorption. Absorption depends on the electronic constitutions of atoms and molecules,the wavelength of radiation,the thickness of the absorbing layer,internal parameters (i.e temperature or concentration). In order to describe the effectiveness of absorption mode,absorption coefficent is used as a term. Absorption is the event that allows a laser or other light source to lead a potentially damaging effect on a tissue. In the absence of absorption,no energy transfer to the tissue

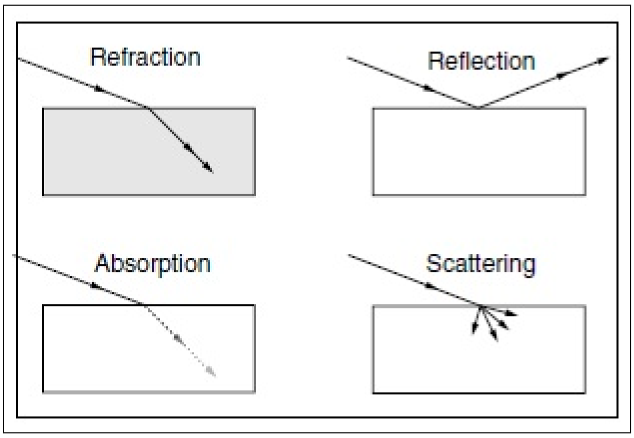

Figure 2.13 The four possible phenomias of interaction between light and biological tissue [9] .

occurs and the tissue is left unaffected by the light source or laser. If the incoming light is transmitted or reflected from the tissue,it will not lead to thermal effect on tissue. On the other hand,if the incident light is absorbed by tissue,converted heat may cause irreversible effects on tissue. In the ultraviolet (UV region), the absorption increases with shorter wavelength due to protein, DNA and other molecules. In a biological tissue,almost 75 % of it consists of water. In the infrared wavelengths,the absorption ascends by longer wavelenghts according to amount of water that tissue contains. Water content determines the absorption (Figure-2.14) [9] . According to the properties of wavelength,it was seen that the minimal value for absorption coefficient of hydroxapatite ($Ca_{10}(PO_4)_6(OH)_2$) is observed in visible and near infra-red region .

In the red to near-infrared (NIR), absorption is miminal. This region is called the diagnostic and therapeutic window. originally by John Parrish and Rox Anderson. The optical properties of biological tissues are determined by the optical properties of components,amounts and the distrubution of substances within the tissue. In the absorption spectrum of Hydroxyapatite ($Ca_{10}(PO_4)_6(OH)_2$) and enamel, absorption coefficent for every part of the tooth differs from each other. Thus, from the given figure below, it is clear that tooth structure composition is not homogenous, because amounts of both organic and inorganic substances shown in dentin are different from that present in enamel (Figure 2.15) .

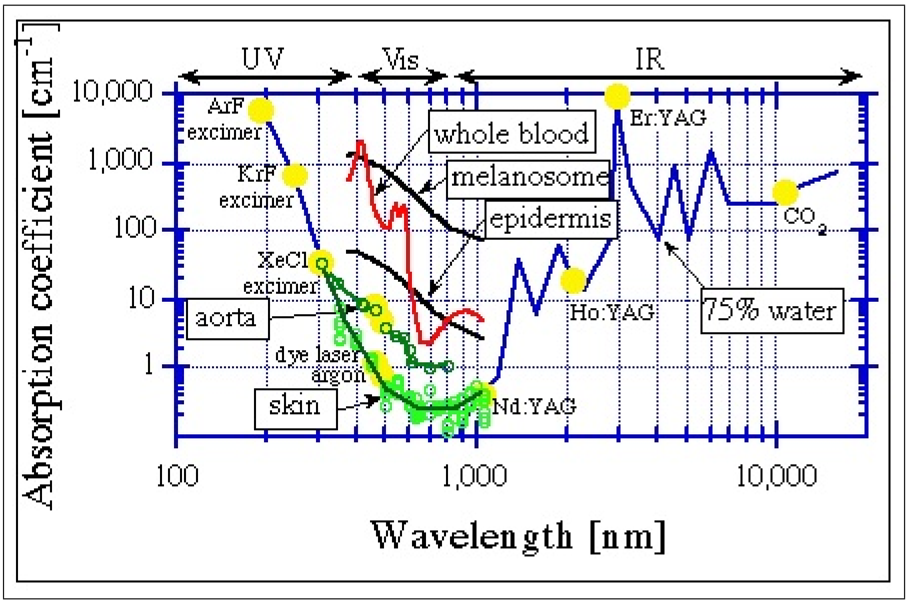

Figure 2.14 Absorption coefficent spectrum of biological tissues [9].

Since the invention of the ruby laser in the early 1960's,crucial innovations have been made in optical laser technology. By using these innovations , orthodontists have found various uses for dental lasers. The use of lasers in orthodontics eliminates lots of problems during orthodontic bracket debonding . Lasers that are used for orthodontic bracket removing can reduce the needed debonding force ,risk of the enamel damage , incidence bracket fracture anda re potentially less traumatic and painfull [37] . As given in the previous sentence, during orthodontic bracket debonding , one can face the enamel cracks (Figure-2.16) [9] . Laser-aided debonding can overwhelm and minimize the occurance probability of enamel cracks after debonding process .

On the other hand, besides plenty of advantages, dental lasers have also some drawbacks : potential pulpal damage due to heat production, expensive units and laser hazards. Thermal effect during laser irradiation on dental tissues can cause irreversible conclusions. There are lots of studies about thermal effects of laser usage during debonding. Zach and Cohen worked on monkey teeth in their study. According to the study ,increase in intrapulpal temperature was accepted as 5.5 °C as a safety threshold value in order to prevent undesirable results after orthodontic treatment. Under that threshold temperature ,no histological changes were discernible with an intrapulpal temperature increase of 1.8 ± °C . Also,it had been concluded that an increase in

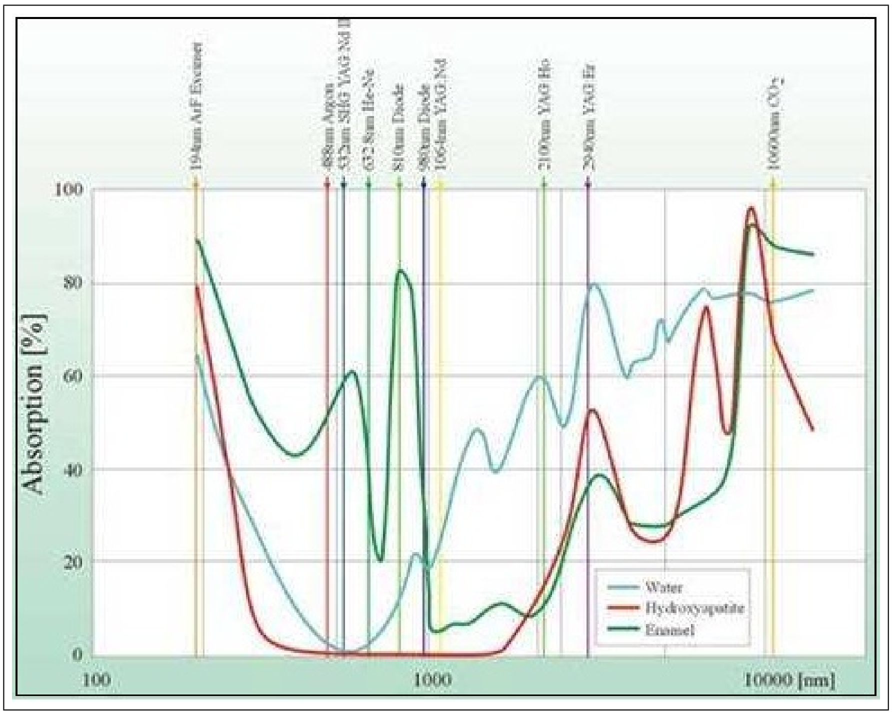

Figure 2.15 Absorption Spectrum of Hydroxyapatite and Enamel [9].

intrapulpal temperature of 11.1 ± °C , 60 % teeth showed abscess formation. At an 16.6 °C elevation pulpal necrosis occurred in all of the teeth [38] .

Strobl et al. debonded ceramic brackets by using both CO_2 and Nd:YAG lasers. Their results showed that laser-aided debonding significantly reduced debonding force by thermal softening of the adhesive material. It was also revealed that with the Nd:YAG laser, approximately 69-75 % of the incoming laser light reached the enamel surface, which has the potential to cause pain or damage to the tooth structure. In their studies where they used a CO_2 laser and modified debonding pliers, Rickabaugh et al. [39] and Ma et al. [40] showed that there is a linear relationship between lasing time and intrapulpal temperature change.In this study,it was stated that the ceramic bracket could be removed from the tooth with the aid of the pliers as soon as the adhesive softening temperature is reached. This quick removal prevented the heat energy stored within the bracket from transmitting onto the tooth. H. Jelinkova, and T. Dostalova [39] used three continuously running lasers, i.e. the diode-pumped Tm:YAP (wavelength 1997 nm), Nd:YAG (wavelength 1444nm), and GaAs diode (wavelength 808 nm) with the ceramic bracket removal after irradiation by laser radiation was com-

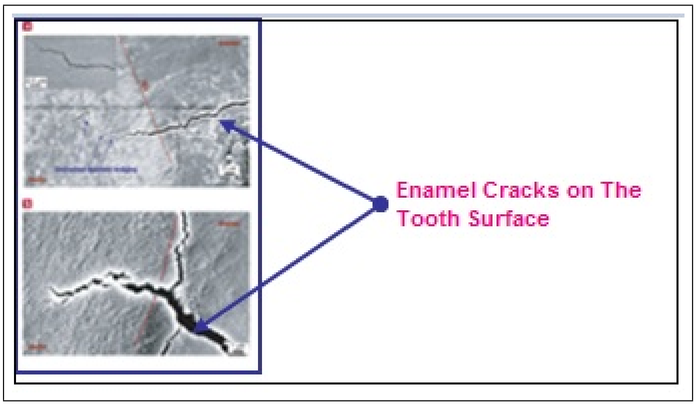

Figure 2.16 Enamel Cracks.

pared. From the results it follows that continuously running diode-pumped Tm:YAG or Nd:YAG laser generating wavelengths 1997 nm or 1444 nm , respectively, having the output power 1 W can be good candidates for ceramic brackets debonding. In this study,it has been found that the near infrared radiation from the GaAlAs laser is transmitted through the bracket and bonding agent, and heat generated by this radiation is concentrated into the tooth resulting in unacceptable increase tooth temperature but in no effect for debonding. (upon 14 °C for 60 s irradiation) By choosing the appropriate laser parameters, one can easily minimize and overwhelm the risks of the laser irradiation that are given above during laser debonding and laser use can be more efficent way to debond ceramic brackets without side effects of the procedure .

