CERAMIC BRACKET DEBONDING WITH INFRARED LASERS

by

AYŞE SENA KABAŞ SARP

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APPROVED BY:

Assoc. Prof. Dr. Murat GÜLSOY (Thesis Advisor)	....
Assoc. Prof. Dr. Ata AKIN	....
Prof. Dr. B. Cem ŞENER	....

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ABSTRACT

CERAMIC BRACKET DEBONDING WITH INFRARED LASERS

Orthodontics is a specialized branch of dentistry aiming to produce a healthy, functional bite, creating greater resistance to disease and improving personal appearance. Orthodontic brackets are small attachments used in orthodontics to fasten an arch wire. One of the types that used is ceramic brackets provide higher strength, more resistance to wear and deformation, better color stability and preferred for cosmetic reasons. After treatment ceramic brackets needs to be debonded from the enamel surface. Debonding may be unnecessarily time consuming and damaging to the enamel if performed with improper techniques or carelessly. There are several methods for debonding orthodontic brackets. All these techniques have their own advantages and limitations.

Since the early 1990s, lasers have been used experimentally for debonding ceramic brackets as a new and established method. Using Lasers in debonding procedure reduces required debonding force and risk of enamel damage but thermal effect during the laser radiation on dental tissues can cause undesirable results.

The aim of this study is to develop a better technique for ceramic bracket debonding. A new fiber laser (1070-nm Ytterbium fiber Laser (IPG Laser, YLM-20-SC, GmbH)) was tested, debonding procedure was quantified with a universal testing machine and intrapulpal temperature was monitored for limiting the injury or pain in present study .

Experiments were performed in two sections according to the type of lasing mode: Adjusted Laser power was applied in Continuous Wave (CW) and in Pulse Mode. Debonding force, debonding time and work done by universal testing machine was significantly decreased by irradiation in both sections. Lasing caused a 50 %

of reduction in required load for debonding and showed a 3- fold decrease in time. Intrapulpal temperature changes are below the accepted threshold value (5.5 °C) until the level of 3.5 watts of laser power in continuous wave mode. Also applying more than 3.5 watts of laser power showed a rapid increase in total applied laser energy. It can be reported that a sensible striding is observed after 88.6 joules of total energy applied on the ceramic brackets in both modes. Moreover, during debonding, the work done by universal testing machine is diminished up to 5 times by irradiation. Most of the groups in CW Mode and all groups are below the threshold value in pulse mode.

Laser applications in debonding require further improvement because Laser could mean very rapid and painless debonding without the risk of either enamel tear outs or bracket fractures. If debonding can be achieved with lasing alone, mechanical operations during bracket removal become unnecessary, alleviating patient discomfort at bracket removal.

Keywords: Laser, Debonding, Ceramic Brackets.

ÖZET

KIZILALTI LASER İLE SERAMİK BRAKETLERİN ÇIKARILMASI

Dişhekimliğinin bir dalı olan ortodontinin amacı sağlıklı, fonksiyonel bir ısırışın sağlanması, hastalıklara karşı direncin kuvvetlendirilmesi ve kişisel görünümün daha düzgün bir hale getirilmesidir. Ortodontik braketler ortodontide kullanılan ufak aygıtlardır. Bunlardan bir tanesi olan seramik braketler deformasyonlara karşı daha güçlü ve daha dayanıklıdır, renk kalıcılığı daha fazladır ve kozmetik açıdan daha çok tercih edilir.

Tedavi bitiminde seramik braketlerin mine yüzeyinden çıkarılması gerekir. Bunun için çeçitli metodlar vardır. Kullanılan bütün bu tekniklerin kendine özgü avantajları ve kısıtlamaları mevcuttur. Braketlerin çıkarılması esnasında uygun teknik uygulanmaz veya özensiz yapılırsa zaman kaybı çok olabilir ve mineye zarar verilebilir.

1990'ların başlarından beri yeni ve gelişmekte olan bir yöntem olarak laserlerin seramik braketlerin çıkarılmasında kullanılabilmesi için deneysel çalışmalar yapılmaktadır. Laser kullanımı braketin çıkarılması için gereken kuvveti azaltmakta ve mineye zarar verme riskini düşürmektedir ama ışıma sırasında oluşan ısıl etki diş dokuları üzerinde istenmeyen sonuçlara yol açabilir.

Bu çalışmanın amacı seramik braketlerin çıkarılmasında daha iyi bir teknik geliştirebilmekir. Bu çalışmada yeni bir fiber laser olan 1070-nm Ytterbium fiber Laser (IPG Laser, YLM-20-SC, GmbH) denenmiş, braketin çıkarılma işlemi bir çekme aleti tarafından değerlendirilmiş ve pulpa içi sıcaklık zarar verebilecek düzeyi belirlemek için gözlemlenmiştir.

Deneyler ayarlanmış olan ışıma şekline göre Sürekli (CW) ve Darbeli mod olmak üzere iki bölümde yapılmıştır. Her iki bölümde de braketin çıkarılması için gerekli olan kuvvette, zaman aralığında, çıkarılma esnasında çekme aleti tarafından yapılan iş miktarında kontrol grubuna kıyasla anlamlıolarak bir azalma vardır. Işıma kuvvette

Laser mine yüzeyinde yırtılma riskinin ve braket kırılmalarının azaltarak daha kısa zamandasonuca ulaştıran ağrısız bir yöntem olarak braket çıkarılmalarında uygulanabilinen geliştirilebilinecek bir yöntemdir. Bu işlem tek başına yapıldığı takdirde hastaların braket çıkarılması esnasında mekanik işlemler sonucu duyduğu rahatsızlık giderilecektir.

Anahtar Sözcükler: Laser, seramik braketlerin çıkarılması, seramik braket

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LIST OF ABBREVIATIONS

LASER	Light Amplification by the Stimulated Emission of Radiation
nm	nanometer
CW	Continuous Wave
°C	Celcius degree
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
CEJ	Cemento-Enamel Junction
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

Orthodontics is a specialized branch of dentistry aiming to produce a healthy, functional bite, creating greater resistance to disease and improving personal appearance. Orthodontic brackets are small attachments used in orthodontics to fasten an arch wire. Three types of attachments are presently available for orthodontic bracket bonding: plastic based, ceramic based, and metal (stainless steel, gold-coated, titanium) based. Ceramic brackets provide higher strength, more resistance to wear and deformation, better color stability and, most important to the patient superior aesthetics. After treatment ceramic brackets needs to be debonded from the enamel surface. Debonding orthodontic brackets means to remove the attachment and all the adhesive resin from the tooth and restore the surface as closely as possible to its pretreatment condition without any damage. Debonding may be unnecessarily time consuming and damaging to the enamel if performed with improper technique or carelessly. There are several methods for debonding orthodontic brackets. Due to their better aesthetic appearance ceramic brackets created many problems when removed with conventional debonding techniques. All these techniques have their own advantages and limitations. The use of lasers in debonding is a new and established method.

Starting from late 1960s, lasers were introduced to many medical areas. During 1980s and early 1990s the use of lasers was introduced into dentistry as various types were approved by the United States Food and Drug Administration (FDA). Dental lasers have been using in cavity preparation, removing of the fillings, surface etching for micro-retantion, removing of hard dental tissues, removing of soft dental tissues, lythotrypsy, root disinfection and cleaning in endodontics, laser welding of dental bridges and dentures, tooth bleaching, caries management and debonding ceramic brackets. Laser dentistry minimizes bleeding and bacterial infections. Procedures performed using dental lasers may not require sutures and wounds heal faster.

Also Certain laser dentistry procedures do not require anesthesia.

Since the early 1990s, lasers have been used experimentally for debonding ceramic brackets. Various lasers have been used in ceramic bracket removal like CO_2 (10600 nm), Nd:YAG (1060 nm), KrF (248 nm), XeCl (308 nm) Tm:YAP (1980 nm) and GaAlAs (808 nm). A 1070- nm Ytterbium fiber Laser (IPG Laser, YLM-20-SC, GmbH) is chosen for this study because of the advantages of fiber lasers over other types and properties of that wavelength. First of all fiber lasers are easy to use and economical because of their compact size and extended lifetime. Light is already coupled into a flexible so it can be easily delivered to a movable focusing element. Also a fiber laser has high output power that can provide very high optical gain. High optical quality of this type laser is also another advantage because the fiber's waveguiding properties reduce or eliminate thermal distortion of the optical path. 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. However, besides lots of advantages, thermal effect during the laser radiation on dental tissues can cause undesirable results.There are lots of studies about thermal effects of laser on dental tissues. According to the previous studies [5] no histological changes were discernible with an intrapulpal temperature increase of 1.8 °C. 5.5 °C was accepted as threshold temperature of pulpal damage in most studies. They have also found that an increase in intrapulpal temperature of 11.1 °C showed abscess formation. That is why 5.5 °C was accepted as threshold temperature in this study.

1.2 Outline

Chapter 2, gives general information about dentistry, orthodontics and lasers used in dentistry. Also, techniques used for bonding and debonding orthodontic brackets are explained in this chapter.

Chapter 3, gives detailed information about materials used in proposed study. And experimental setup and method of experiments are given in detail.

Chapter 4 includes results and Chapter 5 includes discussion of proposed study and finally in Chapter 6 conclusion and future works of the study are given.

2. BACKGROUND

2.1 Dentistry

Dentistry is a branch of medicine that involves diagnosis, prevention, and treatment of any disease of teeth, oral cavity, and associated structures.

Oral cavity is the inside of the mouth, bounded by the palate, teeth, and tongue. The upper jaw is called maxilla and the lower jaw is called mandible. The teeth of the upper arch are called maxillary teeth, because their roots are embedded within the alveolar process of the maxilla. Those of the lower arch are called mandibular teeth because their roots are embedded within the alveolar process of the mandible. Teeth have evolved different functions - incisors for biting, canines (eyetooth) for tearing, molars and premolars for chewing (Figure 2.1).

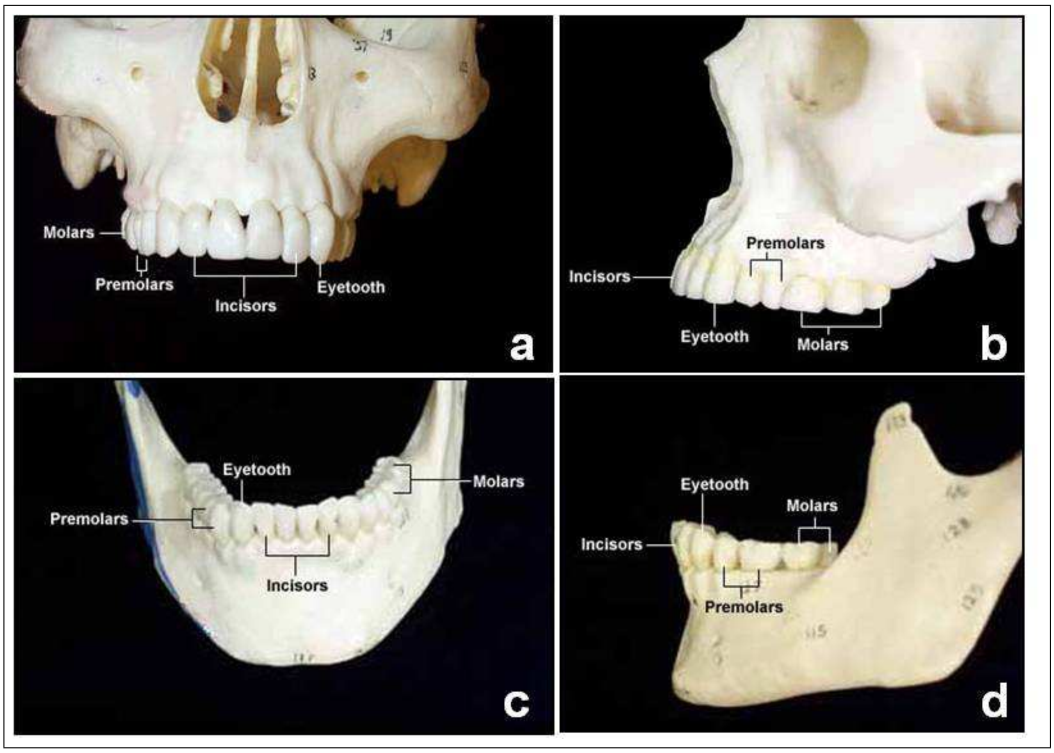


Figure 2.1 (a) Frontal view of maxilla, (b) Lateral view of maxilla, (c) Frontal view of mandibula, (d) Lateral view of mandibula.

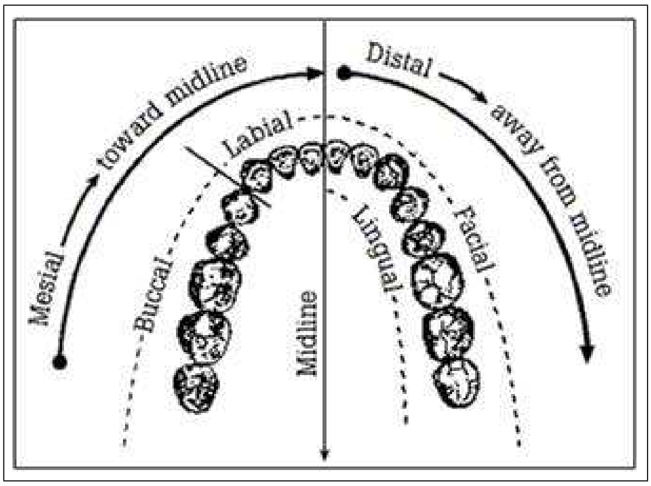


Figure 2.2 Midline, Mesial, Distal, Lingual, Labial, Facial, and Buccal terms are shown.

The imaginary plane which is accepted in the center dividing the dental arch right 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 and distal 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 or lips. Labial is facial surface of anterior teeth (toward the lips) and buccal means facial surface 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.

Each tooth has a core of pulp, surrounded by dentine, which is covered with enamel over the crown (exposed surface of the tooth). Anatomical crown is the portion of the tooth covered with enamel. Clinical crown is the visible part of the tooth above the gum line. Enamel makes up the anatomic crown. It is the hardest material in the human body which is capable of remodeling and repair. Dentin makes up bulk of tooth which is covered by enamel on crown and cementum on the root. It is not as hard as enamel. Dentin is often sensitive to cold, hot, air and touch (via dentinal tubules) if it is exposed. Cementum covers root of tooth and overlies the dentin and joins the enamel at the cemento-enamel junction (CEJ). Primary function of cementum is to

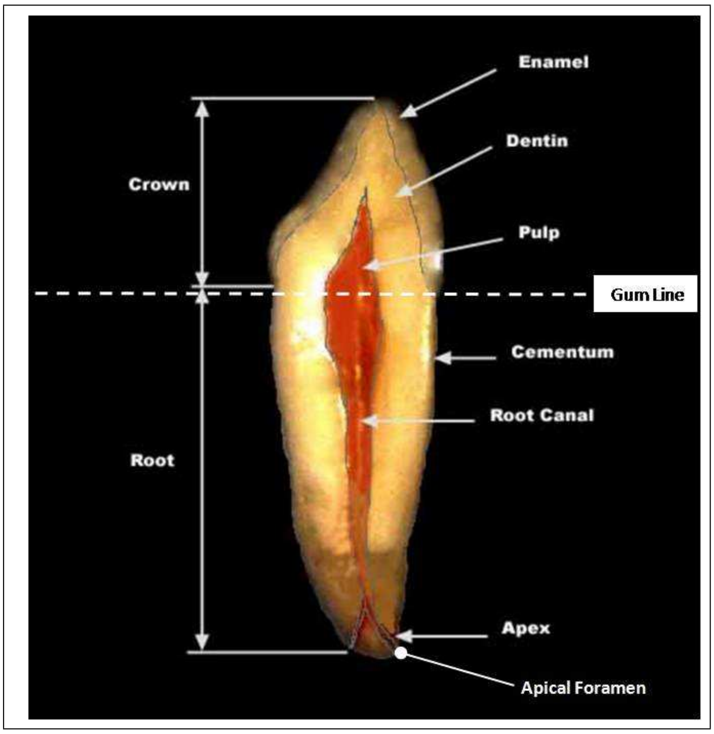


Figure 2.3 Anatomy of tooth.

anchor the tooth to the bony socket with attachment fibers. Root is the part of the tooth embedded in the alveolar process and covered by cementum. Apex is the end of root tip and apical foramen is the opening at the root tip. Pulp is made up of blood vessels and nerves entering through the apical foramen.It contain connective tissue, which aids interchange between pulp and dentin.

The composition of tooth structure is not homogeneous. The amounts of both organic and inorganic components present in dentin differ from the amounts of these components present in enamel. The hardest substance in the body, enamel as the cover of the crown, contains 95 % hydroxyapatite ($Ca_{10}(PO_4)_6(OH)_2$) , 4 % water and 1 % organic matter. The layer layer lying under the enamel, dentin, is much softer than the enamel and composed of 70 % hydroxyapatite ($Ca_{10}(PO_4)_6(OH)_2$), 20 % organic matter (mainly collagen fibers) and 10 % water (Figure 2.4). Hydroxyapatite is a mineralized compound with the chemical formula ($Ca_{10}(PO_4)_6(OH)_2$). Its substructure consists of tiny crystallites which form so called enamel prisms. Crystal lattice intruded by Cl^-, F^-, Na^+, K^+.

	Enamel	Dentin
Components	95 % hydroxyapatite 4 % water 1 % organic matter	70 % hydroxyapatite 10 % water 20 % organic matter

Figure 2.4 The amounts of both organic and inorganic components present in enamel and dentin.