2.6 Lasers in Debonding of Orthodontic Brackets

By the investigation of the first ruby laser, application of the lasers in dentistry has began. However, after many years, the laser was really used for dental treatments. For soft tissue surgery in the oral cavity, continuously running Nd:YAG laser [21], for tooth drilling- pulsed Er:YAG and Ti:Sapphire [11, 17, 38, 40, 37, 5] for endodonty - Er:YAG and alexandrite lasers [41, 42] or diode laser [42]-[43] , for bleaching - alexandrite or diode laser [39, 44, 45] were used by orthodontists. Also,in orthodontic treatments,a new method for tooth alignment by using brackets had been introduced to the field,the effect on the tooth of the laser irradiation was investigated. The usefulness of the lasers for tooth alignment is in laser radiation assistance in debonding of brackets [17] . Up to now,various types of debonding methods have been investigated. These methods contain debonding pliers for mechanical debonding, hand sealers and ultrasonic and electrothermal debonding [46, 26]. In order to debond brackets, radiations of gas laser- CO_2 , excimer KrF or XeCl laser, a solid state Nd:YAG laser, diode laser or halogen lamp and LED system were used by clinicans [27, 28, 47, 29, 16, 48] . During the experiments and innovations,it was revealed that not all laser radiations can be efficent for bracket debonding from the enamel.Thus, new possibilities and studies for bracket removing are still being investigated. Laser irradiation is needed to soften the adhesive resin in ceramic bracket debonding.Laser irradiation should be positively efficent on the adhesive material-composite resin - melt it , and make lower the cohesive enamel strength. From the physical and chemical point of view, the bond between the composite resin and the enamel could be broken by laser irradiation. This break can be realised through absorption of the laser radiation and according to Tocchio et al.,there are three ways to degrade the adhesive material by using the laser energy: Thermal softening,thermal ablation and photoablation [21] . The type and the efficency of interaction depends on the wavelength and power density of the laser radiation used. In thermal softening, laser energy is used until adhesive material softens. When adhesive material degrades, the bracket cannot overcome the gravity. So,it drops from the enamel surface. In thermal ablation,the fast irradiation is the important point because when laser irradiation is done fast enough,increasing in the temperature of the adhesive resin occurs fastly and this results in evoporation of adhesive material before debonding by thermal softening occurs. Photoablation occurs when the high energy laser affects the energy level bonds between the adhesive resin the result is decomposition of the material. Thermal ablation and photoablation are more rapid process, so very little thermal effects can be observed during debonding. Thermal softening is the most accepted interaction during bracket debonding. However, it can cause a large rise in temperature , because it is slower process when compared to the photoablation and thermal ablation. According to the study of Tocchio et al. [49] in 1993, by using 248, 308 and 1060 nm of radiations, monocrystalline and polycrystalline ceramic brackets were irradiated. It was concluded that debonding of monocrystalline brackets at wavelengths of 248 and 308 nm occurred by ablation with power densities ranging from as low as 2.7 W/cm^2. They were not able to distinguish between thermal ablation and photoablation at all power densities when using the 248-nm wavelengths and power densities above 17 W/cm^2 for the 308-nm wavelength. Using the 308-nm wavelength at powers at or below 9 W/cm^2 caused debonding to occur by thermal ablation. Debonding of monocrystalline brackets with a 1060-nm wavelength occurred by either photoablation or thermal ablation at power densities greater than 26 W/cm^2 and also thermal softening at lower power. Debonding of polycrystalline brackets at wavelengths of 248, 308, and 1060 nm using laser power density of 32.6 W/cm^2 resulted from thermal softening. Strobl et al.,[3] revealed that the laser debonding mechanism of was thermal softening of the adhesive resin. However, Mimura et al. and Obata suggested that both thermal softening and resin contraction from ceramic brackets are responsible for the debonding mechanism when using Super-bond (Sunmedical,Kyoto, Japan) (MMA containing 4-META). MMA resin contraction from the ceramic bracket is caused by the differences in the thermal expansion. There are lots of studies done by using various types of dental lasers in orthodontic bracket debonding up to now (Figure-2.17). Studies can be classified due to time span,effects on the pulp, time lag between lasing and removing brackets,different bonding adhesives used.

The removal of ceramic brackets from the enamel surface by means of laser heating was investigated with the use of CO_2 and YAG lasers. In 1992, Strobl et al. used CO_2 and Nd:YAG lasers for both types of ceramic brackets that were monocrystalline and polycrystalline ceramic brackets in order to remove them from the enamel surface [49] . In this study, it was aimed that with the help of the dental lasers, force to debond the ceramic brackets was significantly reduced. Motorized translation stage to break the bonding between ceramic bracket and enamel with a speed of 1 mm/sec.In both laser groups,debonding was started by application of a torque force after 2 seconds of laser irradiation. In the preliminary studies, 14.1 W laser energy power for 2 seconds application was selected for removing ceramic brackets. By application of CO_2 laser with a power of 14 W and a laser duration of 2 seconds, the required average debonding force to remove polycrystalline ceramic brackets was decreased and bracket failures were not observed in the results. During debonding process of monocrystalline ceramic brackets by application of laser at a power of 7 W average debonding force was

Year	Researcher	Laser
1992	Strobl *et al.*	CO2 (10600 nm) / Nd :YAG (1964 nm)
1993	Tocchio *et al.*	KrF (248 nm) / XeCl (308 nm) / Nd :YAG (1964 nm)
1995	Obata *et al.*	CO2 (10600 nm)
1995	Mimura *et al.*	CO2 (10600 nm)
1996	Rickabaugh *et al.*	CO2 (10600 nm)
1997	Ma *et al.*	CO2 (10600 nm)
1999	Abdul-Kader and İbrahim *et al.*	Nd : YAG (1064 nm) / CO2 (10600 nm)
2005	Hayakawa *et al.*	Nd : YAG (1064 nm)
2008	Xianglong *et al.*	Nd : YAG (1064 nm)
2008	Dostalovaa *et al.*	Tm : YAP (1980 nm) / GaAlAs (808 nm)
2008	Dostalovaa *et al.*	Tm : YAP (1980 nm) / GaAlAs (808 nm) / Nd : YAG (1064 nm)
2010	Ayse Sena Kabas Sarp *et al.*	Ytterbium Fiber Laser (1070 nm)
2010	M.Oğuz Öztoprak *et al.*	Er:YAG (2940 nm)
2011	F. Ahrari *et al.*	CO2 (10600 nm)

Figure 2.17 Studies done up to now on laser debonding orthodontic ceramic brackets.

decreased when compared with control group that was non-laser group but monocrystalline brackets required lower laser energy for debonding than polycrystalline ones in this study. The reasons of different conclusions were explained as a result of differences in the layout of two brackets. After all experiments,it was concluded that the average torque force ro debond polycrystalline brackets was decreased 25 % with a CO_2 laser irradiation at 14 W for 2 seconds. Finally, that resulted in removing of all ceramic brackets without failure. In 1993, Tocchio et al. [49] debonded polycrystalline and monocrystalline ceramic brackets that had been bonded with Dynabond (3M Unitek) by using 248 nm ,308 nm and 1060 nm of laser radiation. In this study, either 0 or 0.8 Mpa externally applied stress was used. The ceramic brackets were removed by irradiation of their labial surfaces at power densities between 3 and 33 W/cm^2 and also with the laser radiation of 248, 308 and 1060 nm. The laser beam was applied perpendicularly to the bonding interface. Suitable laser power was 32.6 W/cm^2 at the bracket surface in order to remove all polycrystalline ceramic brackets and also 60% of monocrystalline ceramic brackets. In this study,debondng times of both groups were also evaluated. The debonding times for polycrystalline ceramic brackets were different. The average debonding time for them was 3.1 seconds during 248 nm laser irradiation. During 308 nm laser irradiation, it was 4.8 seconds. Debonding time was 2.37 seconds when samples were irradiated by 1060 nm wavelength. According to the experiments, it was concluded that longer debonding times gained when debonding polycrystalline brackets with 1060 nm. In any samples that were experimentally used in this study showed no bracket failure. For monocrystalline ceramic brackets,debonding types were less than 1 seconds at three different wavelenghts. When 248 nm of laser irradiation was applied, no changes in debonding time were observed. In 308 nm laser irradiation , increase in time observed at power of 9W/cm^2 or less and also for 1060 nm of laser irradiation increases in time seen while the power was decreasing. Intrapulpal temperature changes were not considered in this study. SEM (scanning electron microscopy) observations released that no enamel damage was observed in samples.

Obata et al., [50] studied the effects of CO_2 laser on ceramic bracket debonding by using two different bonding agents in 1995. The selected bonding agents were Bis-GMA composite resin and 4-META MMA resin.In this study,by using the gained information from preliminary experiments,for the 4-META MMA adhesive resin group,suitable laser power was 3 W and laser duration was 3 seconds. Obata suggested that both thermal and adhesive resin contraction from ceramic brackets are responsible for debonding mechanism like Mimura[35] , because of the differences in the thermal expansion between MMA resin and the ceramic bracket, at 60 $\pm$ °C the resin changed from expansion to contraction. In the control group which was debonded by conventional method,after measuring shear forces,enamel and bracket breakages were observed.Intrapulpal temperature changes during laser irradiation were observed. No histological differences were reported between laser irradiated and control group teeth.In both laser groups that were irradiated by CO_2 laser had no bracket fractures in the experiment. On the other hand,in the control group,enamel cracks and bracket failures were observed. Results of this study showed that the CO_2 laser debonding method is safe for tooth pulp, and the use of 4-META MMA resin is safer than that of Bis-GMA resin because lower laser irradiation was necessary to cause debonding.

Rickabaugh et al. [50] studied at varying tensile forces in order to advance in vitro usage of the laser debonding pliers with a CO_2 laser in the debonding of ceramic brackets in 1996. In this study, CO_2 laser were experimentally used to remove polycrystalline ceramic brackets that were bonded by using MMA type of adhesive material.Intrapulpal temperature changes during laser irradiation were evaluated and compared.5.5 $\pm$ °C was accepted as a threshold level for safety. Laser application were achieved perpendicularly to the bracket surface.For debonding of the ceramic brackets,a set of instron machine with a speed of 1 inch a minute was preferred.In this study,a control group with debonding and also with no lasing were added. The average debonding force for this group was 4.88 Mpa (13.04 pounds). During debonding of the specimens,bracet breakage were observed in 3 of the 10 teeth. During lasing by a modified debonding plier, a tensile debonding force was applied.These debonding forces by a modified debonding plier were chosen according to the debonding force applied to the control group. The three of the groups were irradiated and debonded by the CO_2 laser at 20 W and a static tensile force of 3 pounds,1,5 pounds and 0,75 pounds. In group with 3 pounds of tensile force, the average debonding time for the group was 1.64 seconds and the average increase in intrapulpal temperature was 1.80 °C. In this group, only one specimen exceed the threshold value that was determined as safety temperature 5.5$\pm$ °C . In the group with 1.5 pounds tensile force,the average debonding time was 1.83 seconds and the average increase in intrapulpal temperature change was 3.01 $\pm$ °C . Only one specimen indicated a temperature change that exceeded the threshold value in intrapulpal temperature change. Also,when bracket breakage or wing tie was evaluated,no broken tiw wings were reported. Group with a static tensile force of 0.75 pounds, had an average debonding time of 3.42 seconds and an average of 4.47$\pm$ °C temperature. 3 specimens in that group had temperature changes more than 5.5$\pm$ °C that exceeded the threshold or safety value. No broken tie wings were reported for that group. 3.0 and 1.5 pound tensile force applied groups needed significantly less time than 0.75 pound tensile group. For 1.5 pound tensile force group,a good relationship was observed between time and temperature for the 1.5 pound group and 0.75 pound groups.In conclusion,1.5 pound tensile force group had the best results in terms of debonding time and increase in intrapulpal temperature change. Finally, this study concluded that it agreed with those of Strobl et al. and Tocchio et al. in the laser influences thermally soften the adhesive material to debond the ceramic bracket readily. In 1997, Tsun Ma et al. [40] studied a method to overwhelm or reduce the breakage of ceramic brackets during debonding that is accepted as an disadvantage. The purpose of their study was to determine the amount of laser duration needed to achieve an exact decreasing of debonding while keeping the intrapulpal temperature changes below the threshold value. In this study,mandibular bovine teeth and human mandibular first premolars were used. Ceramic brackets were bonded to those specimens by a photoactivated bonding agent. In the control group brackets were debonded in the tensile mode with a crosshead speed of 0,2 inch (0,51 cm) per minute, with debonding pliers attached to the instron machine and no laser the application. For every specimen, the tensile force that was needed to debond the ceramic brackets was measured during experiment. Then,the average and standard deviations of the control group with no laser application were 15.31 pounds that was equal 68.1 N. For the experimental group , a tensile debonding load of 4 pounds which was 25 % of control group was selected. With CO_2 laser in CW power at 18 W power level was selected for this study. In order to position the laser light onto the ceramic bracket , modified debonding pliers were used . The laser was activated for the specific period of time. By using thermocouple , intrapulpal temperature changes were recorded . Lasing time required to keep the maximum intrapulpal temperature rise below $2 \pm$ °C was chosen as a factor of safety. After examination of the debonded samples , it was concluded that adhesive faiure at the resin-bracket interface was observed. According to the results of this study , there was no incidence of bond failure at the resin/enamel interface. In the control group , tie- wing fractures were observed and three specimen had mounting block fractures. 20 % of the specimens did not debond with the 4-pound debonding force , which is equal to a tensile stress of 15.19 Kg/cm^2, or 1.48 Mpa. The remaining 80 Abdul Kader and Ibrahim [51] removed ceramic brackets by using carbondioxide laser at a power of 50 W in 1999. A time interval of 2 seconds were determined as an exposure time. On the bracket surface , the temperature change was saved $93.63 \pm$ °C .On the other hand , increase in the pulp chamber did not exceed $0.7 \pm$ °C . According to their other study , it was stated that significantly higher debonding force was needed when one minute elapsed after laser exposure compared with debonding immediately after laser exposure , whatever the exposure time of laser was [53] . According to the study of Kotaro Hayakawa [7] , the application of high-peak power Nd :YAG laser at 2.0 J or more is effective for removing of ceramic brackets. In this study, two types of ceramic brackets were used : single crystal and polycrystalline . They were attached to bovine mandibular teeth with two different kinds of bonding agent (4-META / MMA and Bis-GMA). On each ceramic bracket , laser irradiation was done on to 2 points with a 1 pulse-per-second shot . At three different laser energy levels 1.0 , 2.0 ,3.0 Joule shear bond strength and thermal effects of the laser irradiation were evaluated . Also , a control group that had no laser application was observed in this study. All in all, the shear test ($p < 0.05$) indicated that every sample in the 2.0 and 3.0 J laser groups underwent a significant decrease in debonding force when compared to the control group . On the other hand , the 1.0 J lasing group , did not show any significant difference. Between different bonding adhesive materials , no significant differences was observed for all groups in the study . Thermal effects were also easily observed on the resin, but they were rather shallow; carbonization-like effects confirmed only on the resin surface. At 2.0 and 3.0 J laser energy levels , the bracket bases showed hollows and black deposits, and the remnant adhesive material had severe carbonization-like effects. The carbonization-like changes to the resin seemed to be deeper than those of the 4-META MMA bonding group .5.1 °C was recorded as the maximum temperature rise on the pulpal walls at the irradiation points by laser .

In 2008 , Dostalova et al. [52] used three laser systems for the bracket removing procedure : the compact diode-pumped Tm :YAP (1997 nm) laser (operating at 1980 nm with a maximum output power of 3.8 W) , a diode- pumped Nd :YAG laser (1444 nm) (operating at 1444-nm with a maximum output power of 2 W) and GaAs laser diode (808 nm) (operating at 808 nm with a maximum output power of 20 W). The output radiation from the particular laser was positioned into the tooth specimens with the orthodontic brackets. Bracket debonding was achieved mechanically after the selecting the proper interval. The heat transmission and absorption observations for bracket, adhesive resin and enamel were done by thermocouple measurements inside the tooth and thermal camera images to explain the thermal energy delivered during debonding.The irradiation time was gradually changed from 30 s up to 90 s in the case of Tm:YAP (1997 nm) and Nd:YAG (1444 nm) lasers. For GaAs diode laser (808 nm), the time interaction was 60 s. There were two configurations that were investigated in this study : irradiation without and with cooling of the sample. After irradiation of brackets by GaAs diode laser (808 nm) , they could not be removed even after 10 W used up to 60 s. Heat was increased up to 14$\pm$ °C in this lasing group. The laser power was 1 ,2 and 10 W for this laser.Also,the laser duration was 60 s. After irradiation by Tm:YAP laser (1997 nm) , inside the bracket the heat was concentrated. After 60 s laser duration, debonding of the polycrystaline ceramic brackets were achieved. Debonding of the brackets was almost same with the increasing exposure time (up to 90 s) or irradiation power of laser (up to 2 W). In the temperature increase in pulpal walls , the difference was observed. When cooling of the tooth was experimented ,the temperature increase was below the threshold value that was accepted as 5.5 °C for the power and the laser duration used 1 W and 60 s. Without cooling of the specimens , the temperature rise exceeds the tolerable increase . For Nd : YAG (1444 nm) laser irradiation , similar results were obtained . Only temperature rise during laser radiation in without cooling configuration was steeper compared to cooling one. Irradiation by Nd:YAG laser caused 17 °C temperature increase without cooling method. For cooling method , it was only 7 °C . Debonding was performed effectively both of these groups by Nd:YAG laser. All in all, after evaluation of SEM measurements , it was revealed that the minimum damage of the enamel for the case of Tm:YAP laser (1997 nm) at 60 s laser duration at the power of 1 W . Both Nd:YAG (1444 nm) and Tm:YAP (1997 nm) lasers with the power 1 W acting 60 s lasing interaction are efficent for bracket removing.

In 2010, Ayse Sena Kabas Sarp et al. [53], studied the effects of Ytterbium Fiber Laser for removing of the ceramic brackets from the enamel surface. In this study, intact 2-year-old bovine mandibular incisors were attached to polycrystalline ceramic brackets (G&H, US) for maxillary lateral incisors by chemically curing Bis-GMA adhesive agent (3M,Unite Bonding Adhesive Set , US). Debonding process was quantified with a universal testing machine , and intrapulpal temperature changes during laser irradiation. Debonding force was recorded by universal testing machinge

.Intrapulpal temperature increase was saved by thermocouple during irradiation in order to prevent irreversible effects of irradiation due to tempereature rise. According to the types of lasing mode ,experiments were done in two parts : continuous wave (CW) and modulated mode. Ytterbium Fiber Laser (1070 nm) was applied on specimens with different stable laser power level in continous mode (CW) . In the second part of the experiments , ceramic brackets were irradiated by modulated mode , in which the laser energy was delivered with on-and-off cycles. The power of laser and duty cycles were set by controlling the current , that was adjusted to 4.99 A of current for 18 W of emission. Output power of Ytterbium Laser was stable during the on cycle (i.e., 18 W) and there was no laser irradiation during the off cycle. Results of the lasin groups were compared to the results of the control group in this study. In debonding force observations , CW laser irradiations indicated a decrease in torque force for almost all parameters but only for both groups of 1.41 ± 0.07 W and 4.20 ± 0.07 W measured power showed significant differences ($p<0.01$ and $p<0.05$, respectively) compared to the control group. Due to debonding time , though CW laser irradiations with appropriate powers resulted in a reduction in debonding force required, they did not lead to any significant decrease in debonding time . The intrapulpal temperature rise for all studied groups except 3.50 ± 0.07 W measured power and 4.20 ± 0.07 of CW mode were below the threshold value 5.5 °C. For CW mode lasin groups, with increasing laser powers, intrapulpal temperature changes increased. In modulated mode lasing groups, no significant difference in intrapulpal temperature changes between the irradiated groups was observed. For all groups, laser irradiation reduced the required load applied for removing of polycrystalline ceramic brackets ;consequently, work done by the testing machine was changed due to laser irradiation in this study. When enamel and bracket's base surfaces were evaluated , residual adhesive was totally observed on the enamel (79.27 %) in most of the samples for all groups,. There were small. No mechanical fracture of enamel was observed . This study also revealed that the mode of operation is as important as the wavelength and the output power of the laser used.