There are several branches which are studying different subjects in dentistry:

1. Endodontics is root canal therapy and study of diseases of the dental pulp,
2. Oral and Maxillofacial Pathology study, diagnosis, and sometimes the treatment of oral and maxillofacial related diseases,
3. Oral and Maxillofacial Radiology study and radiologic interpretation of oral and maxillofacial diseases,
4. Oral and Maxillofacial Surgery study in extractions, implants, and facial surgery,
5. Periodontics study and treatment of diseases of the periodontium (non-surgical and surgical), and placement and maintenance of dental implants,
6. Pediatric Dentistry is a branch of dentistry for children, formerly known as "pedodontics",
7. Prosthodontics study in dentures, bridges and the restoration of implants,
8. Orthodontics and Dentofacial Orthopaedics,
9. Pain Management,
10. Biostimulation,
11. Operative dentistry light sured fillings and bleaching.

2.2 Orthodontics

The formal name of the specialty is orthodontics and dentofacial orthopedics. Orthodontics is a specialized branch of dentistry concerned with the development and management of irregularities and abnormalities of the teeth, jaws and face. Its aim is to produce a healthy, functional bite, creating greater resistance to disease and improving personal appearance. This contributes to mental and physical well-being. It is concerned with the diagnosis, supervision, guidance and correction of malocclusions. Orthodontic procedures are most commonly done on children but in recent years have become very popular for adults. Orthodontic treatment is possible at any age but the adult must be prepared to wear the braces as instructed. Movement of teeth may be slower.

The great technological advance occurred in the last years has brought a number of benefits to orthodontics. Research-based findings have constantly led to the development of new materials and techniques that are aimed at simplifying the clinical procedures. There are several different types of appliances used in orthodontics:

1. Dental braces are device used in orthodontics to align teeth and their position with regard to a person's bite. They are often used to correct malocclusions such as underbites, overbites, cross bites and open bites, or crooked teeth and various other flaws of teeth and jaws, whether cosmetic or structural. Orthodontic braces are often used in conjunction with other orthodontic appliances to widen the palate or jaws or otherwise shape the teeth and jaws. Teeth move through the use of force. Braces involve many different parts that work together to straighten your teeth.

2. An orthodontic archwire, going across the teeth from bracket to bracket, is a wire conforming to the alveolar or dental arch that can be used as a source of force in correcting irregularities in the position of the teeth with dental braces. An archwire can also be used in order to maintain the dental position; in this case it has a retention purpose. Orthodontic archwires can be fabricated with

different alloys. These are most commonly stainless steel, nickel-titanium alloy, and an alloy composed primarily of titanium and molybdenum (also called beta titanium).

3. Brackets are seen clearly on each tooth, holding the archwire Figure 2.5.

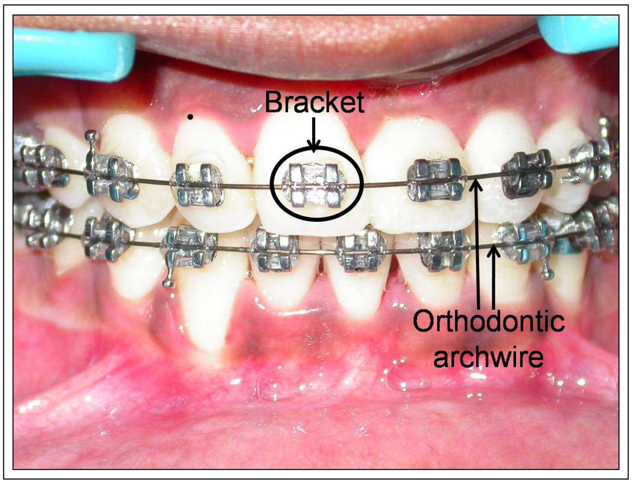


Figure 2.5 Bracket and orthodontic wire shown on an example of orthodontic treatment.

2.3 Bonding Orthodontic Brackets

Procedure of bonding orthodontic brackets on enamel is based on adhesion. Adhesion can be defined as the force that binds two dissimilar materials together when they are brought into intimate contact. This is distinct from cohesion which is the attraction between similar atoms or molecules within one substance [6]. The material or film used to cause adhesion is known as the adhesive; the material to which it is applied is called the adherend [7]. The performance of all dental materials, whether ceramic, polymeric or metallic is based on their atomic structure. [8]. Before bonding to a surface one must make sure it is both clean and dry, otherwise no adhesive bond will form. A clean and dry surface ensures that the adhesive has the best possible chance of creating a proper bond with the solid material. 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 [6]. Stresses that weaken adhesive bonds are caused by differences in thermal expansion coefficients [9]. Two mechanisms of adhesion may be distinguished: chemical and mechanical. Chemical adhesion involves bonding at atomic or molecular level. Mechanical adhesion is based on retention by interlocking or penetration of one phase into the surface of the other. In many cases chemical and mechanical adhesions occur together. Mechanical bonding may also involve other mechanisms such as the penetration of the adhesive into microscopic or submicroscopic irregularities (e.g. crevices and pores) in the surface of the substrate by acid-etching method. Adhesion with composites has been achieved by etching tooth enamel with acids such as phosphoric or acrylic acid [9]. Acid-etching technique process of roughening a solid surface by exposing it to an acid and thoroughly rinsing the residue to promote micromechanical bonding of an adhesive to the surface [7]. Typically 37 % is the preferred etching agent. Concentrations greater than 50 % result in the deposition of an adherent layer of monocalcium phosphate monohydrate on etched surface, which inhibits further dissolution [10]. The adhesion of resins to etched enamel is a result of capillary penetration into surface irregularities. These polymer projections into enamel have been called tags. Resin tags may penetrate 10 to 20 μm into the enamel porosity [11]. This micromechanical bonding mechanism has been commonly used in dentistry because of absence of truly adhesive cements or restorative materials [12]. A more recent example of mechanical debonding is that of resin restorative materials. The acid produces minute pores and other irregularities in the enamel surface into which the resin subsequently flows when it is placed into the preparation. The greatest problems associated with bonding to tooth surfaces are the in adequate removal of etching debris and contamination by water or saliva [13].

Monomer is a chemical compound capable of reacting to form a polymer which is a chemical compound consisting of large organic molecules formed by the union of many repeating smaller monomer units. Polymerization is a chemical reaction in which monomers of a low molecular weight are converted into chains of polymers with a high molecular weight. The physical properties of a polymer are influenced by changes in temperature and environment and by the composition structure, and molecular weight of the polymer. In general, the higher the temperature, the softer and weaker the polymer becomes [14].

There are 3 main groups of adhesives according to their polymerization types:

1. Chemically autopolymerizing paste-paste systems,
2. No-mix adhesives,
3. Light-polymerized adhesives.

Chemically activated polymerization is initiated by mixing two pastes just before use. On the other hand, polymerization with no-mix adhesives occurs with just one type paste.

There are two basic types of dental adhesive resins may be used for orthodontic bracket bonding according to their chemical properties. Both are polymers and are classified as acrylic or diacrylate resins. Both types of adhesive exist in filled or unfilled forms. Acrylic resin, polymethylmethacrylate is a transparent resin of remarkable clearity; it transmits light in the ultraviole range to a wave length of 250 nm. It softens nearly at 125 °C. Between 125 °C and 200 °C depolymerization takes place. At approximately 450 °C, 90 % of the polymer depolymerizes to form the monomer. Poly(methyl methacrylate) of high molecular weight degrades to a lower polymer at the same time that it converts to the monomer [15]. The backbone of the molecule formed in this system can have any shape, but methacrylate groups are found at the end of the chain. Most diacrylate resins are based on the acrylic modified epoxy resin that one of the first multifunctional methacrylates used in dentistry was Bowen‘s resin or Bis-GMA. The Bis-GMA can be described as an aromatic ester of a dimethacrylate, synthesized from an epoxy resin ethylene glycol of bisphenol-a and methylmethacrylate [15]. A fundamental difference is that resins of the first type form linear polymers only, whereas those of the second type may be polymerized also by cross linking into a three dimensional network. This crosslinking contributes to greater strength, lower water absorption, and less polymerization shrinkage [16].

There are some studies done about comparing the debonding adhesives. For example Rueggeberg and Lockwood [17] studied on ten commercial brands of orthodontic materials representing three modes of delivery systems: Two paste, no mix and power liquid types. They bonded stainless steel brackets on bovine teeth. They noted each temperature at debonding when heat was applied to the brackets. They mentioned that a higher temperature was required for two-paste systems than the no mix systems and also the power liquid types required lowest temperature value. In 1995, Mimura et al. [18] studied on the laser aided removal of ceramic brackets and compared two different adhesives. The selected bonding materials were Bis-GMA composite resin and 4-META MMA (4-methacryloxyethyl trimellitate anhydride) resin. They observed that for MMA resin, debonding force was sufficiently at a lowest power of energy than needed for Bis-GMA resin samples. More adhesives remained on the tooth surface in MMA group than Bis-GMA groups. They mentioned that debonding MMA resin with a laser is safer than debonding Bis-GMA resin with a laser.

Orthodontic brackets are small attachments used to fasten an arch wire. There are two wings, a base and a channel for replacing orthodontic archwire (Figure 2.6). These attachments are soldered or welded to an orthodontic band or cemented directly onto the teeth. Three types of attachments are presently available for orthodontic bracket bonding: plastic based, ceramic based, and metal (stainless steel, gold-coated, titanium) based.

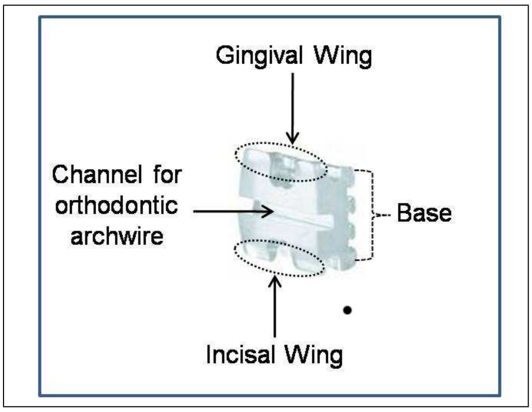


Figure 2.6 Parts of a bracket.

Metal brackets rely on mechanical retention for bonding, and mesh gauze is the conventional method of providing this retention (Figure 2.7). The use of small, less noticeable metal bases helps avoid gingival irritation. For the same reason, the base should be designed to follow the tissue contour along the 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. Corrosion of metal brackets may be a problem and black and gren stains have appeared with bonded stainless steel attachments. Crevice corrosion of the metal arising in areas of poor bonding may result from the type of stainless steel alloy used [19]. Because of the corrosion susceptibility of stainless steel interest is growing in the use of more corrosion-resistant and biocompatible bracket metals such as titanium.

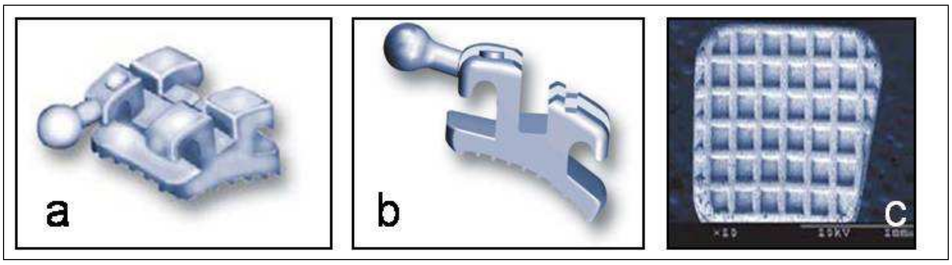


Figure 2.7 (a) Frontal and (b) lateral view of the Elite® OPTI-MIM® Mini-Twin® which is shown as an example of metal brackets, (c) Mesh Base Design of Opti-MIM® bracket base.

During the early 1970s, plastic brackets were marketed as the esthetic alternative to metal brackets. Newman [20] was the first to test the bonding of plastic attachments, polycarbonate brackets, to the buccal surfaces of the teeth and to divulge such a technique. Newman et al. reported that plastic brackets were not resistant enough, being easily fractured or distorted. These polycarbonate brackets constructed from acrylic and later polycarbonate quickly lost favor inherent problems were soon noticed, including staining and odors but more importantly their lack of strength and stiffness resulting in bonding problems, tie wing fractures and permanent deformation. Permanent deformation, or creep, occurs when a material is subjected to a constant load over an extended period of time and is particularly important for thermoplastic materials such as polycarbonate resins. Polycarbonate bracket slots distorted with time under a constant 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 [1].

To compensate for the lack of strength and rigidity of the original polycarbonate brackets, in the mid 1980s, the first ceramic brackets came into the field of orthodontics, offering many advantages over the traditional aesthetic appliances. As in Swartz studies [21], there is apparently a significant difference in tensile strength between ceramics and stainless steel. Also, they were introduced as an esthetic appliance which, unlike plastic brackets, could with stand most orthodontic forces and resist staining.

Ceramic brackets provide higher strength, more resistance to wear and deformation, better color stability and, preferred for cosmetic reasons.

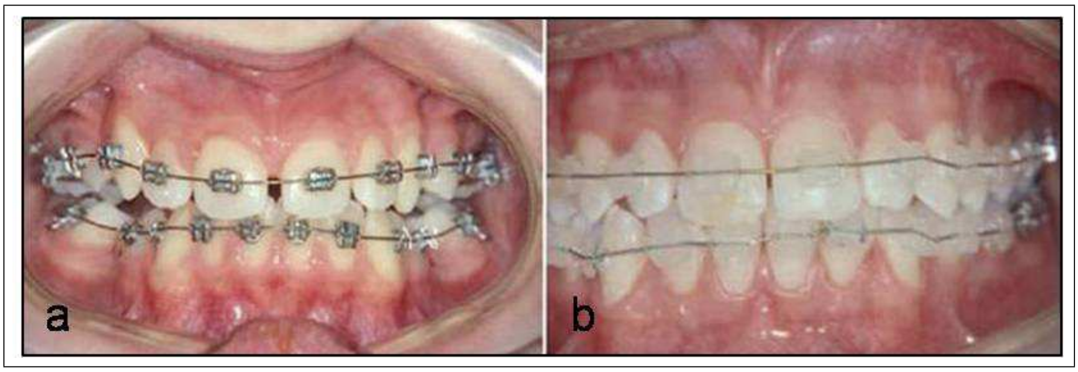


Figure 2.8 Esthetic comparison between bonded appliances, (a) Stainless steel brackets, (b) Ceramic brackets [21].

Ceramic orthodontic brackets are composed of pure alumina that formed when aluminum is added to steel to remove oxygen dissolved in the steel [21]. Ceramic brackets are demanding better aesthetics during treatment because pure aluminum oxide (alumina) is optically opaque when pure.

Ceramic orthodontic brackets are machined from monocrystalline or polycrystalline aluminum oxide. The first ceramic brackets were milled from single crystals of sapphire (monocrystalline) using diamond tools and monocrystalline ones contain a single crystal of aluminum oxide. These were closely followed by polycrystalline sapphire (alumina) brackets, which are manufactured and sintered using special binders to thermally fuse the particles together [1].

The manufacturing process plays a very important role in the clinical perfor-

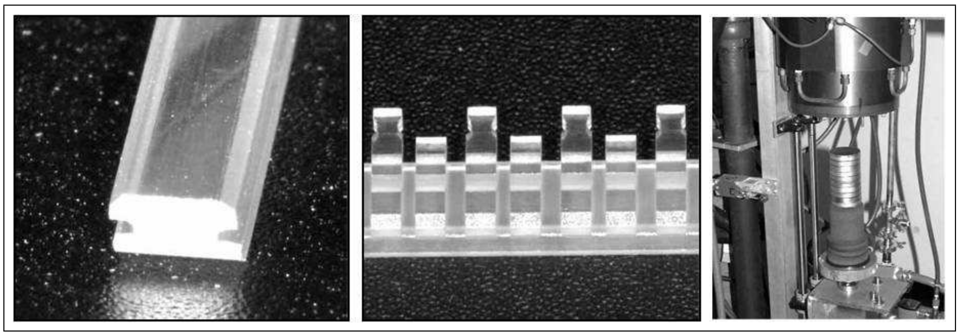


Figure 2.9 Manufacturing steps of ceramic orthodontic brackets, (a) a crystal sapphire rod is cut to shape in preparation of machining brackets, (b) the bracket slot and tie wings are machined into the rod (cuspid bracket with hook), (c) trays containing sapphire brackets are stacked in the furnace [1].

mance of the ceramic brackets. The production of polycrystalline brackets is less complicated, and thus these brackets are more readily available at present. The most apparent difference between polycrystalline and single crystal brackets is in their optical clarity. Single crystal brackets are noticeably clearer than polycrystalline brackets and hence are translucent. Fortunately, both single crystal and polycrystalline brackets resist staining and discoloration [22].

The physical properties of ceramics which are important to the orthodontics include hardness, tensile strength and fracture toughness or brittleness [22].

Ceramics used in orthodontic brackets have highly localized, directional atomic bonds. This oxidized atomic lattice does not permit shifting of bonds and redistribution of stress. When stresses reach critical levels, the interatomic bonds break and material failure occurs. This is called "brittle failure". Fracture toughness in ceramics is 20 to 40 times less than in stainless steel [23, 24], making it much easier to fracture a ceramic bracket than a metallic one. Among ceramic materials, polycrystalline alumina presents higher fracture toughness than single-crystal alumina [25, 26]. The brittle nature of ceramic brackets has resulted in a higher incidence of bracket failure (fracture) during debonding [27, 24, 28, 29]. The fracture toughness of the enamel is lower than that of ceramic [24] and ceramic brackets bonded to rigid, brittle enamel have little ability to absorb stress [23]. Enamel fracture or the appearance of fracture lines during debonding is related to the high bond strength of ceramic brackets and seems to be associated with sudden impact loading [30]. The combination of very hard and brittle properties and high bond strength leads to reports of two significant problems. One is bracket fracture specifically during debonding and another is enamel fracture which may occur during function [31] but mostly during debonding [21, 32].

Hardness is a very important physical property of ceramic brackets that is because of high hardness of aluminium oxide. It is a significant advantage to both ceramic bracket over stainless steel brackets. Swartz et al. showed that ceramic brackets are nine times harder than stainless steel brackets or enamel [33] and as Viazis said severe enamel abrasion from ceramic brackets might occur rapidly, if contacts between teeth and ceramic brackets exist [34].