In 2010 , M. Oğuz Öztoprak et al. [54] studied to develop a new method for ceramic bracket debonding by a scanning method with an Er:YAG laser. In this study , sixty bovine mandibular teeth were distinguished into two groups of 30. On their labial surfaces , polycrystalline ceramic brackets were attached by using the orthodontic composite adhesive material Transbond XT (3M Unitek , Monrovia , Calif)and light cured for 40 s . One of the groups was nonlasing group . The selected laser Er:YAG was applied to the samples at a power of 4.2 W with a wavelentgh of 2940 nm. Irradiation by laser was performed by scanning method throughly the surface of the brackets for 9 seconds . Scanning was applied with horizontal movements parallel to the bracket slot starting from the upper distal wing of the bracket to the upper mesial wing and then to the slot of the bracket and the lowr wings. After 45 seconds laser pulse had been applied on to the samples , the shear test was performed. The shear test was measured by using universal testing machine (Instron , Canton , Mass) in megapascals . After evaluations of the results , they showed significant differences between nonlasing and lasing groups ($p < 0.001$) . Lower debonding forces were observed in the laser group (9.52 Mpa) . In nonlasing group , debonding force was 20.75 Mpa. Also ARI (adhesive remnant index) scores were examined in this study. According to the results of this study , ARI score were significantly different . The laser group had twice as many specimens with adhesive agent,with ARI scores of 2 or 3. A negative correlation was found between bond strengths and ARI scores ($p<0.001$) . When the shear bond strength decreased , the ARI scores increased .

F. Ahrari et al. [55] studied to observe the surface properties of enamel after ceramic bracket removing with or without laser beam. In this study , a total of 90 upper and lower premolar teeth were used. Eighty teeth were used for enamel damage assesment and ten for the temperature measurement. Eighty teeth were attachted with either of the chemically retained or the mechanically retained ceramic brackets and after that procedure removed conventionally or by a carbondioxide laser (188 W , 400 Hz). Irradiation by carbondioxide laser was achieved by scanning method for 5 s. After removing of the polycrystalline ceramic brackets , ARI (adhesive remnant index) scores , the probability of bracket and enamel cracks , and lengths ,frequency , and the directions of enamel cracks were compared among all groups. In chemical retention-conventional removing group ,there was only one sample of enamel fracture.In conventional ceramic bracket removing by debonding pliers , incidences of bracket fracture were 45

3. MATERIALS AND METHODS

3.1 Teeth

In this study,freshly extracted deciduous bovine mandibular incisors are used. These teeth are selected because of their availability , higher hygiene and they are almost similar to human teeth phsiologically. Also,when we use these teeth, no risk of infectious diseas transmission is observed . The use of bovine teeth for bonding material to enamel bonding studies has been validated [40, 33, 51, 56] .

3.2 Orthodontic Ceramic Brackets

Ceramic brackets are distinguished into two groups: Single crystal (monocrystalline) and polycrystal (polycrystalline) [1, 3] . Throughout the experiment, one type of ceramic brackets is used. They are polycrystalline ceramic orthodontic brackets (G&H.US) (Figure 3.1) .Polycrystalline brackets are selected because of their availability and common use [1] . Polycrystalline ceramic brackets are made for maxillar and mandibular incisors. When the previous studies are evaluated, it is obvious that single crystal structure has greater strenght. Polycrystalline ceramic orthodontic brackets need higher debonding force when compared to monocrystalline (single crystal) ceramic brackets. In this study, polycrystalline ceramic brackets are bonded to mandibular bovine incisors. The composite resin that is used to bond the polycrystalline ceramic brackets to tooth surface is the photoactivated type Bis-GMA adhesive resin set (3M, Unite Bonding Adhesive Set, US) because of its high tensile bonding strength, when compared with other ceramic brackets on which this adhesive is used [57].

Figure 3.1 Polycrystalline Ceramic Brackets.

3.3 Specimen Preparation and Bonding Procedure

In the specimen preparation, first of all soft tissue debris,calculus and blood or pulpal tissues around the tooth are cleaned . The surface of the enamel of the bovine incisor teeth are cleaned with a nonfluoridated pumice paste and washed with tap water. Teeth are stored in thymol solution at room temperature until required. In order to avoid the side effects of bacterias, solution has been changed three times per week. By using barbed broach endodontic files cavities are opened on the lingual surface of the tooth to ease the thermocouple placement inside the pulp chamber and identify temperature changes during laser irradiation. Cavity opening are achieved by a round diamond bur with a 1 mm diameter. Then ,every mandibular bovine tooth is embedded in gypsum blocks (Figure 3.2) .

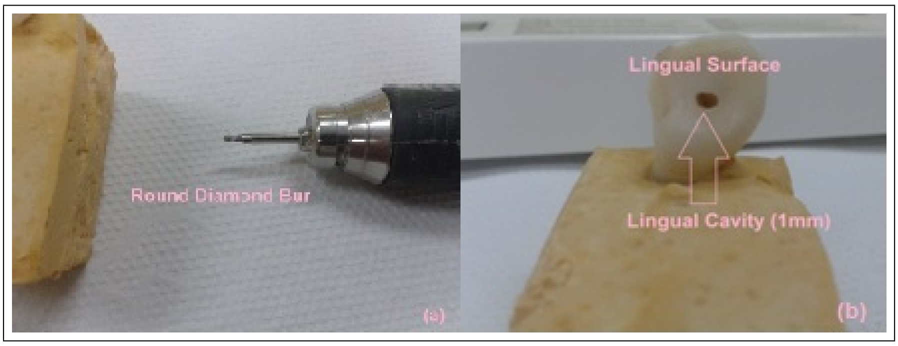

Figure 3.2 (a) Round diamond bur and (b) Lingual cavity were with a 1mm length diameter.

The bovine teeth are placed with their labial surface. They are as parallel as possible to the vertical axis of the block so that debonding would be in a rigid and pure tensile mode (Figure 3.3) [9].

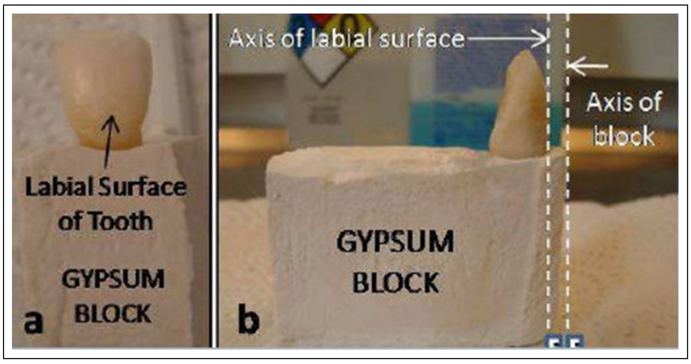

Figure 3.3 (a) Labial view and (b) Lateral view of bovine crown which was embedded in gypsum block .

All orthodontic ceramic brackets are bonded by adhesive bonding resin following the recommendations of manufacturer. Cleaning, etching, sealing and bonding processes are the major steps of the bonding procedure. Before bonding of the ceramic brackets, the front side of the all specimen teeth are polished for 15 seconds with a nonfluoridated pumice paste on labial enamel surface and cleaned, washed by tap water and dried for plaque removing (Figure 3.4) [9].

Figure 3.4 Polishing Process by Pumice Paste.

The teeth were then etched with a 37 % orthoposphoric acid gel for 15 seconds and rinsed by a water syringe and dried. A calciferous and frosty white matt appearance is gained on the tooth surface. Then,bonding agent or sealant is applied over the gained frosty white region on the tooth and also bracket base. After etched surface coated, the polcrystalline ceramic bracket are positioned immediately on the labial surface of the tooth. The orthodontic ceramic bracket is kept stable by tweezers and then adhesive material is applied to the back and smooth base of the bracket that contacts with the tooh. The brackets is placed mesio-distally and inciso-gingivally accurately relative to the long axis of the teeth. The bonding interface of enamel and bracket base is axially centered and placed parallel to the face of the gypsum block. The ceramic orthodontic bracket is positioned at the opposite side of the opened cavity on the labial surface of the tooth (Figure 3.5)

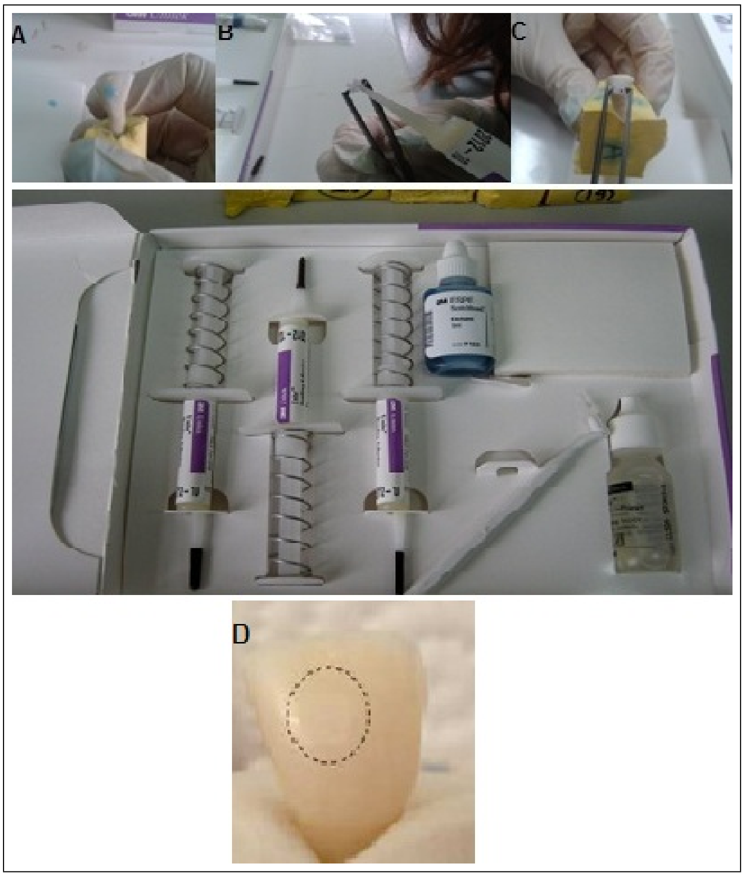

Figure 3.5 (A) (B) (C) Application of self-etching primer (3M,Unite Bonding Adhesive Set,US) on enamel surface of maxillary incisor and bonding process of the ceramic bracket. (D) Calciferous and frosty white matt appearance after etching process

Then, the bracket is pushed tightly towards ot the tooth in one-point contact. Excess adhesive material is cleaned from the edge of the bracket with a dental explorer. In our experiment ,it is important to clean excessive adhesive material around the bracket because during mechanical debonding excessive adhesive material can cause enamel tear outs. In clinical application,a clinician must consider about this step to prevent inflammation that comes close to gingiva [38] . Each tooth in gypsum block with a bonded ceramic bracket was stored in an incubator with sodium chloride solution inside and 100 % humidity and 37 °C for 48 hours before testing to ensure the composite polymerization (Figure 3.6) .

Figure 3.6 Samples are kept in the incubator for the composite polymerization.

3.4 Experimental Set-up

Major instruments required for the study:

1. 1940-nm Thulium Fiber Laser System,
2. Universal Testing Machine,
3. Temperature Measurement System(K-type thermocouple).

A 1940 nm Thulium Fiber Laser is used in this study. In order to control the suitable and chosen laser parameters from the prelimininary studies for the used laser irradiation,at the beginning of the study, the output energy of the 1940 nm Thulium Fiber Laser are measured by a powermeter (Newport, Model 1918-C). The powermeter contains a head to measure the laser power and a monitor to control and read the given data.Also, it contains a metal holder to fix the fiber laser tip. After measuring the laser power, the bonded human teeth are replaced into the testing frame in universal testing machine (Lloyd, LF Plus, UK) (Figure 3.7) . A universal testing machine (Lloyd, LF

Figure 3.7 (A) 1940- nm Thulium Fiber Laser , (B) Universal testing machine (Lloyd, LF Plus, UK) , (C) Computer for collecting and recording data from the thermocouple , (D) Computer that is connected to universal testing machine for controlling and collecting and recording data , (E) Monitor and controller of Powermeter (Newport, Model 1918-C) , (F) Sensitive measuring head of powermeter, (G) K-type thermocouple (OMEGA, OM-CP-0CTTEMP, UK), (H) Shearing Blade.

Plus, UK) were used to measure the SBS (shear bond strength) value of each specimen during debonding. A steel shearing blade is mounted to the upper part of the machine. Moreover, a computer system connected to the testing machine were used to control the machine and record the data.

Intrapulpal temperature changes are recorded by K-type thermocouple (OMEGA, OM-CP-0CTTEMP,UK) during laser irradiation . A computer system is connected to K-Type thermocouple system that reads, collects and records the data obtained from the experiment is included in this study.

Figure 3.8 Universal Testing Machine with a base and a moving part, testing frame (Lloyd, LF Plus, UK).

The thickness of each tooth is measured by using a stopper and mean of the collected values is equal to 2.56 mm. For the study, we choose the teeth which have almost similar thickness according to this measurement for much more reliable results.

3.5 Experimental Procedure

In the first step of the experiment, the power of the 1940 nm Thulium Fiber Laser is measured by a powermeter (Newport, Model 1918-C). The external diameter of the fiber waveguide is 2 mm. It is controlled that the diameter of the laser light beam is always in the same length while distance is changing. In this experiment, experiment is done in one way of the lasing mode: Laser irradiation is applied to the specimens in Continuous Wave (CW) Mode. In the study, experiments are done in continues wave (CW) mode, laser irradiation on samples are done by organized fixed laser power. The proper output energy levels of 1940-nm Thulium Fiber Laser are determined after the preliminary studies due to the intrapulpal temperature changes during the irradiation by this laser. According to the evaluation of the preliminary experiments, real experiments are distinguished into 9 different groups (Figure 3.9) .

Adjusted Laser Power	Laser Duration	Irradiation Method	Number of Samples
3 W	7 / 10 seconds	Scanning / No Scanning	11
2,5 W	7 / 10 seconds	Scanning / No Scanning	11
Control Group	No laser irradiation	No applied method	11

Figure 3.9 Applied and measured power of 1940 nm Thulium Fiber Laser and number of samples for each group in CW mode.

In the laser groups, the 1940 nm Thulium Fiber Laser is used on each ceramic bracket in the experiment at 2.5 and 3.0W for 7 and 10 seconds with the scanning and non- scanning methods. By using this laser, our aim is to eliminate and decrease the damage on the enamel. Moreover,keeping the intrapulpal temperature change below the threshold value is another important purpose. 5.5 °C is accepted as the threshold value for safety in intrapulpal temperature changes. The laser beam is applied perpendicularly to the surface of the ceramic bracket to avoid unnecessary beam reflection and the tip of the laser must be at a fixed distance from the ceramic bracket to keep constant the amount of energy delivered [12, 14, 15, 58, 13] . Laser irradiation is applied to the thinnest part of the bracket center at one point in the center of the bracket for non-scanning lasing groups. The fiber tip of the waveguide is located consistently as close as possible from the labial surface of the polycrystalline ceramic bracket. Moreover, for groups that laser irradiation are performed by a scanning method, the application of 1940-nm Thulium Fiber Laser is done by the horizontal movement parallel to the bracket slot initiating from the upper distal wing of the bracket to the upper mesial wing,and then to the bracket's slot and the lower bracket wings (Figure 3.10) .

The aim of that in vitro experiment is to reduce the intrapulpal temperature increasing by the scanning method [54] and also to develop a different method to debond orthodontic ceramic brackets with a 1940 nm Thuluim Fiber Laser. The shear test,measurement of intrapulpal temperature changes and application of laser irradia-

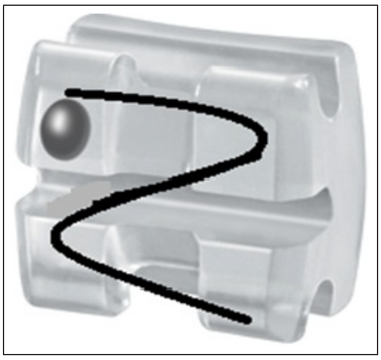

Figure 3.10 Indication of The Scanning Movement On The Ceramic Bracket.

tion are all done at the same time during experiment . Unlike the laser groups, the first group is the control group with no laser application . In every group, 11 teeth specimens are used to evaluate the results properly. After measuring of the adjusted power of 1940- nm Thulium Fiber Laser, fiber tip of laser is replaced in the second part of the experimental set-up that the distance between the fiber tip. Silicon thermal paste (Bakir, R-1260 Silicon Gress, Turkey) is applied by hand into the lingual cavity of every sample to mimic the intrapulpal tissue and to set against the thermal side effects of laser irradiation during experiment (Figure 3.11).

Figure 3.11 Silicon Thermal Paste to Mimic The Tissue.

Then, the specimens with the gypsum block are immobilized around by three screws to the testing frame. The shearing blade is placed and fixed properly on the base of the bracket's gingival wing. The K-Type thermocouple is located into the lingual cavity of tooth and in this respect one must sure that the tip of the thermocouple must touch on to the intrapulpal wall of the tooth. In the laser groups without scanning method application , the fiber tip (with a diameter of 2 mm) of the waveguide is located consistently as close as possible from the labial surface of the polycrystalline ceramic bracket and also laser irradiation is performed on the centre of the ceramic bracket surface which is the thinnest part of it . The position of fiber laser is fixed in the non-scanning laser groups in order to prevent the reflection of the laser light back through the fiber tip , so that the energy of laser could travel much more effectively to the adhesive agent (Figure 3.12) [9] .

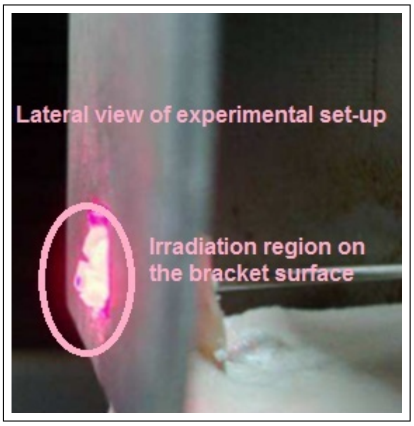

Figure 3.12 Lateral view of the experimental set-up after laser beam is located on the centre of the ceramic bracket surface.

In the study in Continous Wave (CW) Mode, the laser power is set to the determined value that was gained from preliminary studies. After that measurement and setting procedure of laser power, the universal testing machine is initiated to apply a torque (debonding) force on the ceramic bracket for SBS (shear bond strength) test. In SBS test, the crosshead speed of the universal testing machine was set to 1 mm/minute value. The torque force applied on the orthodontic ceramic bracket during experiment is named as debonding force. The universal testing machine is connected to a computer that collected the experiment data during debonding. While shearing test performing, an increasing the torque or debonding force is observed on the monitor of the computer, the selected group of lasing is manually initiated to apply onto the specimen in selected mode. Starting point is the determined moment for this study. At the moment of the debonding of the orthodontic ceramic bracket from the surface of each tooth, lasing and also the shearing test are stopped manually. This moment is named as the breaking point. Breaking load is accepted as the applied load during debonding. By computer that is connected to the universal testing machine, load at the breaking point during debonding is measured and saved for each tooth.

K-type thermocouple that measures the intrapulpal temperature changes during debonding is used in this study. This thermocouple is connected to the other computer with OMEGA program is permanently reading, recording and monitoring the collected data during laser irradiation by 1940 nm Thuluim Fiber Laser. The measurements are done for every 2 seconds automatically. The exact temperature values on the excel sheet are read to determine debonding time much more properly. The difference between the final and also the highest temperature and the temperature at the starting point that is the point for where temperature started to increase gave the intrapulpal temperature change during debonding (Figure 3.13) .