As several studies showed that the tensile strength is much higher in monocrystalline alumina than in polycrystalline alumina, that is significantly more than stainless steel [35]. Tensile strength characteristics of ceramics depend on the condition of the surface of the ceramic [36]. A shallow scratch on the surface of a ceramic bracket reduces the load required 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 % under stress before fracturing, where as ceramic brackets deforms less than 1 % before failing [36].

Due to their advantages, ceramic brackets have some disadvantages too. Ceramic brackets cannot bond chemically with acrylic and diacrylate bonding adhesives due to their inert aluminium oxide composition. Consequently, the early ceramic brackets used a silane-coupling agent to act as a chemical mediator between the ceramic bracket base and the adhesive resins. This chemical retention resulted in extremely strong bonds that caused the enamel/adhesive interface to be stressed during debonding, risking irreversible enamel damage in the form of crack and delamination that often required dental restorations. Consequently, the challenge was to develop a bond between the ceramic bracket base and the enamel that clinically has adequate strength to accomplish treatment but can be broken at debond without 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 bonding agents. 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.

Several researchers have evaluated the bond strength of ceramic brackets with different retention mechanisms and concluded that mechanically retained brackets have adequate bond strength and appear to cause less enamel damage at debond compared to the chemically retained variety [23, 37]. Bond strength can also be modified by the choice of adhesive, different types of enamel conditioning [38] and different etch times [39]. The mean bond strength of metal reinforced brackets is reportedly significantly lower than conventional ceramic brackets and comparable with stainless steel brackets [40]. Polycrystalline ceramics, due to their rougher more porous surface, have a higher coefficient of friction than monocrystalline ceramics and stainless steel, which are comparable. Omana et al. [41] showed that polycrystalline brackets generate significantly greater frictional forces than stainless steel brackets. The frictional characteristics of polycrystalline ceramic brackets are worst with any archwire combination, whether bearing against stainless steel, nickel-titanium, cobalt-chromium or beta titanium archwires, when compared to stainless steel brackets [23, 42]. The low fracture toughness of ceramics leads to a higher incidence of bracket breakages than with stainless steel brackets. Tie wings can easily fracture due to the high torsional forces in ceramic brackets. Also, ceramics are radiolucent and if swallowed or inhaled would not be visible on the radiograph. In addition, bonding ceramic brackets to compromised teeth e.g. endodontically treated teeth, enamel hypoplasia and cracks, or those with large restorations should be avoided if possible, thus reducing the risk of tooth damage during bracket removal [31].

There are two types of ceramic bracket bases available. One type of bracket base is formed with undercuts or grooves that provide a mechanical interlock to the adhesive. The mechanical retention of such brackets is less as compared to other bracket base that are having both micromechanical retention and chemical adhesion. The other type of bracket base has a smooth surface and relies on a chemical coating to enhance bond strength. A silane coupling agent is used as a chemical mediator between the adhesive resin and the bracket base. It has been claimed that chemical adhesion provided higher bond strength when compared with mechanical retention [43]. The brackets that are used in this study have a base type that provide a mechanical retention as well as a chemical coating was used on base to enhance the bond strength. A chemo/mechanical adhesion was used for dimpled based ceramic brackets in this present study.

Bonding of orthodontic brackets has accepted as a clinical technique since 1970 [44]. Bonding has replaced banding and this technique has a lot of advantages compared to traditional banding system including better aesthetics, better hygiene and less time for doctor. However, bond failure is the biggest weakness for this technique.The bonding procedure is based on enamel alteration created by acid etching of enamel as developed by Buonocore in 1955 [2]. The steps involved in direct and indirect bracket bonding on facial or lingual surfaces are as follows:

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

Cleaning of the teeth with pumice removes plaque and the organic pellicle that normally covers all teeth. After the teeth rinsed and the operative field has been isolated, the conditioning solution or gel is applied over the enamel surface for 15 to 30 seconds. At the end of the etching period the etchant is rinsed off the teeth with abundant water spray. A routine etching removes from 3 to 10 μm of surface enamel [2]. After the teeth are completely dry and frosty white, a thin layer of bonding agent (sealant, primer) may be painted over the etched enamel surface. Bracket placement should be started immediately after all etched surfaces are coated. Excess adhesive must be removed (especially along the gingival margin) with the scaler before the adhesive has set.

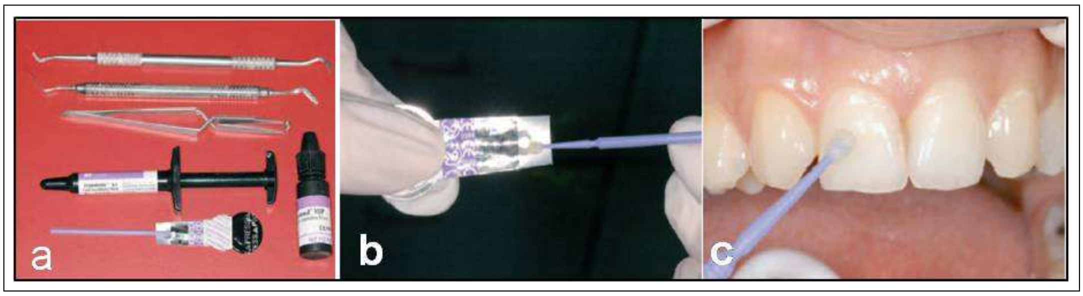


Figure 2.10 (a) Instruments used for bracket bonding with self-etching primer (Transbond Plus) and light-initiated adhesive resin (Transbond XT), (b) and (c) Application of self-etching primer (Transbond Plus) on enamel surface of maxillary incisor [2].

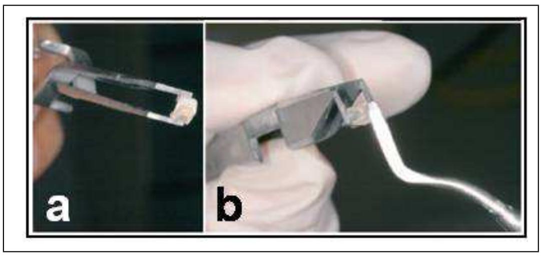


Figure 2.11 Placement of chemically curing adhesive resin on contact surface of bracket [2].

Figure 2.12 Direct bracket bonding [2].

2.4 Debonding Orthodontic Brackets

After treatment ceramic brackets needs to be debonded from the enamel surface. Debonding orthodontic brackets means to remove the attachment and all the adhesive resin from the tooth and restore the surface as closely as possible to its pretreatment condition without any damage. Debonding may be unnecessarily time consuming and damaging to the enamel if performed with improper technique or carelessly [45]. There are several methods for debonding orthodontic brackets:

1. Conventional method: Debonding with pliers (a- lift-off pliers, b- Hows or Weingart pliers and ligature cutters, c- Special debonding pliers),

2. Debonding with electrothermal unit,

3. Debonding with ultrasonic unit,

4. Debonding with debonding agents,

5. Debonding with Lasers.

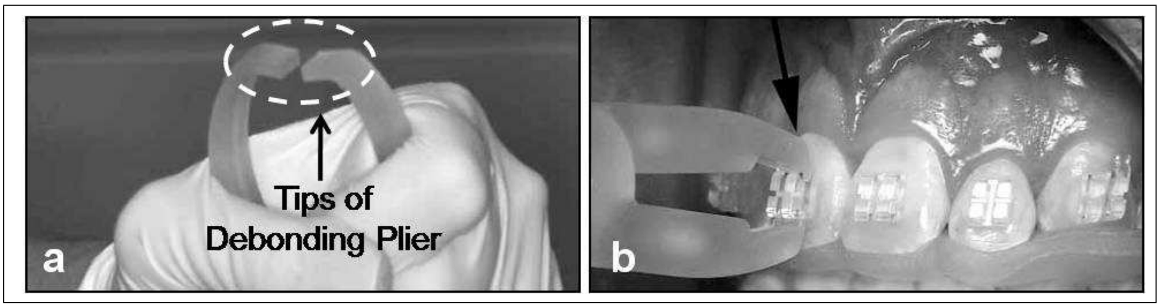


Figure 2.13 (a) The plier handles are squeezed together firmly until they touch, (b) The tips of the debonding plier are engaged under the incisal and gingival tie-wings of the bracket .

The methods used for debonding have advantages and disadvantages (Figure 2.14). All these techniques have their own advantages and limitations. The use of lasers in debonding is a new and established method.

A study for comparing debonding forces belongs to Thomas and Prassana [46] who studied the effects of debonding metal and ceramic brackets on enamel by conventional methods. Four groups of brackets were used. First group were standard metal brackets and the other three groups were different types of ceramic brackets. Enamel damage was seen significantly more in the groups with ceramic brackets than debonding metal brackets. Also they mentioned that ceramic brackets using mechanical retention appear to cause enamel damage less often those using chemical retention.

In 1990, Eliot Storm [1] compared two different etching times (15 seconds and 60 seconds) and two different resins Bis-GMA type resin and 4-META MMA type resin in their study to observe fractures on the enamel. No significant difference was reported between 15 seconds and 60 seconds etched groups. Each resulted in one enamel fracture. Each bonding system also produced one enamel fracture, but heavily filled 4-META MMA type resin was difficult to debond. Using the ceramic debonding instrument was advised in that study specially for ceramic brackets.

Debonding technique	Advantages	Disadvantages
Debonding lift-off pliers	Quick and simple, Standard orthodontic instrument.	Increased debonding force, Risk of enamel damage, Risk of bracket fracture and aspiration of fragments.
Hows or Weingart pliers and ligature cutters	Quick and simple, Standard orthodontic instrument.	Risk of bracket fracture, Risk of aspiration of brackets or bracket fragments, Risk of enamel damage.
Special debonding pliers	Quick and simple, More consistent debonding, Reduced risk, compared to conventional ones.	Additional expense of separate instruments, Brackets and enamel damage still possible.
Electrothermal Debonding	Reduced debonding force, Reduced risk of enamel damage, Reduced incidence bracket fracture, Reduced patient discomfort.	Risk of pulpal damage, Risk of soft tissue burns, Expense of unit, Increased clinical time, Bracket failure may not occur at first attempt.
Ultrasonic	Reduced debond force, Decreased chance of enamel damage, Reduced incidence bracket fracture, Removal of residual resin with same instrument.	Time consuming, Excessive wear of ultrasonic tips, Water spray coolant required to minimise the detrimental heating effect on the pulp.
Debonding agents	Reduced debonding force, Promote failure at the adhesive/enamel interface, Reduced enamel damage.	Questionable effect on the bond strength of the adhesive resins, Increased clinical time, Additional expense of agent.
Laser	Reduced debonding force, Reduced risk of enamel damage, Reduced incidence bracket fracture, Potentially less traumatic and painful.	Potential pulpal damage due to heat production, Expensive units, Laser hazards.

Figure 2.14 Various debonding techniques for orthodontic brackets have been documented and grouped due to their using methods, advantages and disadvantage [3].

2.5 Lasers in Dentistry

Laser is an acronym for "Light Amplification by the Stimulated Emission of Radiation." A laser creates and amplifies a narrow, intense beam of coherent light. The discovery of optic laser technology began with the invention of ruby lasers in the early 1960s [?]. Starting from late 1960s, lasers were introduced to many medical areas. First studies were published in ophthalmology and dentistry. Laser systems can be classified due to their wavelength, active material used, power or mode of operation (Figure 2.15). Wavelength is one of the most important laser parameters that determine how deep laser light penetrates into the tissue.

Parameter	Wavelength	Active Material Used	Power	Mode of Operation
Classification	UV Visible IR	Gas Solid Liquid Electronic	Low Power High Power	Continous Wave Pulse Mode

Figure 2.15 Classifcation of laser systems due to their wavelength, active material used, power or mode of operation.

When matter is exposed to light basically four phenomenas occur: Reflection, Refraction, Absorption and Scattering (Figure 2.16). They are as the same as the phenomenas occur when medical lasers interact with tissue.

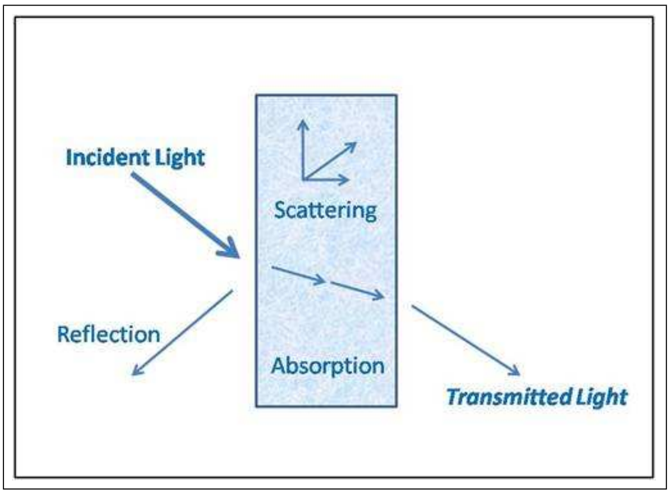


Figure 2.16 Geometry of reflection, refraction, absorption and scattering.

If the incident light is reflected from or transmitted through tissue, it will not cause thermal effect. However, if light is absorbed by tissue, it will be converted into heat. In biological tissues, water molecules or macromolecules such as proteins and pigments are the absorbing agents (Figure 2.17). Absorption coefficient is the term for describing the effectiveness of absorption. The important thing is that nearly 75 % of a tissue tissue content is water. In the ultraviolet, the absorption of light by water is inversely proportional to the wavelength and increasing in the absorption is observed with shorter wavelength due to protein, DNA and other molecules. In the infrared region, the absorption increases with longer wavelengths according to tissue water content. In the red to near-infrared (NIR), absorption of light by water molecules reaches its minimal value. This region is also called the diagnostic and therapeutic window [?]. The way in which the light interacts with the tissue depends on its wavelength.

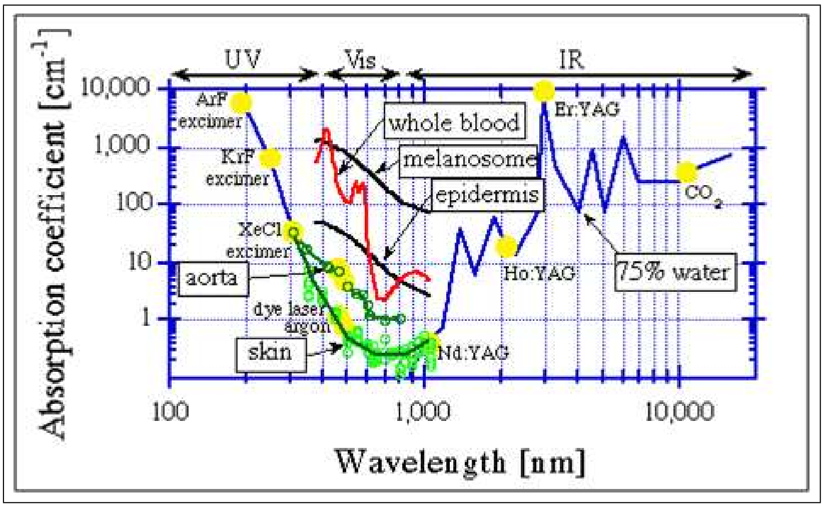


Figure 2.17 The graph above shows the primary absorption spectra of biological tissues. Also shown are the absorption coefficients at some typical laser wavelengths.

The optical properties of the tissues are influenced by the optical properties of its component substances and the concentration and distribution of those substances within the tissue. The composition of tooth structure is not homogeneous. The amounts of both organic and inorganic components present in dentin differ from the amounts of these components present in enamel so the absorption coefficient for each layer differs from the other (Figure 2.18).

During 1980s and early 1990s the use of lasers was introduced into dentistry

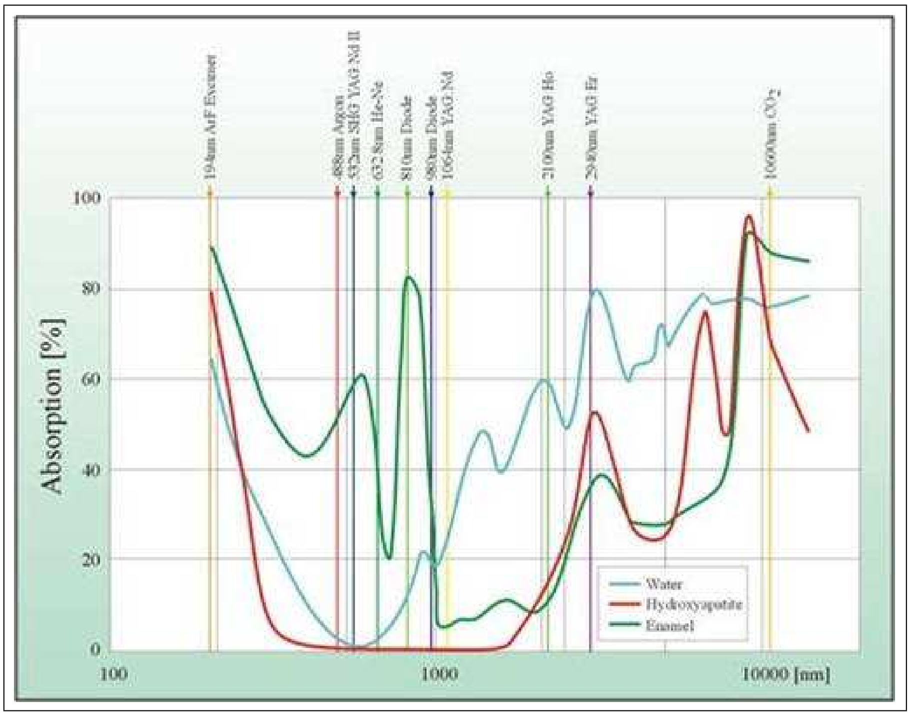


Figure 2.18 Absorption Spectrum of Hydroxyapatite and Enamel.

as various types were approved by the United States Food and Drug Administration (FDA). Dental lasers have been using in experimental and clinical studies like:

1. Cavity preparation and disinfection,
2. Removing of the fillings,
3. Surface etching for micro-retantion,
4. Removing of hard dental tissues and calculus,
5. Removing of soft dental tissues,
6. Lythotrypsy,
7. Root disinfection and cleaning in endodontics,
8. Tooth Bleaching,
9. Laser Welding of Dental Bridges and Dentures,
10. Caries management,

11. Debonding ceramic brackets.

Here are some of the major benefits associated with laser dentistry:

1. Procedures performed using dental lasers may not require sutures,
2. Laser dentistry minimizes bleeding because the high-energy light beam aids in the clotting (coagulation) of exposed blood vessels, thus inhibiting blood loss,
3. Damage to surrounding tissue is minimized,
4. Certain laser dentistry procedures do not require anesthesia,
5. Bacterial infections are minimized because the high-energy beam sterilizes the area being worked on,
6. Wounds heal faster and tissues can be regenerated.