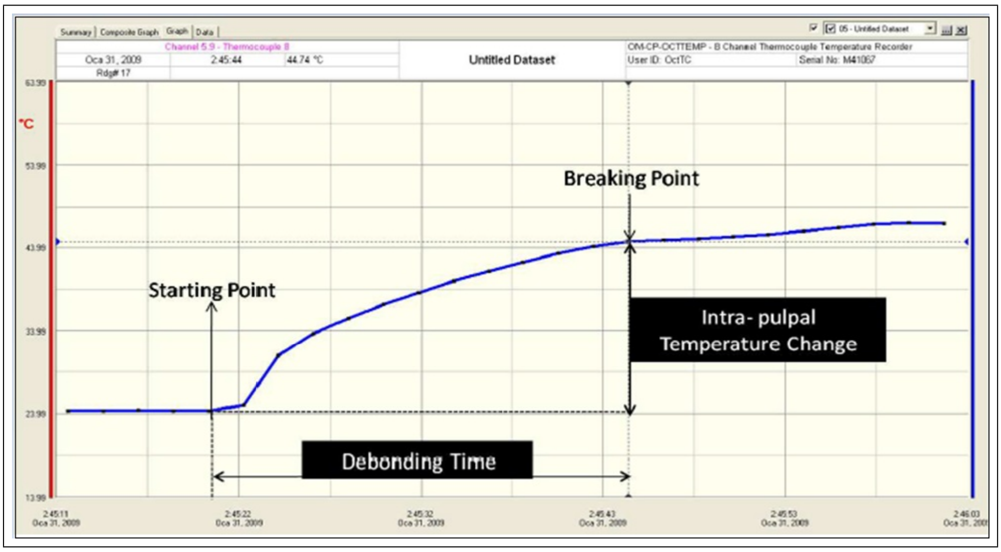

Figure 3.13 Starting point, breaking point, debonding time and intra-pulpal temperature change are indicated on the graph on the monitor of the computer which is connected to the K-type thermocouple.
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The load in order to debond the ceramic bracket,the intrapulpal temperature changes during irradiation and the breaking time for ceramic bracket are interpreted after experiment. Standard deviations and average or mean values for intrapulpal temperature changes collected by K-Type thermocouple,residual load need to debond the bracket are recorded by universal testing machine and breaking time are calculated for each group. After all calculations, a T-test is performed to determine statistically significant differences.The statistical significance level is accepted at $p<0.05$ for this study for confidence.universal testing machine and breaking time are calculated for each group. After all calculations, a T-test is performed to determine statistically significant differences.The statistical significance level is accepted at $p<0.05$ for this study for confidence. After debonding performed, the orthodontic ceramic bracket bases and enamel surfaces are examined and post lasing photos of the brackets are taken . Failures of brackets are reported. Amount of the adhesive remnant on the enamel surfaces after debonding is interpreted .

4. RESULTS

4.1 Laser Power,Laser Duration and Application Method

Applied Laser Energy (Joule)	Laser Power (Watt)	Laser Duration (seconds)	Application Method
30	3.0	10	Scanning / Non-Scanning
25	2.5	10	Scanning / Non-Scanning
21	3.0	7	Scanning / Non-Scanning
17.5	2.5	7	Scanning / Non-Scanning
No laser application	-	-	-

Figure 4.1 Groups of Experiment in different Laser Power, Laser Durations and Application Methods.

Experiments are performed in continuous wave (CW) Mode, laser is applied on samples with an arranged constant power in CW mode in different laser powers. The results of the preliminary studies determine the suitable laser parameters for debonding procedure by 1940-nm Thuluim Fiber Laser. In this study, experiments are done and evaluated at rates between 17. 5 and 30 J laser energy levels. The laser energy applied on to the samples is calculated by the measured output power level of laser and irradiation time. Laser parameters include laser duration, application method (scanning or non-scanning) and power of 1940-nm Thuluim Fiber Laser . Then according to laser parameters and laser energy levels , samples are divided into 9 different groups. One of these groups is the control group or nonlasing group. In control group, debonding of the orthodontic ceramic brackets is performed without laser application (Figure 4.1). Basal area of polycrystalline ceramic brackets are calculated 10.35 ± 0.41 mm^2

4.2 Load at Breaking Point

Shear bond stregth (SBS) test or tension test is done by universal testing machine. Maximum value at the debonding moment of orthodontic ceramic brackets for load is named as "Load at breaking point" . Then, means and standard deviations for all groups are calculated. The calculated results are given in the figure below (Figure-4.2) .

Groups	Mean	S.D	Min. / Max.	Irradiation Method	Laser Duration (sec.)
Control Group	69,61	15,26	47,8 / 69.61	Non-lasing group	-
3 W*	35,44	18,30	7,8 / 45,2	Scanning	10
3 W*	21,7	8,26	8,7 /32,2	Non- scanning	10
2.5 W*	44,52	14,79	26,87 / 63,86	Scanning	10
2.5 W*	36,68	10,15	18,96 / 49,46	Non-scanning	10
3 W	44,79	16,34	12,93 / 64,73	Scanning	7
3 W	56,10	22,25	16,06 / 83,14	Non- scanning	7
2.5 W	68,28	17,62	41,74 / 85,43	Scanning	7
2.5 W	60,36	25,33	12,78 /94,92	Non-scanning	7

Figure 4.2 Statistics of shear bond strength value of each group (Newton) . Stars (*) indicate that lasing groups are significantly different from control group ($p < 0.05$)(student t-test).

In CW mode experiments by 1940-nm Thulium Fiber Laser , both 3.0 W 10 seconds scanning and non-scanning groups are debonded with much more significant and lower debonding force than the nonlasing group . Among other laser groups, both 2.5 W 10 seconds scanning and non-scanning groups also have significantly lower force compared to control group ($p < 0.05$, student t-test) . In these four laser irradiated groups that are explained above, it is reported that Laser groups having the energy of 25J or above have been found to be effective in terms of debonding force . Reduction in debonding force is almost more than 50 %. When same laser durations and same methods but different laser powers are applied on the samples , application method does not create significant differences and reductions in the debonding force. In addition , when laser power is increased from lower level laser energy to higher level laser energy (i.e from 2.5 W to 3.0 W) ,debonding force is significantly decreased ($p < 0.05$, student t-test) to desired values in accordance with the control group. Results in different laser durations but with same configurations (laser power and application method) reveal that longer application time of laser irradiation in the studied group causes significant reduction in the debonding force when all other parameters are kept constant. ($p < 0.05$, student t-test) . As a conclusion, after assesment of different application methods of 1940-nm Thulium Fiber Laser , no effective debonding force reduction is observed among all studied groups . When application time of laser light is increased in the groups that have same configurations , debonding force is significantly reduced .

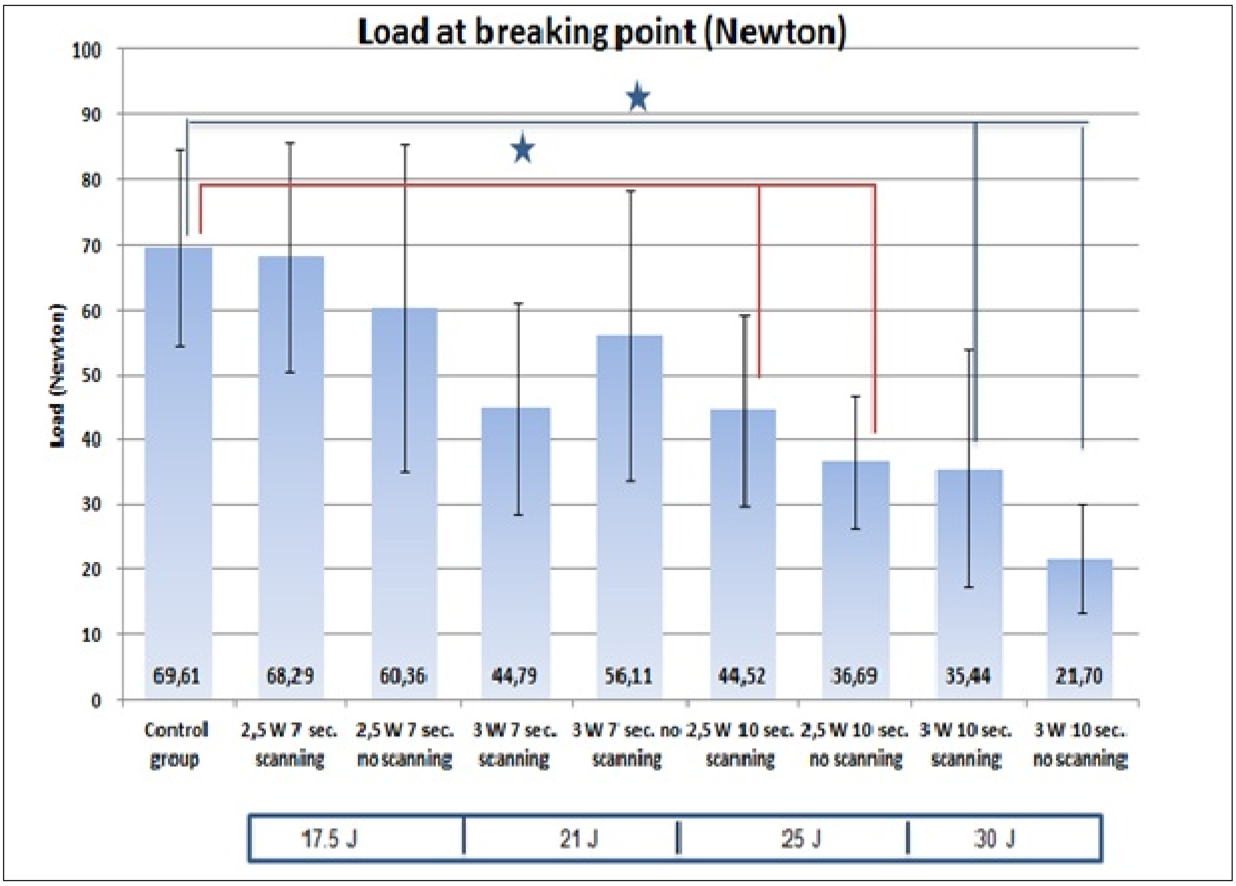

Figure 4.3 Average and standard deviation of load at breaking point for each CW mode group are shown on graph above. Group 3 W 10 sec. with scanning and no scanning are significantly different than Control Group ($p < 0.05$, student t-test).

In addition, increase in the laser power with other stable laser parameters in the group reports significant reduction in the required debonding force. In this study, these findings confirm that laser irradiation could decrease the load applied for orthodontic ceramic bracket removing compared to the nonlasing group (Figure-4.3) .

4.3 Intra-pulpal Temperature Changes

Laser System	Laser Power (Watt)	Laser Duration (sec.)	Mean Temp.Increase (°C)	S.D	Irradiation Method
1940-nm Thulium Fiber Laser	3	10	6,21	3,45	Scanning
	3	10	5,52	3,66	No scanning
	2.5	10	5,57	2,06	Scanning
	2.5	10	5,10	2,17	No scanning
	3	7	3,22	1,34	Scanning
	3	7	5,16	3,75	No scanning
	2.5	7	6,50	3,19	Scanning
	2.5	7	4,18	1,31	No scanning

Figure 4.4 Laser power , irradiation durations and average temperature rise during laser irradiation for brackets 1940-nm Thulium Fiber Laser ($p < 0.05$, student t-test).

During laser application in the experiment , the intrapulpal temperature changes inside the pulp are measured and saved by K-type thermocouple. Intrapulpal temperature change is temperature difference between measured temperature at the beginnning of the irradiation and at the breaking or debonding moment. In this study, 5.5 ± °C is accepted as a benchmark value for all specimens in order to prevent pulpal damage during laser application according to Zach and Cohen's studies [1]. Mean and standard deviations for intrapulpal temperature changes during laser irradiation for each group are given in (Figure-4.4).

In this study, when comparison of the results are done between two lasing groups :3.0 W 10 seconds scanning and 3.0 W 7 seconds scanning groups, irradiated samples exhibit significant difference in intrapulpal temperature change ($p < 0.05$,student t-test) . For 3.0 W laser power group , longer laser duration (10 s) causes higher increase in intrapulpal temperature with same laser power and application method . The intrapulpal temperature changes for 3.0 W 7 seconds scanning and non-scanning,2.5 W 7 seconds and 2.5 W 10 seconds non-scanning irradiated groups are below the safety threshold value that is equal to 5.5 ± °C .

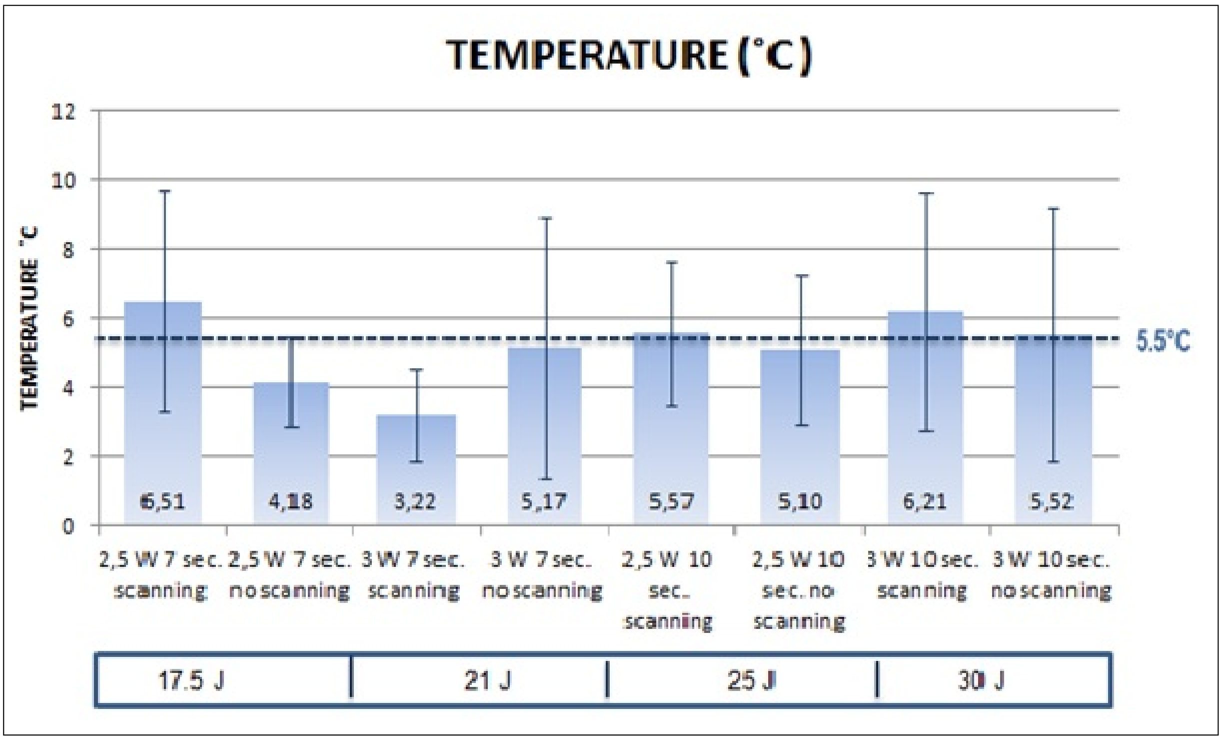

Figure 4.5 Graph of the temperature increse of the pulp chamber wall at lasing in CW mode with 1940-nm Thulium Fiber Laser . ($p < 0.05$, student t-test)..

In addition, intrapulpal temperature increase for both 3.0 W 10 seconds and 2.5

W 10 seconds non-scanning laser groups have almost close to temperature value of the threshold level (Figure-4.5) . Moreover , almost more than 50 % of the lasing groups have good and closer results when compared to the threshold value.

4.4 Examination of Enamel Surface and Bracket Base

In the evaluation of the enamel surfaces and bracket bases, in more than 50% of samples with energies 25 J or more, adhesive remnant hasn't been observed on enamel surfaces for the laser groups. In the bases of brackets , adhesive remnant has been observed (Figure 4.6) .

Figure 4.6 Indication of Remnant Adhesive Material On the Bracket Base.

And also the bracket's wing broken and carbonization effect on the bracket base are rarely observed during the debonding procedure (Figure-4.7) [9].

Most of the surfaces are clean and out of remnant adhesive material (Figure 4.8).

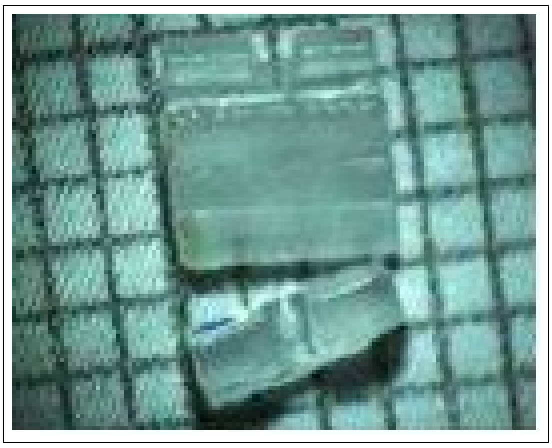

Figure 4.7 Broken Wing Of The Ceramic Bracket .

Figure 4.8 Enamel Surface After Debonding Procedure .

4.5 Summary

1. In this study , both 3 W 10 seconds scanning and non-scanning lasing groups need significantly less debonding force than non-lasing group . These laser groups have the best findings among all other laser irradiated groups due to the needed debonding or torque force. Moreover , both 2.5 W 10 seconds scanning and non-scanning lasing groups also have significantly lower force according to control group ($p < 0.05$,student t-test) .

2. Increase in the power of laser causes to observe lower debonding force than the nonlasing group.

3. Results in different laser durations but with same laser power and application method exhibit that longer duration of laser irradiation in the lasing group leads to significant decrease in the debonding force when all other configurations are kept constant. ($p < 0.05$, student t-test) .

4. Irradiation of the polycrystalline orthodontic ceramic brackets by 1940-nm Thulium Fiber Laser with different application methods do not create any remarkable differences between laser groups .

5. Increase in intrapulpal temperature was generally below the accepted threshold value 5.5 °C in most lasing groups .

6. In the evaluation of the enamel surfaces and bracket bases , in more than 50 % of samples with energies 25 J or more, adhesive remnant hasn't been observed on enamel surfaces for the laser groups.

5. Discussion

Previous studies have shown that dental lasers can significanlty decrease the required debonding force to remove orthodontic ceramic brackets from the surface of the enamel. In the present study, we debond the polycrystalline ceramic brackets by using previously untested 1940-nm Thulium Fiber Laser . We seek to investigate the side effects and advantages of this laser in removing of the ceramic brackets . The selected laser for this study is 1940-nm Thulium Fiber Laser because of the advantages of fiber lasers over other types and properties of that wavelength. In addition, fiber lasers are easier to use because they have extended lifetime and compact size. They show high vibrational stability. Light is already coupled into a flexible fiber , it means that the light is already in a fiber allows it to be easily delivered to a movable focusing element. Moreover, fiber lasers can have active mediums several kilometers long, and thus can provide very high optical gain. According to the properties of the wavelength of this laser,it is known that the minimal value for absorption coefficent of hydroxapatite and enamel is observed in visible and near infra-red region (NIR) . The scanning movement is preferred in this study in order to transmit the heat to the total surface of the ceramic brackets and thus preventing large rise in intrapulpal temperature changes during laser irradiation [54]. Laser irradiation is required to soften the adhesive material in the bonding interface between enamel and adhesive agent. Any process that degrades the bonding resin makes the debonding procedure easier. It should be positively efficent on the adhesive material composite resin melt it and make lower the cohesive enamel strength. All in all, from the physical and chemical point of view, the bond between the composite resin and the enamel could be broken by laser irradiation. According to the study of Tocchio et al, debonding mechanisms that enable adhesive material to degrade by laser energy can be achieved by thermal softening, thermal ablation and photoablation. The type and the efficency of interaction depends on the wavelength and power density of the laser radiation used. Decomposition of the adhesive material is gained by heat transmitted through the orthodontic bracket in thermal softening. Laser energy is used until the adhesive material softens in thermal softening. When adhesive agent degrades, the ceramic bracket cannot overcome the gravity. As a result, it drops from the surface of the enamel. The fast irradiation is the crucial point in thermal ablation because it is induced by high-peak lasing and a sudden temperature increase. Photoablation occurs when the high energy laser affects the energy level bonds between the adhesive resin. The result is decomposition of the material. Thermal softening is the most accepted and used interaction during ceramic bracket debonding. Thermal softening is accepted as a responsible debonding mechanism for this study because softened adhesive agent on the base of the ceramic brackets after debonding process were observed results of this mechanism. On the other hand, thermal softening can cause excessive rise in temperature because it is slower process when compared to the photoablation and thermal ablation. Neither thermal ablation nor photoablation was observed during the present study. The fibrous form of the softened adhesive material was observed on the base of the bracket after the debonding process, thus thermal softening is accepted as being responsible for the debonding mechanism in this study. By using the results of this experiment , it can be concluded that this study agree with the study of Tocchio et al., Rickabaugh et al. and Strobl et al. [3, 33, 50] . According to their studies, laser irradiation during debonding procedure can effectively and thermally soften the adhesive resin to cause ceramic bracket removing. Ma et al and Rickabaugh et al. used carbondioxide laser and debonding pliers to remove the ceramic brackets in their study. They stated that the bracket could be debonded from the enamel with pliers as soon as the adhesive degradation temperature had been reached. Mimura et al. [18] and Obata et al. [50] stated both thermal softening and resin contraction from orthodontic ceramic brackets could be responsible for the mechanisim of debonding. On the contrary, Hayakawa started debonding after lasing, not during the lasing. He mentioned that the mechanism of laser debonding was not traditional thermal softening because they observed some specimens debonded immediately after laser irradiation without mechanical effects [30] . In this study , the bracket debonding is performed at the same time with the temperature measurement and laser irradiation because the results of the previous studies stated that a 1- min debonding interval between irradiation of laser and debonding force application resulted in higher torque force [66] . Moreover, it is one of the major advantages to apply a debonding force during the laser irradiation in order to minimize unnecesseary heat transfer from the ceramic bracket to the tooth. Findings are assessed by observing the load at breaking points, intrapulpal temperature changes, applied laser energies and different irradiation methods .