A 1070-nm Ytterbium fiber Laser (IPG Laser, YLM-20-SC, GmbH) is used in this study. A fiber laser or fibre laser is a laser in which the active gain medium is an optical fiber doped with rare-earth elements such as erbium, ytterbium, neodymium, dysprosium, praseodymium, and thulium [19]. Applications of fiber lasers include material processing, telecommunications, spectroscopy, and medicine. The advantages of fiber lasers over other types include:

- Light is already coupled into a flexible fiber: The fact that the light is already in a fiber allows it to be easily delivered to a movable focusing element. This is important for laser cutting, welding, and folding of metals and polymers.
- High output power: Fiber lasers can have active regions several kilometers long, and so can provide very high optical gain. They can support kilowatt levels of continuous output power because of the fiber's high surface area to volume ratio, which allows efficient cooling.

- High optical quality: The fiber's waveguiding properties reduce or eliminate thermal distortion of the optical path, typically producing a diffraction-limited, high-quality optical beam.
- Compact size: Fiber lasers are compact compared to rod or gas lasers of comparable power, because the fiber can be bent and coiled to save space.
- Reliability: Fiber lasers exhibit high vibrational stability, extended lifetime, and maintenance-free turnkey operation.

However, besides lots of advantages, thermal effect during the laser radiation on dental tissues can cause undesirable results. There are lots of studies about thermal effects of laser on dental tissues. Zach and Cohen [5] while applying external heat on teeth of monkeys confirmed no histological changes were discernible with an intrapulpal temperature increase of 1.8 °C. Also with an increase of nearly 5.5 °C in pulpal temperature they mentioned that pulpal necrosis had occurred 15 % of teeth. According to the results of this study, 5.5 °C was accepted as threshold temperature of pulpal damage in most studies. They have also found 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.

Fraunhofer et al. [47] measured the heat produced at the dentinal pulpal wall during etching of dental enamel with an Nd:YAG laser in preparation for direct bonding of orthodontic brackets. An Nd:YAG laser at the power of 0.80 W, 1W, 2W and 3W was used for 12 seconds of irradiation. The thermocouple, against the dentinal pulpal wall used to measure temperature changes on both buccal and lingual surfaces of the teeth. The recorded temperature measurements showed that an increase in power output from the laser unit caused an increase in heat. Power outputs of 1-3 W showed possible irreversible pulpal damage in this study. And also the equality in temperature rise for the buccal and lingual surfaces was reported.

Sulieman et al. [48] measured the surface and pulp chamber temperature increases in vitro on upper and lower anterior teeth during a tooth whitening procedure using a diode laser. They used a thermocouple to measure the temperature increase on the surface and in the pulp chamber. The diode laser was set at three different power settings: 1W, 2W and 3W. They saw that the increase in surface temperature readings ranged from 37 °C (1W) to 86.3 °C (3W) with no bleaching gel present and pulp chamber temperature increases ranged from 4.3 °C (1W) to 16 °C (3W). Then they also examined presence of the bleaching gel reduced temperature increases at the tooth surface and within the pulp. They reported the increase in the pulp chamber temperature with the laser used at 1-2W was below the critical temperature 5.5 °C but a power of 3W produced a pulp chamber temperature increase above the threshold of 16 °C.

Walsh et al. [49] examined intrapulpal thermal changes in human molar teeth by irradiating with a CO_2 at 2W, using a pulsed mode of operation. Two types of duty cycles were chosen for this study. The 5 % duty cycle comprises ten 5 millisecond exposures per second, with 950 milliseconds per second for cooling (95 milliseconds after each laser pulse). In contrast, the 1 % duty cycle setting comprises two 5 millisecond exposures per second, allowing 990 milliseconds per second for cooling (495 seconds after each laser pulse). And 5, 10, 30 and 60 seconds were selected for lasing times of groups. Results of measurements either lasing of crown or irradiation of root showed that temperature rises for groups with lasing times of 5, 10 and 30 seconds were below accepted pulpal injury threshold in that study. However, group with 60 seconds irradiation time was over 2.2 °C. According to those results Walsh mentioned that there was a linear relationship between laser exposure and pulp temperature rise for each of the laser parameters investigated in their study. The times taken to reach maximum temperature were related to the duration of laser exposure. Cooling times were related to the duration of laser exposure, with longer lasing periods associated with longer cooling times. Walsh confirmed that times to maximum temperature may be even longer in the mouth than those recorded in this study, due to cooling from pulpal blood flow and saliva and these same two factors would, similarly, be expected to reduce cooling times. Temperature rises for the 1 % duty cycle were less than those for the 5 % duty cycle. In conclusion, according to the results of that study, Walsh recommended pulsed lasing modes for dental hard-tissue procedures because of the opportunity for cooling between exposure pulses.

Srimaneepong et al. [50] investigated pulpal space pressure and temperature after application of Nd:YAG laser at 1 and 3 Watts of power, and high-speed diamond bur to remove dentin surface. Results were based on the monitoring for pressure and measuring for temperature changes with a pressure transducer and thermocouple. It was observed that laser irradiation and the use of a high-speed diamond bur caused an increase in pulpal space pressure and temperature. Laser irradiation of 3 watts (1.75 kPa and 1.31 °C) caused greater changes than 1 watt (0.53 kPa, 0.34 °C). Also, both pulpal space pressure and temperature increased as remaining dentin thickness decreased.

Selecting the appropriate laser, resin and bracket combination can minimize risks and make the debonding more efficient.

2.6 Lasers in Debonding Ceramic Orthodontic Brackets

Since the early 1990s, lasers have been used experimentally for debonding ceramic brackets. Various lasers have been used in ceramic bracket removal like CO_2 (10600 nm), Nd:YAG (1060 nm), KrF (248 nm), XeCl (308 nm) Tm:YAP (1980 nm) and GaAlAs (808 nm) (Figure 2.19).

Laser energy can degrade the adhesive resin by 3 methods: Thermal softening, thermal ablation and photoablation [51]. Most accepted mechanism is the softening of the bracket adhesive. For an example it is mentioned that at the CO_2 laser wavelength bracket material is highly absorptive. The energy is absorbed and converted into heat in a very thin surface layer of bracket. The energy that absorbed softens the composite at the opposite side of the bracket. The high energy density from the laser on the bracket and adhesive can have a resultant deleterious thermal effect on the pulp of the tooth which may lead to pulpal death [51, 52].

Laser Debonding Orthodontic Ceramic Brackets		
Year	**Research**	**Type of the Laser**
1992	**Strobl *et al.***	**CO_2(10600 nm) Nd : YAG (1064 nm)**
1993	**Tocchio *et al.***	**KrF (248 nm) XeCl (308 nm) Nd : YAG (1064 nm)**
1995	**Obata *et al.***	**CO_2 (10600 nm)**
1995	**Mimura *et al.***	**CO_2 (10600 nm)**
1996	**Rickabaugh *et al.***	**CO_2 (10600 nm)**
1997	**Ma *et al.***	**CO_2 (10600 nm)**
1999	**Abdul –Kader and Ibrahim.**	**CO_2 (10600 nm) Nd : YAG (1064 nm)**
2005	**Kotaro Hayakawa *et al.***	**Nd : YAG (1064 nm)**
2008	**Xianglong et *al.***	**Nd: YAG (1064 nm)**
2008	**Dostálováa *et al.***	**Tm: YAP (1980 nm) GaAlAs (808 nm)**
2008	**Dostálováa *et al.***	**Tm: YAP(1980 nm) GaAlAs(808 nm) Nd: YAG (1064 nm)**

Figure 2.19 Studies on laser debonding orthodontic ceramic brackets.

Debonding ceramic brackets by lasers reduces the time spend during applications, and increases deboding forces as a result of this risk of enamel damage and bracket fracture is reduced.

In 1992 Strobl et al. [51] studied on the reducing of the required debonding forces by using CO_2 and Nd:YAG lasers on monocrystalline and polycrystalline ceramic brackets while debonding from the enamel surface. They used a motorized translation stage to break the bond between ceramic bracket and the tooth surface with a speed of 1 mm/sec. In lased groups, debonding was initiated by applying a torquing force after 2 seconds of irradiation with both lasers. It was confirmed that both brackets were very absorbvative because of the wavelength in lasing with CO_2 so energy was converted into heat and this softened the composite. On the bases of previous preliminary studies a laser energy power level of 14.1 Watts application for 2 seconds was selected for additional debonding experiments. While lasing of CO_2 laser with a power of 14 W for 2 seconds, the average torque force needed to remove polycrystalline brackets was decreased and no bracket failure was reported. While debonding monocrystalline ceramic brackets by lasing at a power of 7 W averages torque was decreased when compared with non-lased group but monocrystalline brackets required lower laser energy for debonding than polycrystalline ones in this study. They explained the causes of different behaviours as a result of differences in the design of two brackets. According to the results of this study, CO_2 laser aided bracket debonding techniques resulted in significantly lower residual debonding force when compared with non-lased debonded group. During the study with Nd:YAG, laser aided bracket removal with a 5 second exposure with a total energy of 120 joules did not produce any significant additional reduction in the debonding force. No enamel damage in the enamel surface or missing enamel fragments were observed during the debonding of brackets of either type. The results of their study have shown that laser-aided debonding significantly reduced the debonding force and the risk of enamel damage (due to ARI scores). However, the thermal effects on the pulp were not investigated.

In 1993 Tocchio et al. [51] used 248 nm, 308 nm and 1060 nm of radiation while debonding the bonded monocrystalline and polycrystalline types of ceramic brackets from the labial surfaces of bovine incisors. The laser beam was centered on the anterior surface of the bracket and was directed perpendicularly to the bonding interface. The laser was closed when debonding occurred. While irradiating on their labial surfaces at densities between 3 and 33 W/cm^2 at three wavelengths, an externally stress of either 0 or 0.80 MPa was applied, too. A laser power of 32.6 W/cm^2 at the bracket surface was used to debond all the polycrystalline brackets and 60 % of monocrystalline brackets. Debonding times were also measured during these experiments. It was mentioned that the debonding times of both types were significantly different from each other. Longer debonding times found in that study when debonding polycrystalline brackets with 1060 nm. The debonding times of poly crystalline brackets at different wavelengths were different.The mean debonding time for polycrystalline brackets was 3.1 seconds during 248-nm lasing, 4.8 seconds during 308-nm of irradiation and was 23.7 seconds as lasing with 1060-nm of laser. Debonding types of monocrystalline brackets debonded with an applied load were always less than one second at all wavelengths. No change in debonding time observed in as the power of 248-nm of laser decreased. For 308-nm radiation, increase in time observed at power of 9W/cm^2 or less. And for 1060-nm of radiation also increases in time seen as the power decreased. It was reported that all the polycrystalline brackets were debonded by sliding down off the tooth under the effect of applied load whereas all the mono crystalline samples with 0 or 0.80 MPa load were debonded by bracket blow off (except one during 308-nm lasing). When the surface of each enamel and bracket were observed with both light and scanning electron microscopy to determine the extent of bracket and enamel damage, no enamel or bracket damage observed. No intrapulpal temperature change was measured in those experiments.

In 1995, Obata et al. [52] studied on the effects of CO_2 laser on ceramic bracket removal from the tooth surface for two different bonding adhesives: 4-META MMA and Bis-GMA for determination of optimal laser energy power levels and exposure time for both adhesives. From prelimineray studies for the 4-META MMA resin group, application of 3 watts for 3 seconds had been found suitable. Thermal expansion properties of resins were examined. It was informed that ceramic brackets and Bis-GMA resin showed linear expansion properties as the temperature increased. Because of the differences in the thermal expansion between MMA resin and the ceramic bracket, at 60 °C the resin shifted from expansion to contraction. As a result of this, Obata suggested that both thermal and resin contraction from ceramic brackets are responsible for debonding mechanism like Mimura [53]. Also, temperatures were measured during laser irradiation for safety. Pulp cavity increasing temperatures were reported more than bracket surface increasing temperatures. Furthermore, no histological difference was seen between irradiated tooth pulps and non-irradiated tooth pulps. In the control group on the tooth surfaces after the shearing force measurements showed enamel fracture and brackets cracked. The laser irradiated group that included the use of either 4-META MMA or Bis-GMA had no fractures. Findings in that study confirm 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.

In 1996 Rickabaugh et al. [53] studied to investigate the in vitro use of the laser debonding pliers with a carbon dioxide laser at varying tensile force levels in the removal of ceramic brackets. They used a carbon dioxide laser to debond poly crystalline ceramic brackets which were bonded with a MMA type of adhesive resin. They measured all lasing time and intrapulpal temperature increases and compared, accepting 5.5 °C as safety threshold in the intapulpal temperature. Laser was always at a fixed distance, perpendicular to the bracket. An instron machine set at a speed of 1 inch per minute was used for debonding. There was a control group with debonding and with no lasing in a tensile mode. The mean debonding force for the control group was 13.04 pounds (4.88 Mpa). Broken tie wings were noted in 3 of the 10 samples at debonding. During lasing by a modified debonding plier, a tensile debonding force was applied. Those force levels were selected according to the debonding forceapplied to control group. The three experiment groups were debonded with the CO_2 laser at 20 W and a static tensile force of 3 pounds, 1.5 pounds and 0.75 pounds. The length of lasing time and the increase in intra pulpal temperature were measured. The increase in intrapulpal temperature was recorded by a thermocouple. After test, all brackets were inspected for tie wing breakage. In group with 3 pounds of tensile force, the mean debonding time for this group was 1.64 seconds. And the mean increase in intrapulpal temperature was 1.80° C. Only one sample reported that exceed the 5.5 °C safety threshold and this same sample had the only broken tie wing in this group. For group with 1.5 pounds, the mean debonding time was 1.83 seconds and the mean increase in intrapulpal temperature was 3.01° C. Just one sample had a temperature that exceeds the 5.5 °C safety threshold. No broken tie wings reported. Group with a static tensile force of 0.75 pounds, had a mean debonding time of 3.42 seconds and a mean of 4.47° C temperature. However, 3 samples were more than 5.5 °C safety threshold. Again, no broken tie wings reported. 3.0 and 1.5 pound force applied groups required significantly (used post hoc comparisons of Scheffe) less time than 0.75 pound group. There was a good relationship confirmed between time and temperature for the 1.5 pound group and 0.75 pound groups, but not with the 3.0 group. As a summary 1.5 pound (6.67 N) force found with the best combination in terms of debonding time and increase in intrapulpal temperature. The results showed that this study agree with those of Strobl et al. and Tocchio et al. in the laser effects thermally soften the adhesive to remove the ceramic brackets easily.

Examined Variable	3.0 joule Group	1.5 joule Group	0.75 joule Group
Mean debonding time (seconds)	1.64	1.83	3.42
Mean increase in temperature (° C)	1.80	3.01	4.47
Number of broken tie wings	1	No	No

Figure 2.20 Examinated values for mean debonding time, mean increase in temperature and number of broken tie wings during the experinments in the study of Rickabaugh et al. [4].

In 1997 according to previous studies in laser debonding of ceramic brackets, Tsun Ma et al. [4] tried to find a method to reduce the fracture of ceramic orthodontics brackets during debonding procedures which is an important disadvantage of debonding. The aim of their study was to determine the amount of lasing time required to achieve a significant reduction of debonding while keeping the intrapulpal temperature increase within a safety limit. Ceramic brackets were bonded to mandibular bovine teeth and human mandibular first premolars with a photoactivated bonding resin. 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, without the application of laser energy. The tensile force required to debond the bracket was measured for each sample. The mean and standard deviations of the control group were 15.31 pounds (68.1 N). By using this predetermined tensile force based on the data from the control group, a tensile debonding load of 4 pounds which was 25 % of control group, was chosen for the experimental group. A power level of 18 watts was used with the CO_2 laser in the continuous power output mode used in laser debonding group. Modified debonding pliers was used to accurately position the laser beam onto the ceramic bracket. The laser was activated for the specific period of time.

The intrapulpal temperature was recorded with a thermocouple. Lasing time required to keep the maximum intrapulpal temperature rise below 2 °C was chosen as a factor of safety. Visual examination of the debonded specimens revealed adhesive failure at the resin/bracket interface. There was no incidence of bond failure at the resin/enamel interface. In the control group two specimen had tie-wing fractures and three specimen had mounting block fractures. 20 % of the samples did not debond with the 4-pound tensile load, which is equivalent to a tensile stress of 216.2 psi, 15.19 Kg/cm^2, or 1.48 Mpa. The remaining 80 % of specimens were debonded successfully without any incidence of bracket fractures or enamel damage. Also, Ma et al. showed that there is a linear relationship between lasing time and an increase in intrapulpal temperature while lasing polycrystalline brackets Ma also reported that the bovine teeth indicated a greater increase in intrapulpal temperature for the same lasing time as the human premolars. A mean intrapupal temperature increase of 0.91 °C was observed after 1 second lasing, 1.74 °C after 2 seconds and 2.67 °C after 3 seconds. In experiments with bovine teeth a mean intrapupal temperature increase of 1.65 °C was observed after 1 second lasing, 3.31° C after 2 seconds and 5.15 °C after 3 seconds. As a summary of those results, this study confirmed that thermally softening of the orthodontic adhesive by the carbon dioxide laser reduced the tensile debonding force with minimal increase in intrapulpal temperature. The mean intrapulpal temperature rise was below 2 °C. Visual examination of debonded specimens revealed bond failure occurred only at the bracket/resin interface and the incidence of ceramic bracket breakage in the control group was 20 %.

In 1999 Abdul Kader and Ibrahim [54] debonded ceramic brackets using a CO_2 laser at a power of 50 W. They decided a time interval of 2 seconds as exposure time. The temperature of bracket rose up to 93.63 °C although the temperature in the pulp chamber did not exceed 0.7 °C. Also in another study of Abdul Kader and Ibrahim, they reported that whatever the exposure time of laser was, significantly higher force was required for debonding ceramic brackets when one minute elapsed after laser exposure compared with debonding immediately after laser exposure. (It meansless debonding force was required before the adhesive resin material resolidifaction) [55].

In 2005, Hayakawa et al. [56] studied to develop an effective method for debonding ceramic orthodontic brackets with a high-peak power Nd:YAG laser. Single crystal and polycrystalline ceramic brackets were bonded to mandibular bovine teeth with 4-META MMA and Bis-GMA types of bonding resins. Hayakawa used Nd:YAG laser at 3 different energy levels 1.0, 2.0 and 3.0 joules. Bond strength and thermal effects of the laser on the dentin surface were assessed at these 3 laser energy levels. Also, there was a non irradiated control group. Laser was applied a pulse duration of 1.2 ms and 1 pulse per location. The bond strength of each specimen was measured with a testing machine which the crosshead speed was 1 mm/minute. The thermal effects of lasing were indirectly measured through the change in pulp wall temperature. Bond strength was decreased in the 2.0 J and 3.0 J groups when compared with the non irradiated group. However, no difference was reported with 1.0 J group. No significant differences were observed among the brackets of used in 3.0 J group although in 2.0 J group, the single-crystal bracket group showed a significant decrease in bond strength compared with the poly crystalline group. All brackets bonded with Bis-GMA showed almost no reaction when debonded with 1.0 J laser energy. No significant differences were found among the both adhesive resins in the 2.0 J and 3.0 J groups. Thermal effects were also easily observed on the resin, but they were rather shallow; carbonization-like effects con- firmed only on the resin surface. The maximum temperature rise measured on the pulpal walls at the lasing points was 5.1 °C.The mean intrapulpal temperature increases of each group can be shown as:

At 2.0 and 3.0 J, the bracket bases exhibited hollows and black deposits, and the remnant resin 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 As a result the application of a high-peak power Nd:YAG laser at 2.0 J or more was confirmed as effective for debonding ceramic brackets.

Xianglong Han [57] studied on the efficiency of Nd:YAG laser-aided ceramic brackets debonding technique with both ceramic brackets and metallic brackets. They used three different debonding techniques were used in that study: Metallic brackets were debonded with shear debonding force (Group 1), ceramic brackets were debonded with shear debonding force (Group 2), and ceramic brackets were debonded with Nd:YAG laser irradiation (Group 3). Nd:YAG laser was applied with a pulse width of 0.2 ms, and 3W for 3 s. Shear bond strengths, ARI scores and scanning electron microscopy examinations were also observed in that study. In Group 1 the mean shear bond strength for debonding brackets was 9.78 Mpa, in Group 2 the mean shear bond strength for debonding brackets was 11.63 Mpa and in Group 3 the mean shear bond strength for debonding brackets was 5.13 MPa. Xianglong Han mentioned no pulpal injury was occurred when Nd:YAG laser with that values was applied . ARI scores showed that application of laser irradiation had the most desired results. Moreover, scanning electron microscopy observations showed that Nd:YAG laser diminished the amount of remnant adhesive without damaging enamel structure.

Dostálová et al. [58] used two types of lasers: A laser diode (808 nm) and a diode-pumped Tm: YAP (1.9 nm) in their study. Laser radiation was applied for 30, 60, and 90 seconds for Tm: YAP irradiation and 60 seconds for GaAlAs diode at a maximum power of 1 W. While irraditon with a GaAlAs diode laser at a power of 1W, for 60 seconds rised the temperature 18 °C. A power of 2W, lasing for 60 seconds caused a 29 °C increase in temperature and a power of 10W, lasing for 60 seconds caused a 114 °C increase in temperature. Irradiation by the GaAlAs diode laser did not show a significant effect on debonding of ceramic orthodontic brackets even with a power of 10 W for 60 seconds. During lasing with Tm: YAP laser the temperature changes of irradiation without and with water cooling of the tooth tissue were done. At 1 W of power without cooling the tooth tissue no bracket removal observed in the group with 30 second of irradiation duration. In 1W of power lasing for 60 second group without water cooling, brackets debonded with a 31 °C of increase in temperature. There was water cooling of the tooth tissue in all other groups. Group with a power of 1W of laser irradiation for 60 seconds showed an increase of 2 °C. Group with a power of 1W of laser irradiation for 90 seconds showed an increase of 5 °C and group with a power of 2W of laser irradiation for 60 seconds showed an increase of 9 °C. All in these 3 groups bracket removal was observed. From the SEM measurement the minimum damage of enamel was found after Tm:YAP. During the thermal debonding procedure, irradiation by the continuously running Tm:YAP laser gave good results in the brackets debonding.

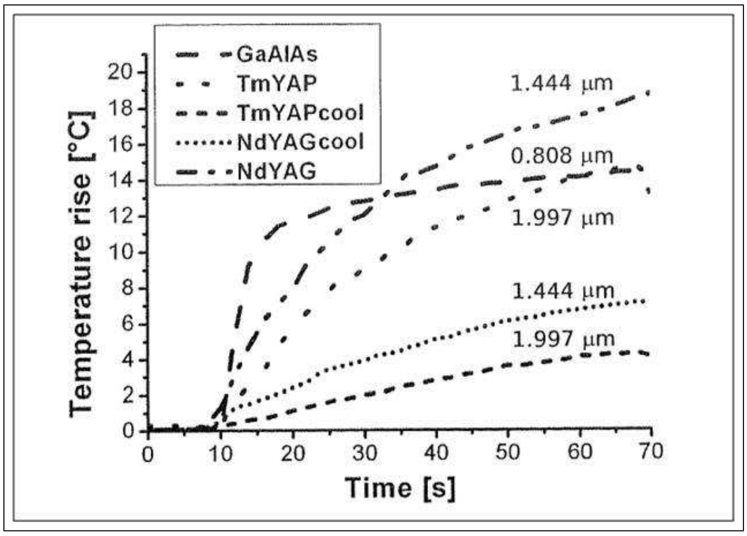


Figure 2.21 Graph of time dependence of the temperature rise during irradiation by various types of radiations which was reported in the study of Dostálová et al.

In another study of Dostálová et al. [59] in 2009, the laser radiations of a diode-pumped Tm:YAP microchip laser (Operating at 1980 nm with a maximum output power of 3.8 W), diode pumped Nd:YAG laser (Operating at 1444-nm with a maximum output power of 2 W) and a GaAlAs diode laser (Operating at 808 nm with a maximum output power of 20 W) were examined for debonding effect. 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. Aa a result diode-pumped Tm:YAP microchip laser with an output power 1W can be a good candidate for the ceramic bracket debonding procedure. For GaAlAs diode laser no debonding was reported even for 60 s irradiation and heat was increased up to 14 °C. During experimens with Tm:YAP, it was observed that bracket could be removed after 60 s because of the heat was concentrated inside the bracket and adhesive resin. When irradiation time increased up to 90 s or power up to 2W the debracketting speed did not changed. The optimum value reported for Tm:YAP was with a exposure time of 60 s and water cooling application. Similar results was obtained for Nd:YAG laser only the difference between in with and without cooling systems was more. Because of the unefficiency radiation of GaAlAs diode in debonding just other SEM measurements reported that the results for other two lasers were similar. The minimum damage was seen with Tm:YAP laser at 1 W and 60 s interval.

3. MATERIALS AND METHODS

3.1 Teeth

Intact 2-year-old bovine mandibular incisors were used instead of human teeth because of their availability, higher hygiene and near physical properties to human teeth. The use of bovine teeth for resin-to-enamel bonding studies has been validated [51, 4, 54, 57]. Also, previous reports found no clinically significant differences in the debonding forces of brackets attached to incisors and molars [52]. After extraction, the teeth were washed and scaled off calculus, soft tissue debris and blood and then rinsed. The labial surfaces of bovine incisor teeth, which had been were polished with pumice slurry and washed. With reference to previous studies, isotonic sodium-chloride solution was selected for storing at room temperature until bonding [60, 61]. Solution had changed three times a week to avoid the reproduction of bacteria.

3.2 Orthodontic Ceramic Brackets

Polycrystalline ceramic brackets (G&H.US) for maxillary lateral incisors were bonded to bovine incisors in this study (Figure 3.1a-b). Due to the previous studies showed that debonding of polycrystalline brackets required higher force than mono crystalline ones, polycrystalline brackets were selected because of their availability [84]. Also, with the recommendations according to the results of those studies and generality of usage, chemically curing Bis-GMA resin set (3M, Unite Bonding Adhesive Set,US) was used in our tests (Figure 3.1).

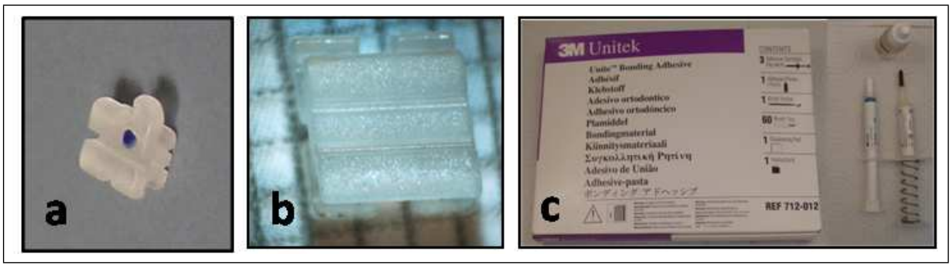


Figure 3.1 (a) Ceramic bracket, (b) The base of ceramic bracket, (c) Bonding Adhesive Set with Etching agent, Adhesive Primer and Adhesive Resin.

3.3 Bonding Procedure

Each bovine crown was embedded in gypsum blocks as the labial surface of the enamel was positioned so that it was as parallel as possible to the vertical axis of the block so that debonding would be in a rigid and pure tensile mode(Figure 3.2). Cavities were opened on the lingual surface of teeth by a round diamond bur with a 1mm diameter (Figure 3.3).

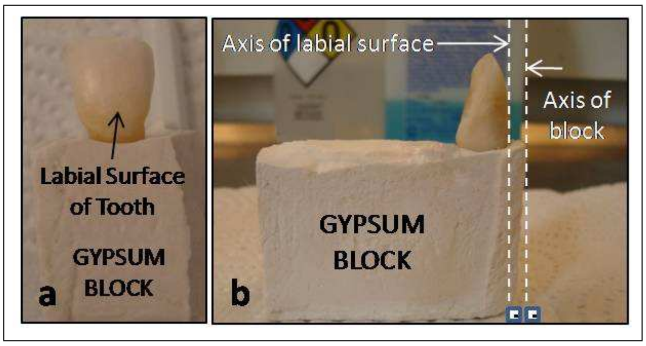


Figure 3.2 (a) Labial view and (b) Lateral view of bovine crown which was embedded in gypsum block.

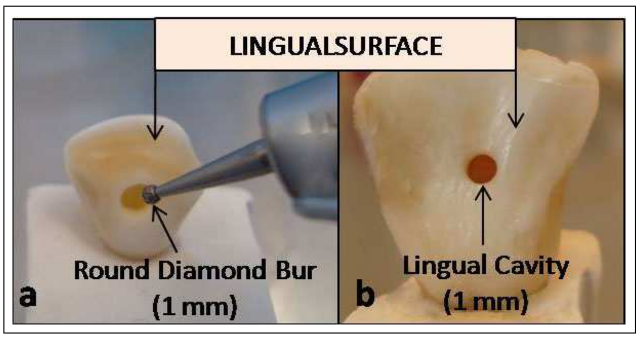


Figure 3.3 (a) Round diamond bur and (b) Lingual cavity were with a 1mm length diameter.

All brackets were bonded with orthodontic adhesive following the manufacturers' recommendations. The steps of bonding procedure were involved in direct bracket bonding on lingual surfaces are as follows: Cleaning, Etching, Sealing and Bonding.

Before the bonding operation, the bonding surfaces of all specimens were cleaned and polished with a pumice paste for 15 seconds, rinsed with water, and dried to remove plaque and the organic pellicle that normally covers all teeth. As previous studies mentioned that there was no significant difference between etching 15 seconds or 30 seconds, each enamel surface was etched for 30 seconds [82] and rinsed with a syringe, and dried. A chalky and frosty white appearance was obtained on the enamel surface. However, it was known that teeth that did not appear dull and frosty white shallow on the enamel should be re-etched; there was no need to repeat this step in our experiments (Figure 3.5).

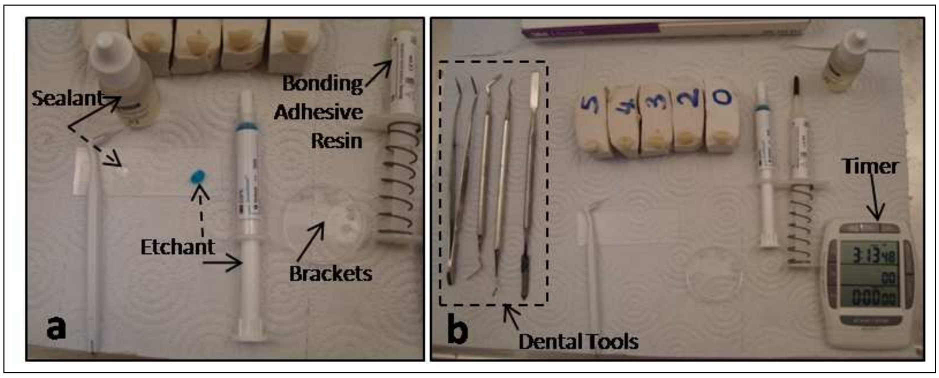


Figure 3.4 (a) Bonding adhesive resin, sealant, etchant and brackets that were used in experiments, (b) Dental Tools that were used during bonding ceramic brackets on the enamel and timer was used to control etching time.

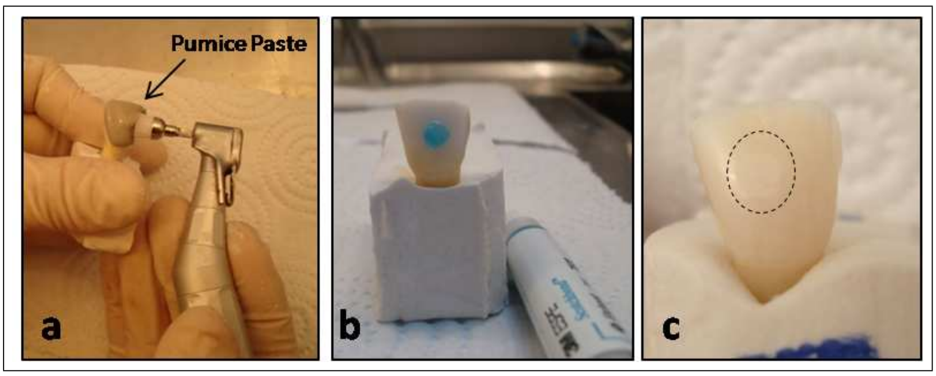


Figure 3.5 (a) Polishing the enamel with pumice paste for 15 seconds, (b) Etching enamel for 15 seconds, (c) Dull and frosty white shallow on the enamel after etching.

After the etching-applied area was completely dry and frosty white, a thin layer of bonding agent (sealant) was painted over this surface and bracket base. Bracket placement started immediately after etched surface coated as recommended [46]. The recommended bracket bonding procedure consists of the following steps: Transfer, Positioning, Fitting and Removal of excess. The bracket was hold by tweezers and then bonding adhesive was applied to the back of the bonding base. The bracket was immediately placed on the tooth close to its correct position. The brackets were positioned mesio-distally and inciso-gingivally accurately relative to the long axis of the teeth. The bonding interface was axially centered and positioned parallel to the gypsum blocks face. For bonding, the bracket was placed at exactly at the opposite side of the cavity, on the labial surface. Next, the bracket was pushed firmly towards to the tooth surface with one-point contact. Excess resin was cleaned from the edge of the bracket before polymerization with a hand scaler. It was not important in experiments but in vital conditions removal of excess adhesive reduces periodontal damage and the possibility of decalcification. Clinically significant gingival hyperplasia and inflammation rapidly occur when excess adhesive comes close to the gingiva and is not removed properly [46].

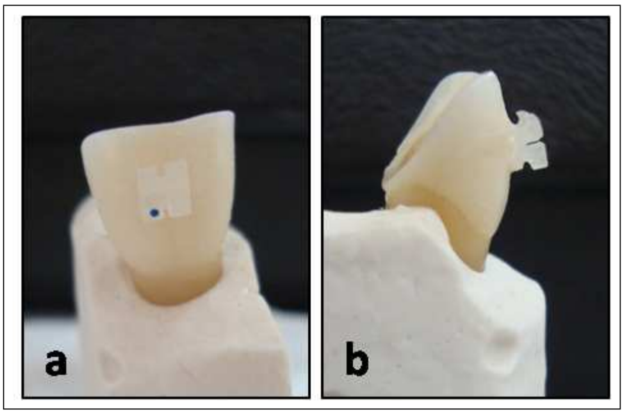


Figure 3.6 Ceramic brackets were bonded on the enamel, (a) Labial view, (b) Lateral View.

After the adhesive resin was cured, the samples were replaced in a beaker with sodium chloride solution inside and stored at 100 % humidity and 37 °C for 2 days to ensure composite polymerization (Figure 3.7).

Figure 3.7 (a) A group of teeth after bonding procedure, (b) Groups, inside a beaker, were kept in an incubator for 2 days.