5.1 Load at Breaking Point

All the brackets are debonded under the influence of the applied load . The usage of the 1940-nm Thulium Fiber Laser in ceramic bracket debonding has been found effective in reduction of the needed debonding force in both 25 -30 J lasing groups during the irradiation of polycrystalline ceramic brackets . Experiments are done in CW (continous-wave) mode. The laser energy applied on to the samples is calculated by the measured output power level of laser and irradiation time. In the study of Pickett [59] , the differences between in vivo and in vitro studies were investigated . As a result of this study , it was reported that in vitro bond strength values might be higher than those obtained in vivo . Thus ,it is normal to obtain higher debonding force values in vitro studies those than gained in vivo studies . In present vitro study , an average force of 69.61 ± 15.26 N is required to debond the ceramic brackets without lasing . Both 3 W 10 seconds non- scanning and scanning lasing groups (30 J) produce significanlty the best declined debonding force with 1940-nm Thulium Fiber Laser application compared to the control group . In other levels of lasing , both 2.5 W 10 seconds non-scanning and scanning laser groups (25 J) also have significantly considerable decrease in bond strength when compared to the nonlasing group . Statistical analysis indicates the significant differences to be at the 0.05

5.2 Effects of Application Methods In Intrapulpal Temperature Changes and SBS Values

In current study, in order to reduce the heat transmission to the pulp of the tooth, the side effects of the laser energy are tried to be decreased by scanning application method through the surface of the ceramic bracket in four laser groups. When all results of presented study are examined , no significant differences are observed between two configurations : scanning applied and not applied groups due to intrapulpal temperature changes and required debonding forces ($p < 0.05$ student t-test). Conversely , results of the proposed study are not consistent with the study of M. Oğuz Öztoprak et al. [54], in the previous study it was reported that the application of the Er:YAG laser (2940-nm) with the scanning method was effective for debonding of ceramic brackets by degrading the adhesive through thermal softening and also this laser might be an effective way to reduce the debonding force from higher values to the desired level.

5.3 Thermal Effects on Pulp Chamber During Laser Irradiation

Hayakawa [7] reported that the temperature of the pulp wall initiated to increase to its maximum point immediately after irradiation by laser. Unlike the given conclusion by Hayakawa , Obata [50] reported that the temperature rise in the pulp chamber starts 3 seconds after lasing. The average temperature changes of the pulp walls of the laser groups were compared with the results of previous studies of Zach and Cohen . According to their study, no pulp damage was found with an intrapulpal temperature increase of 1.8 °C when laser irradiation were applied on the samples. The histological study of Zach and Cohen on monkeys [39] showed that the increase in intrapulpal temperature changes should be below 5.5 °C to prevent irreversible side effects on pulpal walls. Also, it had been concluded that an increase in intrapulpal temperature of 11.1 °C , 60 % teeth showed abscess formation. At an 16.6 °C elevation pulpal necrosis occurred in all of the teeth. In the proposed study, 5.5 °C temperature increase during laser irradiation is accepted as a threshold or benchmark value . Only between 3W 10 seconds scanning and 3W 7 seconds scanning applied groups , irradiated samples exhibit significant difference in intrapulpal temperature change. ($p < 0.05$,student t-test) . Desired intrapulpal temperature changes during debonding by 1940-nm Thulium Fiber Laser are observed in shorter laser duration in the 3W lasing group . Another type of comparision is the study of Ma et al. , a linear relationship between lasing time and an increase in intrapulpal temperature were reported. Polycrystalline ceramic brackets were lased by carbondioxide laser at a power level of 18W in their study in CW mode. After 1 second irradiation, an average intrapulpal temperature increase of 0.91 °C were observed. Intrapulpal temperature changes after 2 seconds 1.74 °C and 2, 67 °C after 3 seconds were reported [37] . On the contrary to the study of Ma et al. , we are not able to gain the exact linear relationship in the assessment of the change in temperature . It might be the results of the different application methods and immobile application of the laser irradiation during scanning method. 3W 10 seconds scanning applied and no scanning applied , 2.5 W 7 seconds scanning applied , 2.5 W 10 seconds scanning applied lasing groups have an average pulp chamber wall temperature increase that is above 5.5 °C . In these scanning applied lasing groups , undesired increase in temperature change could be the result of the mobile and manuel application of the fiber laser tip through scanning . When the fiber tip of laser is immobilized , increase in intrapulpal temperature changes during irradiation have more desired values compared to the other lasing groups. Moreover, undesired increase in intrapulpal temperature change during laser irradiation migh be explained with the crystal structure of polycrystalline ceramic brackets. Energy transmissibility of them might not be efficent enough . The maximum rise in the present study is 6.51 °C in 17.5 J scanning applied laser group . There is no significant difference in intrapulpal temperature changes between irradiated groups except 3 W 10 seconds scanning and 3 W 7 seconds scanning applied lasing groups. No significant differences in temperature changes are observed between both different application methods. ($p < 0.05$ student t-test) .

5.4 Applied Laser Energy

The applied laser energy on to the specimens is calculated by the measured output power level of laser and laser duration. In the present study , applied laser energy levels are selected between 17.5 - 30 J values. In higher laser energy levels irradiation , lower and significant debonding forces than the needed debonding forces of nonlasing groups are observed to remove the polycrystalline ceramic brackets . When laser energy levels are decreased , higher debonding or torque forces are obtained. In 30 Joule scanning applied and not applied lasing groups, significant decrease in debonding forces are observed when compared to the results of control or nonlasing group. Moreover , in 25 J scanning applied and not applied groups , apparent and significant decrease in required torque forces are obtained compared to the control group. Besides these laser energy levels , no significant decrease in debonding force is observed . In the study of Hayakawa [4] , lasing groups have a significant decrease in debonding force when compared to the nonlasing and lower energy level groups. Hayakawa used an Nd : YAG laser (1940 nm) with three different laser energy levels :1 , 2 or 3 Joule. The 3 J laser applied group indicates lower and significant bond strentgh compared to the control group and 1 J lasin groups. Hayakawa [7] explained debonding was achieved more effectively in the 3 J group because the higher output level meant less energy loss during transmission through the polycrystalline brackets; the bond surfaces received the laser energy more effectively . In higher laser energy levels of the proposed study , lower debonding forces are observed to remove the polycrystalline ceramic brackets. When laser energy levels are decreased , higher debonding or torque forces are obtained. All in all, it is revealed that findings of the present study are in agreement with the study of Hayakawa.

5.5 Examination of Enamel and Bracket Surface

In evaluating the probable risks of enamel damage , assesment of the adhesive remnant on the enamel surface is important criterion .In more than 50 % of samples with energies 25 J or more, adhesive remnant hasn't been observed on enamel surfaces for the laser groups. It is accepted as a good result because of the fact that after debonding process we do not need a bur to clean the enamel surface . Using a bur to clean the surface can lead to enamel cracks . Thus ,1940-nm Thulium Fiber Laser irradiation for ceramic bracket debonding could be an effective way for degrading the adhesive material .

5.6 Conclusion and Further Works

1940-nm Thulium Fiber Laser irradiation on polycrystalline ceramic brackets with scanning and non-scanning methods confirms the following results. According to the findings of the proposed study , it is obvious that lower debonding forces are observed in lasing groups compared to nonlasing or control group. All in all , 3W 10 seconds non-scanning and scanning lasing groups (30 J) had the best significant decrease in debonding force with 1940-nm Thulium Fiber Laser application compared to the control group . Lasing of the samples with scanning or non-scanning method leads to almost more than 50 % reduction in the needed load for removing of the orthodontic ceramic brackets . Different application methods during laser irradiation on the samples do not create any significant differences and advantages due to debonding force and intrapulpal temperature increase. Longer application time of the selected laser is effective in reduction of the debonding force when the same laser power is applied on the specimens. In the evaluation of the intrapulpal temperature change between 17.5 and 30 J laser energy levels , the maximum rise is 6.51 °C in 17.5 J scanning applied laser group. Moreover , the most desired intrapulpal temperature increase is observed in 3.0 W 7 seconds scanning applied laser groups . The minimum rise is 3.22 °C that is below the benchmark value of 5.5 °C . No significant difference is observed due to intrapulpal temperature rise among laser groups except 3W 10 seconds and 3W 7 seconds scanning applied lasing groups . So , according to the results , application of longer laser duration causes higher and undesired temperature increase in the pulp walls between these laser groups. On the other hand , in lower laser energy levels , it is expected to have desired intrapulpal temperature rise but it could not be observed properly in 2.5 W 10 seconds scanning applied laser group . This can be the result of the manual application of the fiber laser tip by scanning method. Moreover , undesired rise in intrapulpal temperature change during irradiation of the samples might be explained with the crystal structure of polycrystalline ceramic brackets. Energy transmissibility of them might not be efficent enough. All in all , these issues must be studied further before the clinical use of this method . In order to obtain more desired values due to intrapulpal temperature changes in higher laser energy levels that are equal to or exceed 30 J , air cooling could be an effective process during laser-aided debonding of ceramic brackets . Water cooling might not be sufficent to obtain desired values because of the water absorption of 1940-nm Thulium Fiber Laser In ceramic bracket removing ,application of laser need to have further improvement because of the advantages of dental laser in clinical treatments. If debonding can be performed effectively with only lasing procedure , conventional methods that lead to patient discomfort and irreversible enamal damages become unnecessary.

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