3.4 Experimental Set-up

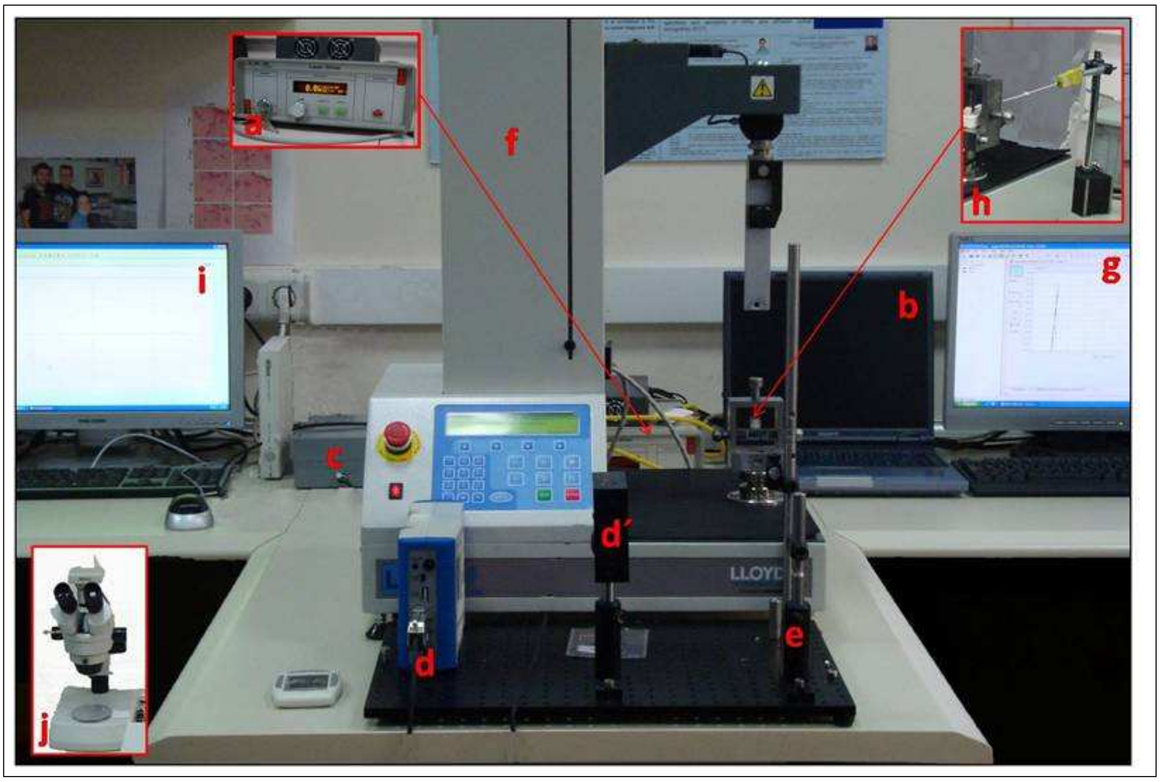


Figure 3.8 (a) 1070- nm Ytterbium fiber Laser (IPG Laser, YLM-20-SC, GmbH), (b) Computer for adjusting the duty cycle of pulses in the experiments in pulse mode, (c) Pulse modulator to control the duty cycle, (d) Monitor and controller of Powermeter (Newport, Model 1918-C), (d') Sensitive measuring head of powermeter, (e)) Holder for laser tip, (f) Universal testing machine (Lloyd, LF Plus, UK), (g) Computer that is connected to universal testing machine for controlling and collecting and recording data, (h) K-type thermocouple (OMEGA, OM-CP-0CTTEMP, UK), (i) Computer for collecting and recording data from the thermocouple, (j) Stereomicroscope.

Experimental set-up consist of three parts:

1. 1070- nm Ytterbium fiber Laser system,
2. Universal testing machine,
3. Temperature measurement system.

A 1070- nm Ytterbium fiber Laser (IPG Laser, YLM-20-SC, GmbH) was a computer and a pulse modulator to control the duty cycle of pulses in the experiments in pulse mode. takes place in this system. Before each test a powermeter (Newport, Model 1918-C) was used to measure the output energy level of the 1070- nm Ytterbium fiber Laser (IPG Laser, YLM-20-SC, GmbH) during irradiation (Figure 3.9). The powermeter consisted of a head for measuring the power and a monitor for controlling and reading the data. There was also a metal holder used for fixing the laser tip.

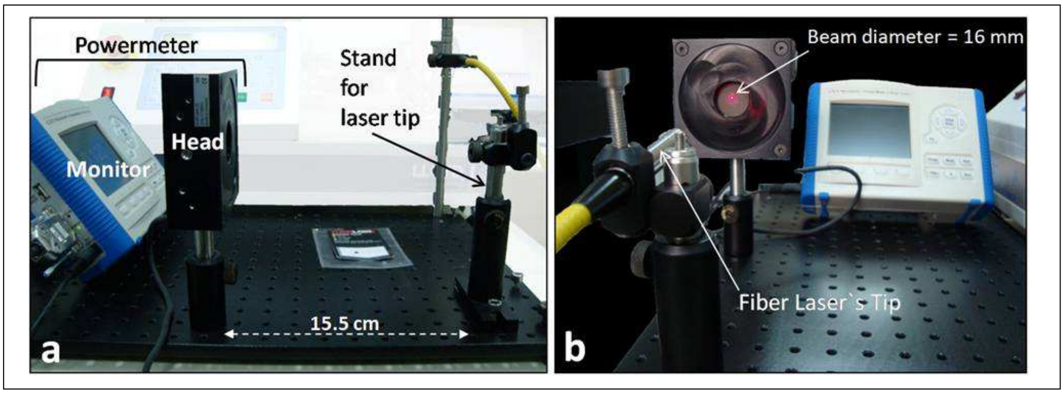


Figure 3.9 (a) Powermeter (Head and monitor parts) and holder for fixing laser tip. Also, the distance between power meter's head and stand was 15.5 cm; (b) Diameter of the laser beam was 16 mm.

In the second part of the experimental set up (Figure 3.10) , a universal testing machine (Lloyd, LF Plus, UK) was used to measure the shear bond strength of each specimen during debonding. It was consisted of an upper moving section and a rigid base section. A steel jaw was prepared especially for the base section as a testing frame. To debond the brackets from teeth, the gypsum blocks were placed in this testing frame. Also a steel shearing blade (with 19, 5 cm length) with a square hole was mounted to the upper part of the machine. In addition to this, a computer connected to the testing machine was used for controlling the machine and reading,collecting and recording the data.

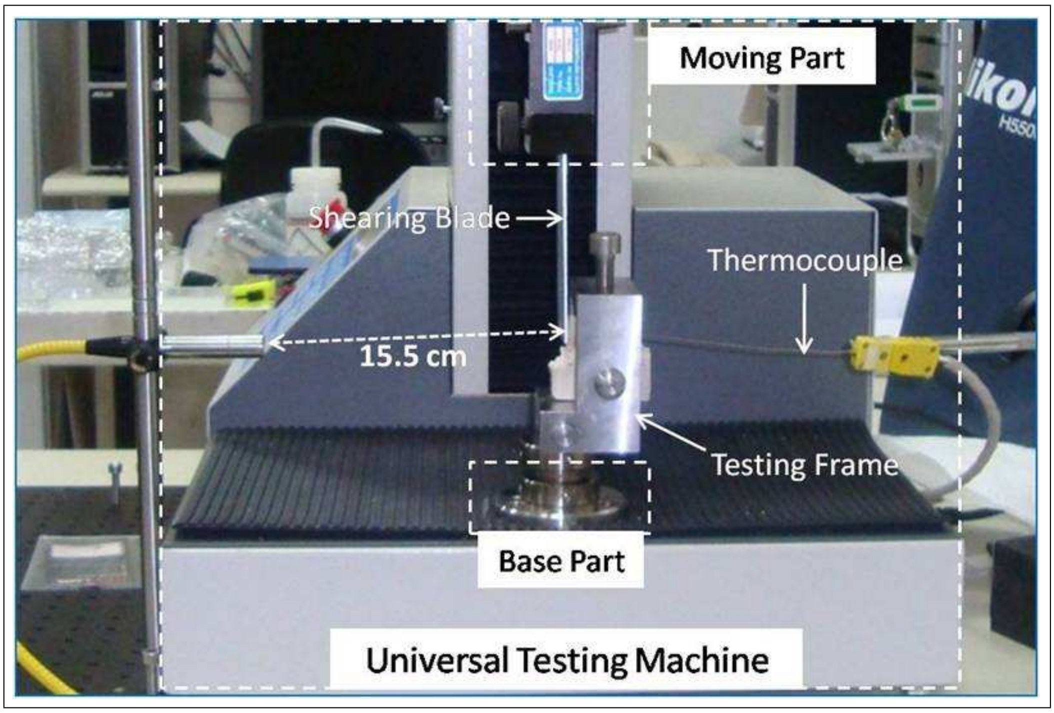


Figure 3.10 Universal testing machine with a base and a moving part, testing frame, K-type thermocouple and shearing blade.

K-type thermocouple (OMEGA, OM-CP-0CTTEMP, UK) was a material that used in temperature measurement system to measure intra pulpal thermal changes during irradiation. There was a computer that read, collect and record the data from and the K-type Thermocouple.

Moreover, stereomicroscope for observing enamel surfaces and base of the brackets after lasing and camera was connected to it for and also the 1070-nm fiber Laser‘s controller was in the last part.

3.5 Experimental Procedure

First of all, the output energy level from the fiber tip was measured by the powermeter at least three times before each sample by fixing the tip on the holder. The external diameter of the fiber waveguide was 1.6 mm. It was controlled that the diameter of the Laser beam was always in the same length while distance was changing.

In this study, experiments were performed in two sections according to the type of lasing mode:

1. Section 1: Adjusted Laser power was applied in Continuous Wave (CW) Mode,
2. Section 2: Adjusted Laser power was applied in Pulse Mode.

Section 1: Experiments in CW Mode			
Group name	Adjusted power (W)	Measured power (W)	Number of samples
Control	No lasing	No lasing	13
2	2	1.41 ± 0.07	13
3	3	2.14 ± 0.07	13
4	4	2.78 ± 0.07	13
5	5	3.50 ± 0.07	13
6	6	4.20 ± 0.07	13

Figure 3.11 Adjusted power, measured power and number of samples for each group in CW mode.

In first section, experiments in continues wave (CW) Mode, laser was applied on samples with an arranged constant power in continuous mode. The output energy levels used in this section were selected after the preliminary study about the intrapulpal thermal changes with the application of this laser. According to those energy levels, samples were divided into 5 different groups (Figure 3.11). Beside those groups that brackets debonded with laser energy, there was also a control group debonded without laser application. 13 specimens from each combination type were randomly assigned to each group.

Section 2: Experiments in Pulse Mode				
Group name	Adjusted current (A)	Measured power (W)	On Time (ms)	Off time (ms)
200/600	4.99	18.0 ± 0.1	200	400
300/900			300	900
400/1200			400	1200

Figure 3.12 On time and off time intervals for lasing, adjusted current and measured power values for each group in pulse mode.

In the second set of experiments in Pulse Mode, laser power was set by adjusting current level and duty cycle. Lasing dose was set as a current value of 4.99 A that had an average measured power of 18.0 ± 0.1 Watts (Figure 3.12). Best fit pulse durations in these experiments were selected by preliminary experiments.

After measuring the adjusted power in the first part, fiber tip was replaced in the second part of the set up where the distance between the fiber tip and the steel jaw was ranged as 15.5 cm again.

After taking the sample out of the incubator, silicon thermal paste (Bakir, R-1260 Silicon Gress, Turkey) was filled manually into the lingual cavity of each tooth in order to mimic the pulpal tissue and to counteract the thermal lost during the procedure. The gypsum block was fixed around by three screws to the testing frame. The shearing blade was placed exactly on the base of the bracket's gingival wing.

The thermocouple was placed into the lingual cavity and ensured that the tip was touching on to the intra-pulpal wall (Figure 3.14). Laser guide beam was positioned to the centre of the ceramic bracket surface which is one of the thinnest parts of it (Figure 3.15). This lasing position was situated so that the laser energy would most

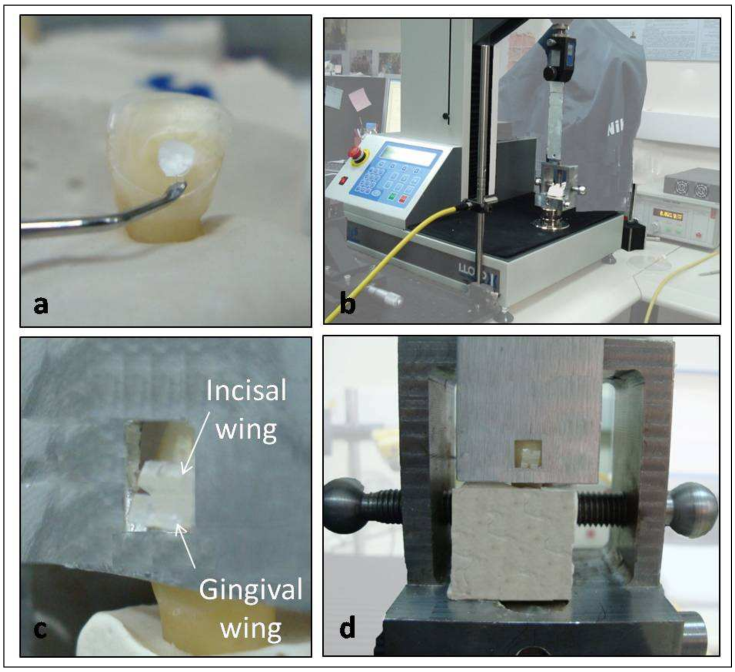


Figure 3.13 (a) Silicon thermal paste was filled into the lingual cavity of each tooth, (b) The gypsum block was fixed around by three screws to the testing frame, (c, d) The shearing blade was placed exactly on the base of the bracket's gingival wing.

effectively travel to the adhesive resin. The laser energy was positioned in order not to be delivered perpendicularly to the labial surface of the bracket, to avoid the reflection of the light back through the fiber.

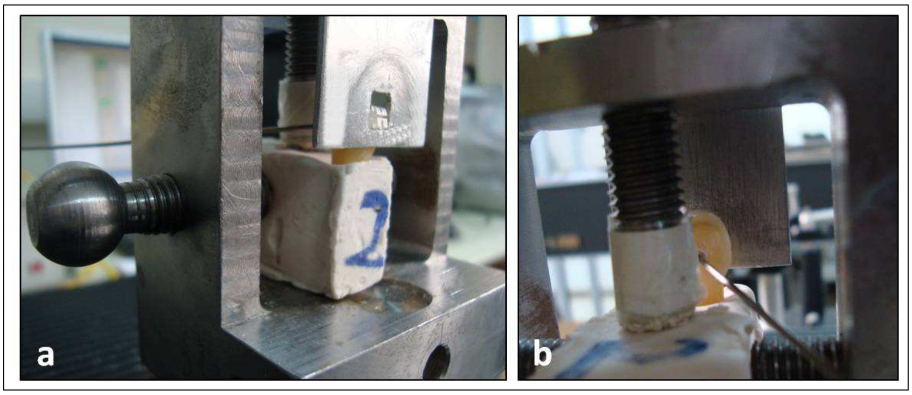


Figure 3.14 (a) Lateral and (b) labial views after the replacement of thermocouple into the lingual cavity.

In experiments in CW Mode, after positioning the guide beam, the output power was set at a desirable value. On the other hand, in experiments in Pulse Mode, after positioning the laser beam, the output current was constantly set at 4.99 Amperes which

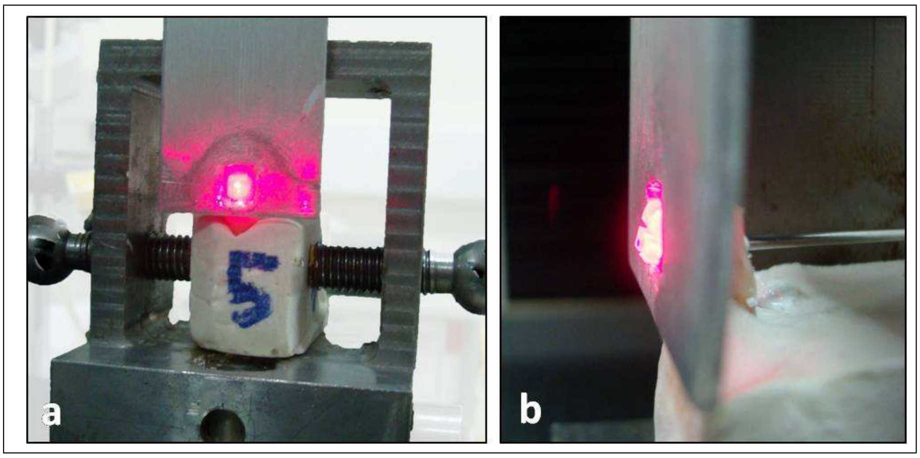


Figure 3.15 (a) Frontal and (b) lateral views after guide beam was positioned to the centre of the ceramic bracket surface.

had a measured output 18.0 ± 0.1 of Watts. The desired on time and off time intervals for pulse cycle for irradiation was set from the computer in part 3 that was controlling the pulse mode of the laser. Then universal testing machine was started to apply a torque force on the ceramic bracket for shear bond strength test. In test, the universal testing machine had a 1 mm/minute crosshead speed. The application of a torque force perpendicular to the bracket enamel interface was chosen as the standard debonding technique throughout this study. The force applied on the ceramic orthodontic bracket during test was called as debonding force. For both experimental groups, one of the computers in part 3 was collecting and recording the data from the universal testing machine which measures the force applied on the bracket while debonding. During shearing test, whenever an increase in debonding force was examined on the monitor of the computer, the selected group of lasing was manually started to apply onto the sample in chosen mode. That moment was defined as the starting point. Lasing and then the shearing test were ended manually at the moment of the detachment of the ceramic bracket from enamel surface. The moment of the detachment was called as the breaking point and the load applied at that moment was defined as breaking load. Time and load at the breaking point were measured and recorded for each sample. Debonding time which was defined as the time interval between the starting point (where the tension starts the commencement of the laser irradiation) and the breaking point. Irradiation and debonding time intervals are equal (Figure 3.16).

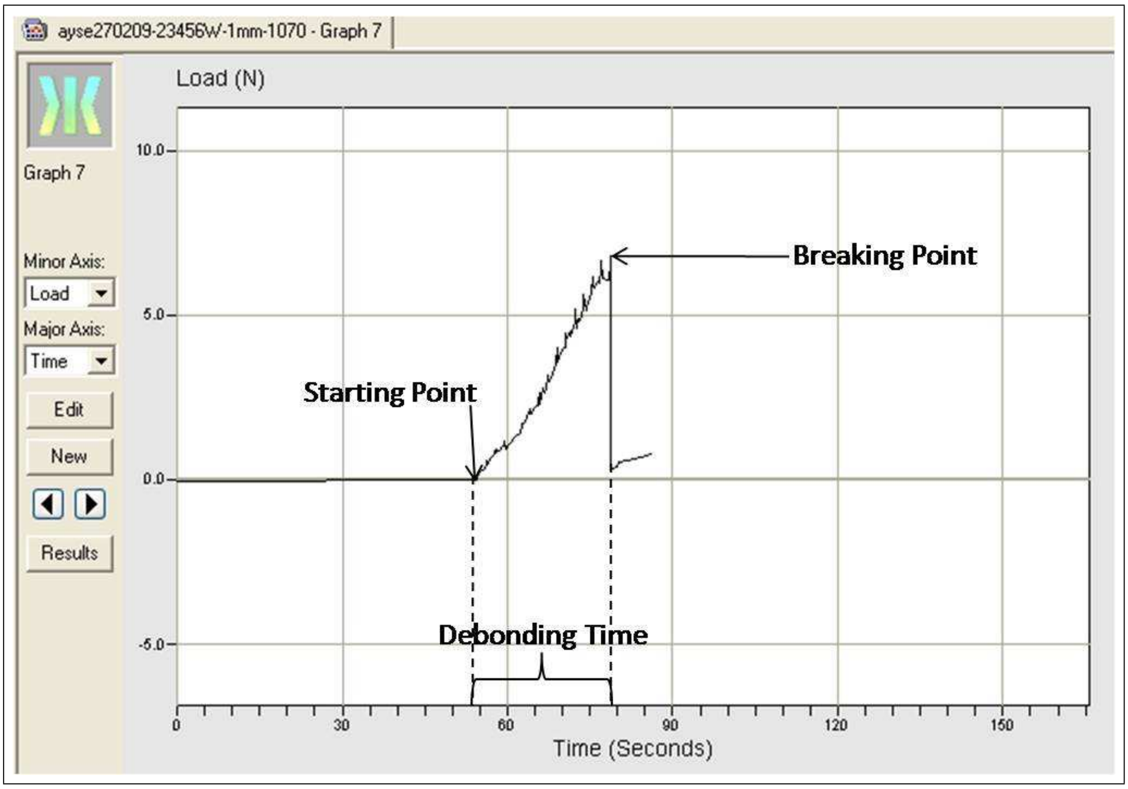


Figure 3.16 Starting point, breaking point and debonding time was shown on the graph that was examined on the monitor of the computer which was connected to the universal testing machine.

The other computer with OMEGA program was continuously collecting, recording and monitoring the data from the K-type thermocouple which measures the intra-pulpal temperature changes during each debonding procedure. The measurement was done for every 2 seconds automatically (Figure 3.17). Exact temperature values on the Excel sheet were tracked in order to figure out debonding time more accurately (Figure 3.18). The equality of irradiation and debonding time intervals helped to find the final temperature at the breaking point. The difference between the final temperature and the temperature at the starting point gave the intra-pulpal temperature change.

The residual load needed to debond the bracket, breaking time of the ceramic orthodontic bracket, the intra pulpal thermal changes while debonding were the evaluated values during experiments. Average values and standard deviations were calculated for shear bond strength, breaking time and intrapulpal temperature change for each group. The work done by universal testing machine and laser energy applied on the ceramic brackets during debonding procedure were calculated. Means and stan-

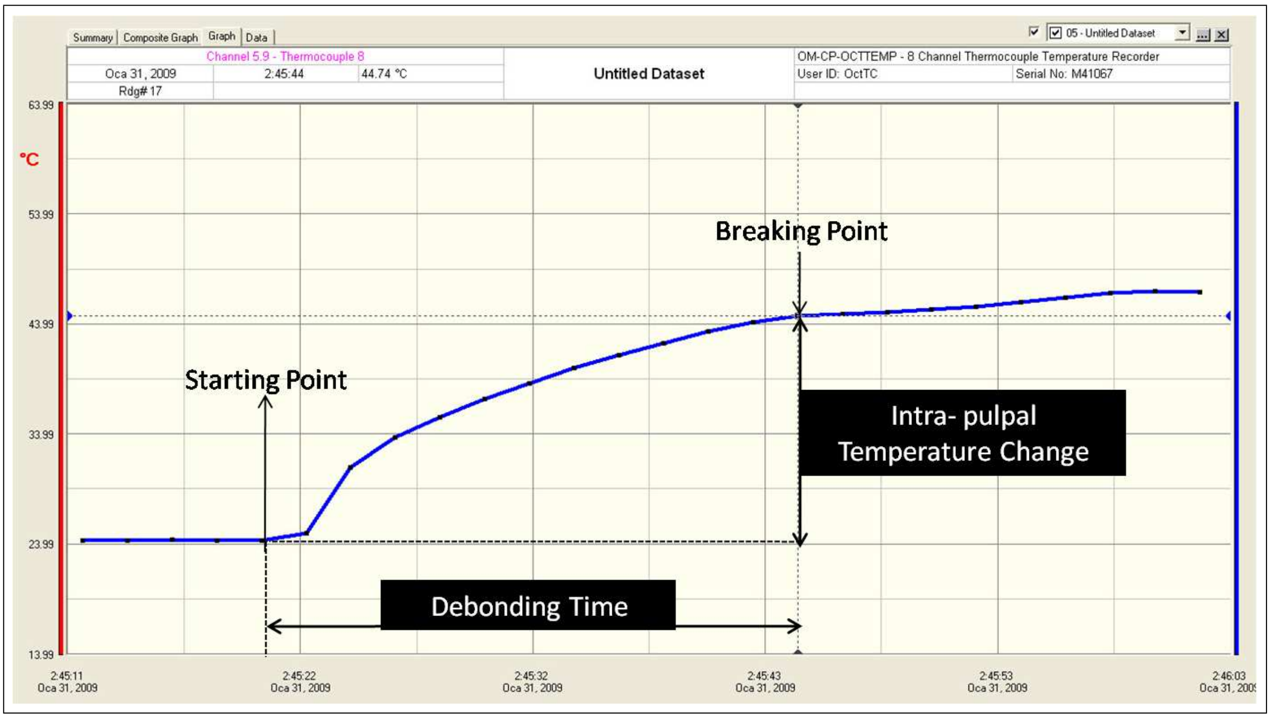
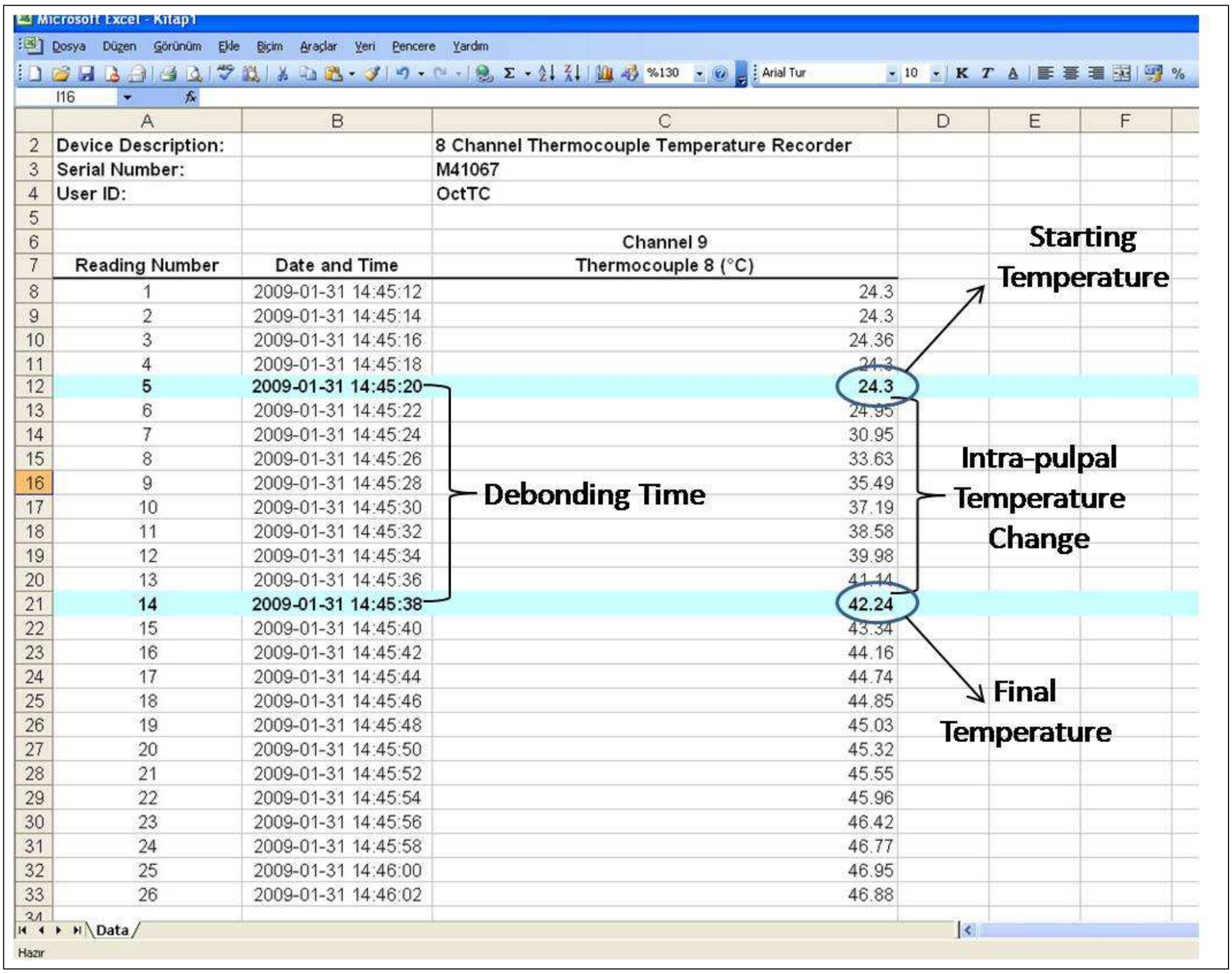


Figure 3.17 Starting point, breaking point, debonding time and intra-pulpal temperature change were shown on the graph on the monitor of the computer which was connected to the K-type thermocouple.

dard deviations were calculated for shear bond strength and intrapulpal temperature change for each group. A student t-test was performed with a. 05 level of confidence to identify statistically significant differences.

The post debonding surfaces of the bracket base and those of the teeth were observed with a stereoscopic microscope, the residual adhesive on the surface of enamel or bracket was evaluated, post lasing photos of the brackets and the enamel were taken and broken bracket wings were reported after experiments.



	A	B	C
2	Device Description:		8 Channel Thermocouple Temperature Recorder
3	Serial Number:		M41067
4	User ID:		OctTC
5			
6			Channel 9
7	Reading Number	Date and Time	Thermocouple 8 (°C)
8	1	2009-01-31 14:45:12	24.3
9	2	2009-01-31 14:45:14	24.3
10	3	2009-01-31 14:45:16	24.36
11	4	2009-01-31 14:45:18	24.3
12	**5**	**2009-01-31 14:45:20**	**24.3**
13	6	2009-01-31 14:45:22	24.95
14	7	2009-01-31 14:45:24	30.95
15	8	2009-01-31 14:45:26	33.63
16	9	2009-01-31 14:45:28	35.49
17	10	2009-01-31 14:45:30	37.19
18	11	2009-01-31 14:45:32	38.58
19	12	2009-01-31 14:45:34	39.98
20	13	2009-01-31 14:45:36	41.14
21	**14**	**2009-01-31 14:45:38**	**42.24**
22	15	2009-01-31 14:45:40	43.34
23	16	2009-01-31 14:45:42	44.16
24	17	2009-01-31 14:45:44	44.74
25	18	2009-01-31 14:45:46	44.85
26	19	2009-01-31 14:45:48	45.03
27	20	2009-01-31 14:45:50	45.32
28	21	2009-01-31 14:45:52	45.55
29	22	2009-01-31 14:45:54	45.96
30	23	2009-01-31 14:45:56	46.42
31	24	2009-01-31 14:45:58	46.77
32	25	2009-01-31 14:46:00	46.95
33	26	2009-01-31 14:46:02	46.88

Figure 3.18 The final temperature at the breaking point was found by the equalance of irradiation and debonding time intervals. The difference between the final temperature and the temperature at the starting point gave the intra-pulpal temperature change.

4. RESULTS

4.1 Laser Power

Two set of experiments with different mode of laser operation were performed. Experiments in continuous wave (CW) Mode, laser was applied on samples with an arranged constant power in continuous mode. According to those energy levels, samples were divided into 5 different groups. Beside those groups that brackets debonded with laser energy, there was also a control group debonded without laser application. In the second section, experiments in Pulse Mode, laser was applied with an arranged ampere and duty cycle in pulse mode. Lasing dose was set as a current value of 4.99 A that had an average measured power of 18.0 ± 0.1 Watts. Best fit 3 types pulse durations in these experiments were selected by preliminary experiments.

GROUP	2	3	4	5	6
Basal Area of Brackets (mm^2)	10.35 ± 0.41	10.35 ± 0.41	10.35 ± 0.41	10.35 ± 0.41	10.35 ± 0.41
Power density (W / cm2)	1.36×10^{-3}	2.07×10^{-3}	2.70×10^{-3}	3.40×10^{-3}	4.07×10^{-3}

Figure 4.1 Power densities of laser energy applied on brackets were found per each group in CW mode.

Groups of experiments in pulse mode	Basal Area of Brackets (mm^2)	Power density (W / cm2)
	10.35 ± 0.41	1.74×10^{-3}

Figure 4.2 Power densities of laser energy applied on brackets were found per each group in pulse mode.

4.2 Load at Breaking Point

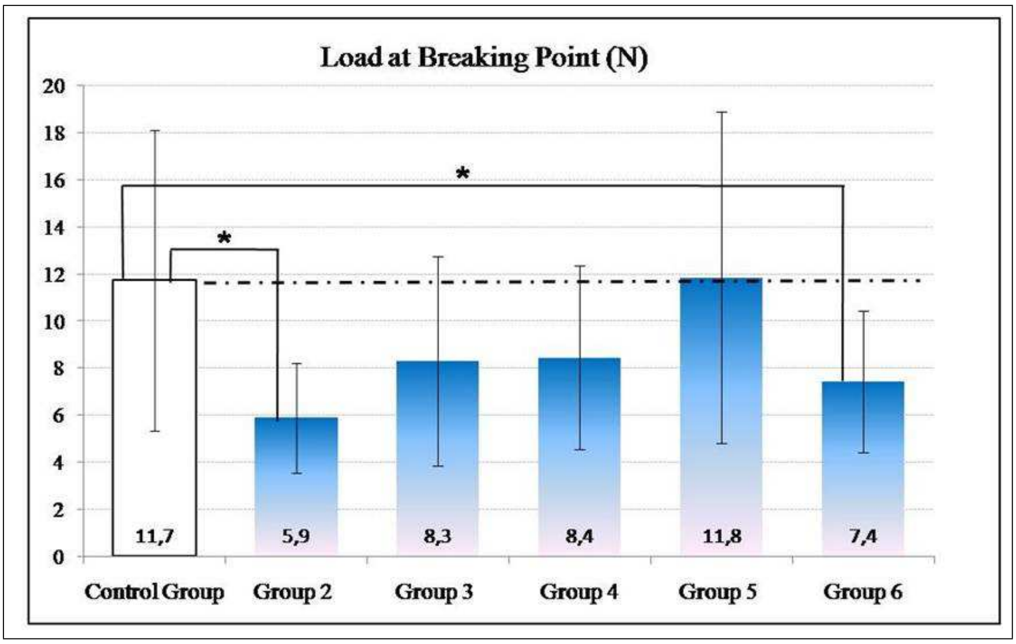


Figure 4.3 Average and standard deviation of load at breaking point for each CW mode group are shown on graph above. Group 2 and 6 are significantly less than the control group (*) ($p \leq 0.05$, student t-test).

In tension tests done by universal testing machine, recorded maximum value for load at the debonding moment of ceramic bracket from the enamel is called "Load at breaking point". Average and standard deviations for each group are calculated. The results are shown below in figures. According to these results, in both sections, Laser energy can significantly decrease the required debonding load for ceramic brackets.

In experiments in CW mode, Group 2 and Group 6 brackets were debonded with significantly less force than the control Group. Among Laser applied groups, load at breaking point of Group 2 is significantly less than Group 4 and Group 5. Besides, Group 6 is significantly less than Group 5. Analysis found no significant differences between the other groups (Figure 4.3).

In second set of experiments, just Group 200/600 and Group 300/900 have a significant decrease in force according to Control group. And maximum load for both groups are less than for Group 400/1200 (Figure 4.4).

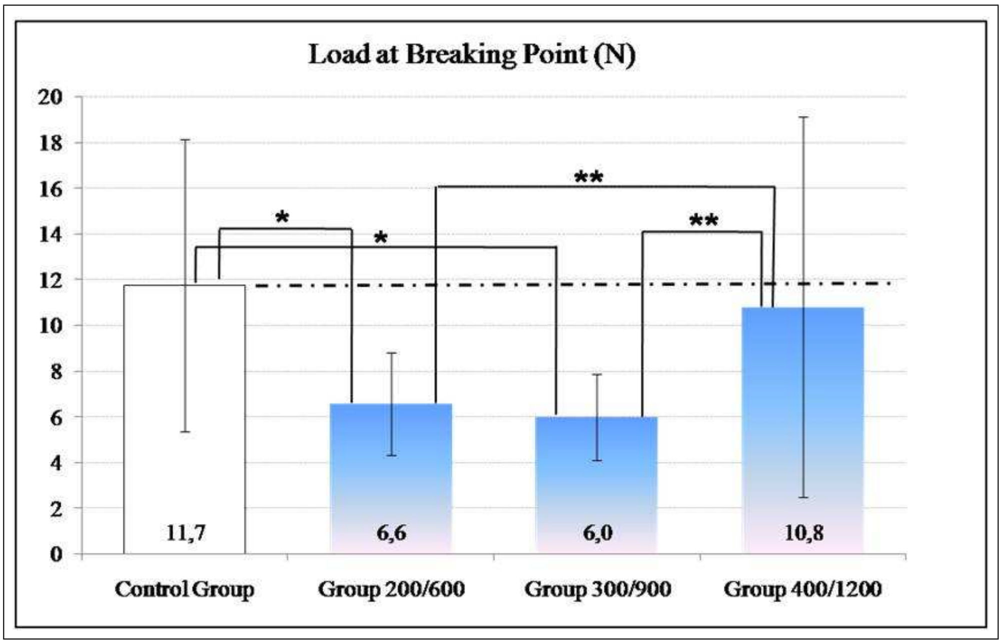


Figure 4.4 Average and standard deviation of load at breaking point for each pulse mode group are shown on graph above. Group 200/600 and Group 300/900 are significantly less than the control group (*)($p \leq 0.05$, student t-test) . Also, Group 200/600 and Group 300/900 are less than Group 400/1200 (**)($p \leq 0.05$, student t-test) .

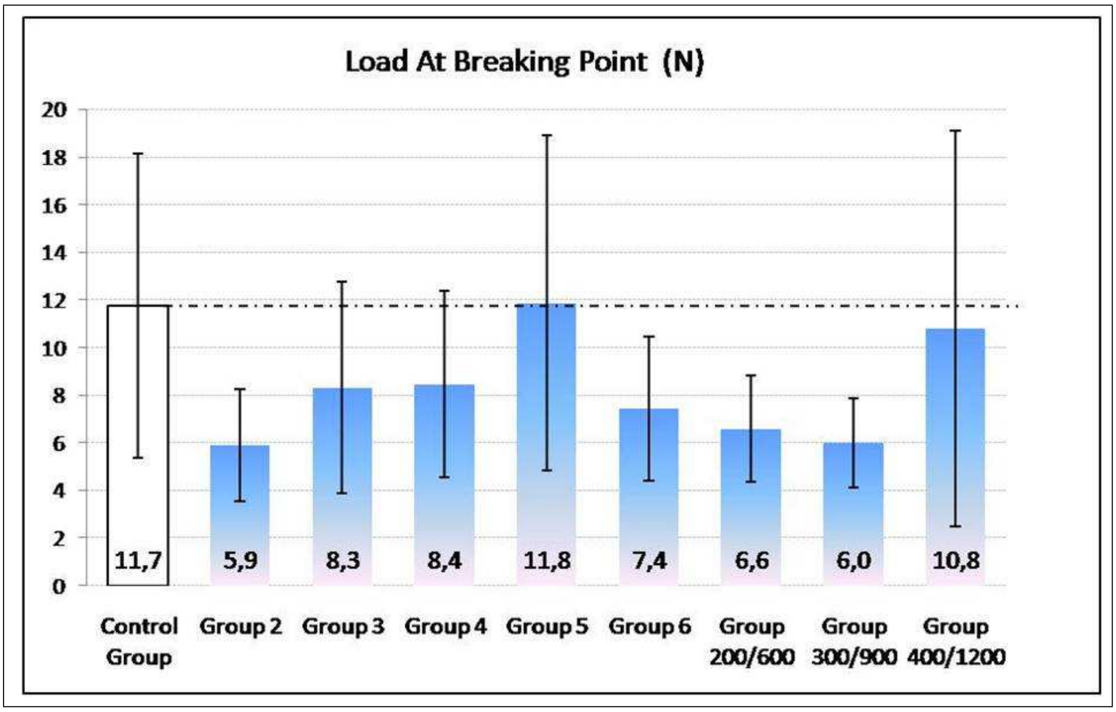


Figure 4.5 Debonding force for Group 2 is significantly less than Group 400/1200. Group 300/900 significantly requires less force than Group 4 and 5(**) ($p \leq 0.05$, student t-test). There is no significant differernce between Group 3 and 6 and the groups in pulse mode.

All in all, while comparing load values among all irradiated groups; debonding force for Group 2 is significantly less than Group 400/1200. Group 300/900 significantly requires less force than Group 4 and 5. There is no significant difference between Group 3 and 6 and the groups in pulse mode (Figure 4.5). There is no significant difference between Group 2 and Group 300/900.

4.3 Debonding Time

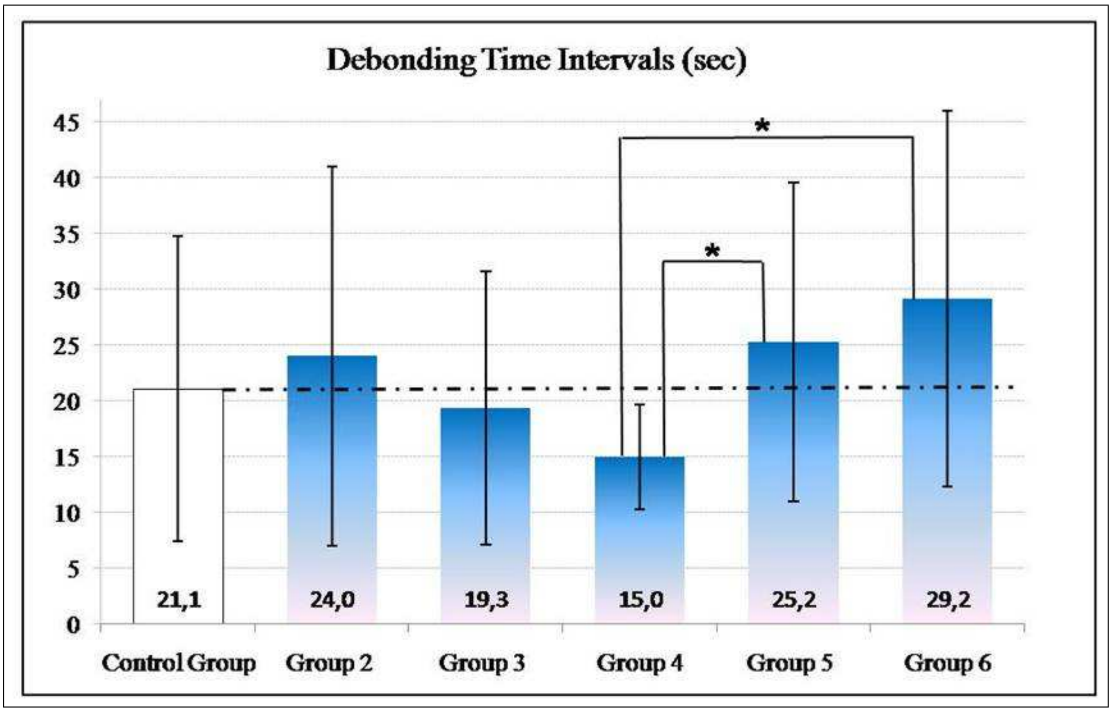


Figure 4.6 Debonding time intervals for each group in CW mode. Irradiation of the bracket by the 1070-nm Ytterbium fiber Laser had no effect on time spent till debonding. The comparisons between the lased groups show that just Group 4 has significantly less time from group 5 and group 6. (*) ($p \leq 0.05$, student t-test).

Each irradiation was started at the first moment of the increase in tension test. Time interval starting from this moment to the moment at the breaking point is called "Debonding Time". Irradiation and debonding time ranges are both equal to debonding time interval. Average and standard deviations for debonding time interval for each group are summarized below.

In first section, irradiation of the bracket by the 1070-nm Ytterbium fiber Laser

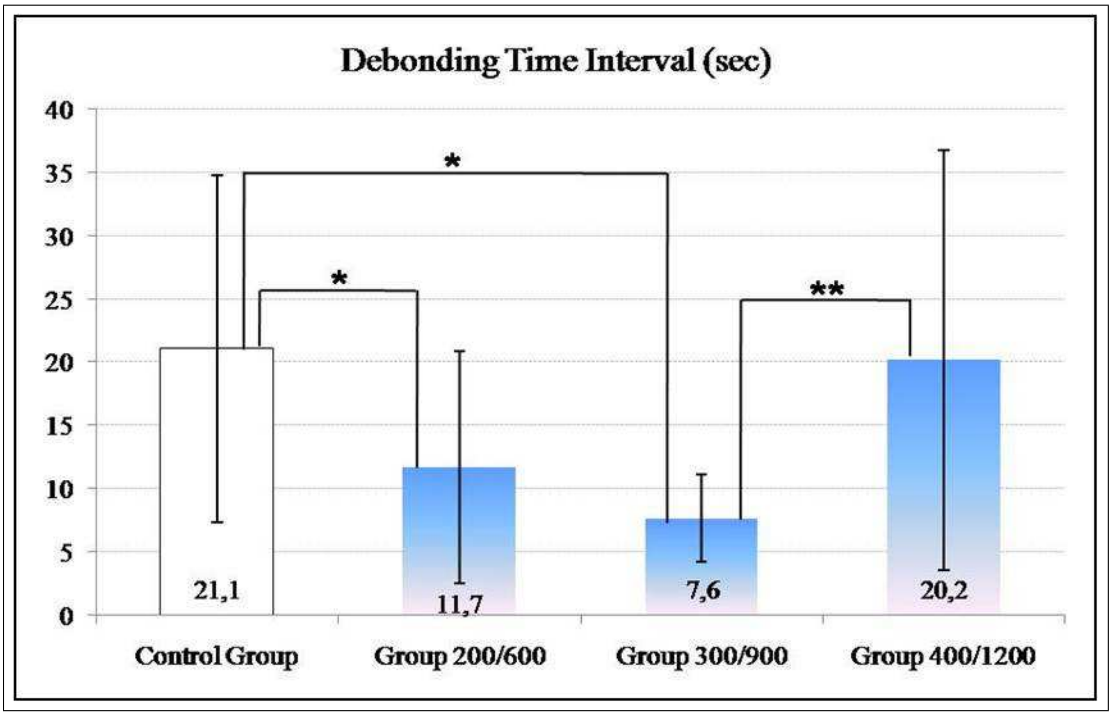


Figure 4.7 Debonding time intervals for each group in pulse mode. Group 200/600 and Group 300/600 are significantly less than the control group (*) ($p \leq 0.05$, student t-test). The comparisons between the lased groups show that just Group 300/900 has significantly less time from Group 400/1200 (**) ($p \leq 0.05$, student t-test).

had no effect on time spent till debonding. The comparisons between the lased groups show that just Group 4 has significantly less time from Group 5 and Group 6 (Figure 4.6).

In second section, Group 200/600 and Group 300/600 requires significantly less time than the control group The comparisons between the lased groups show that Group 300/900 has significantly less time than Group 400/1200 (Figure 4.7).

Moreover, while comparing all irradiated groups among themselves, Group 300/900 in pulse mode has significantly less debonding time than all groups has in CW mode. Also Group 2, 5 and 6 significantly requires more time than Group 200/600 (Figure 4.8).

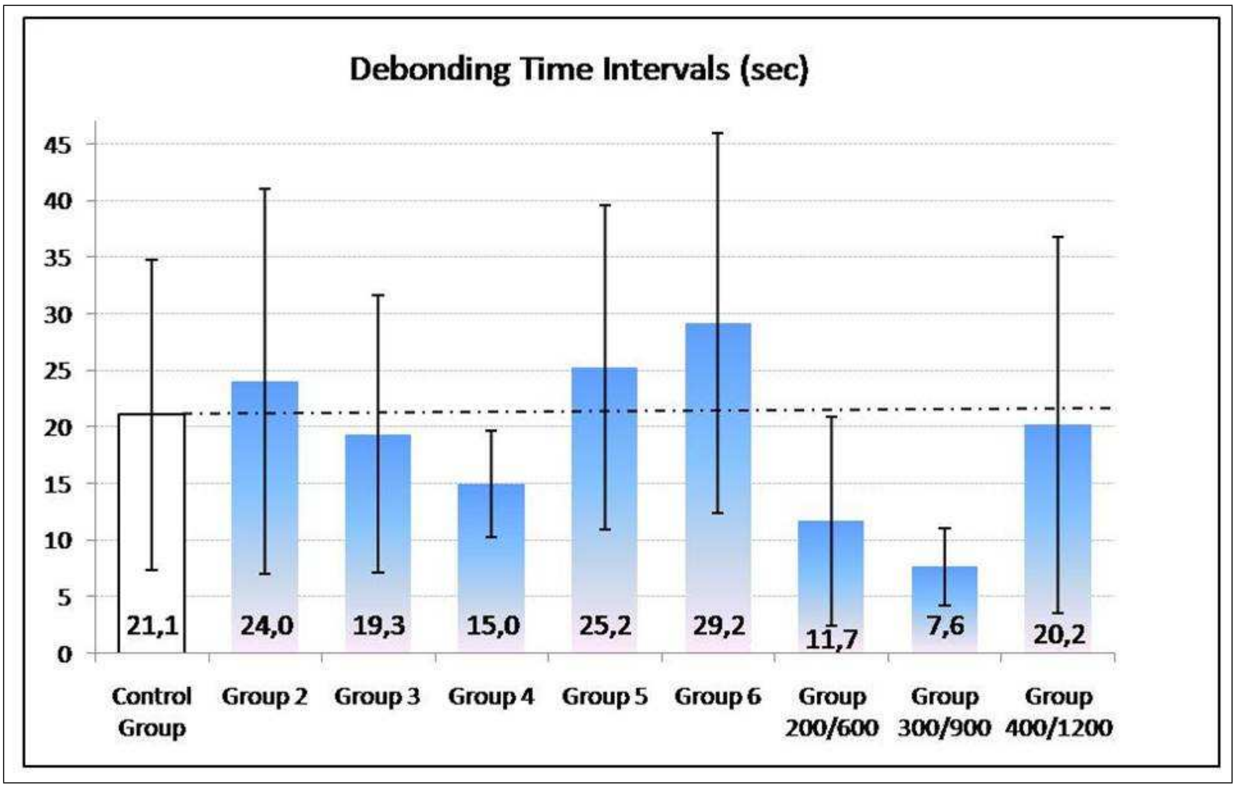


Figure 4.8 Debonding time intervals for each irradiated group is shown above. Group 300/900 in pulse mode has significantly less debonding time than all groups have in CW mode (*) ($p \leq 0.05$, student t-test). Also Group 2, 5 and 6 significantly requires more time tan Group 200/600 (**) ($p \leq 0.05$, student t-test).

4.4 Intrapulpal Temperature Changes

The temperature changes inside the pulp were measured and recorded by a thermocouple during irradiation. Intrapulpal temperature change is the temperature difference between temperature measured at the beginning of lasing and at the breaking moment. According to Zach and Cohen's [1] observations, 5.5 °C is accepted as threshold value for pulpal damage in this study. Also, difference of 2.2 °C which was reported as no histological changes is discernible [2], is taken in account in results. Averages and standard deviations for intrapulpal temperature changes for each group are shown in Figure 4.9, Figure 4.10 and Figure 4.11.

In first section, intrapulpal temperature changes increase directly proportional to the applied Laser powers. There are significant differences between all irradiated

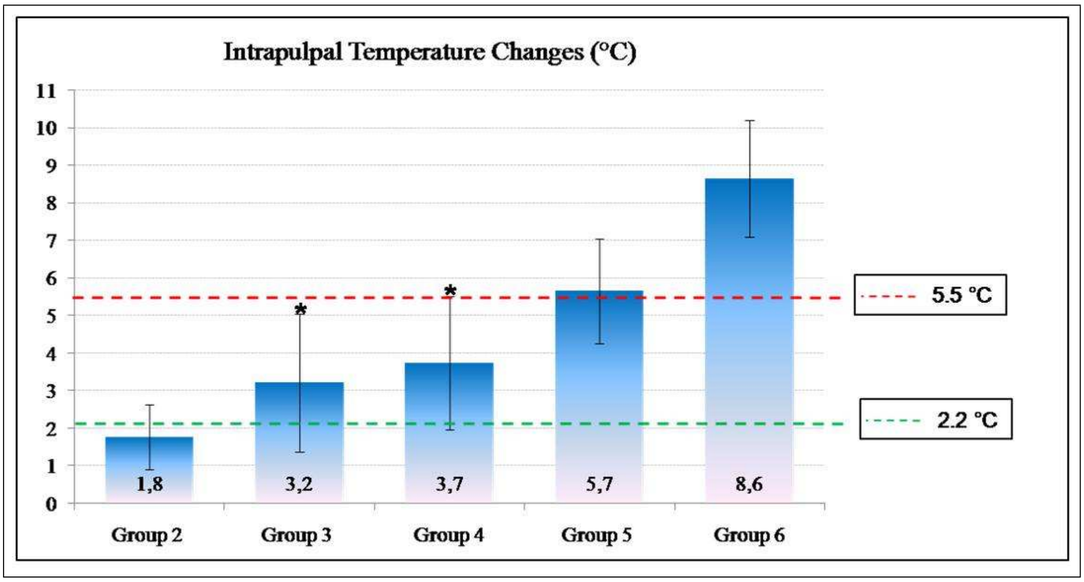


Figure 4.9 Graph of the temperature rise of the pulp chamber wall at lasing in CW mode. There are significant differences between all irradiated groups except between Group 3 and Group 4 (*) ($p \leq 0.05$, student t-test). The intrapulpal differences for Group 2, 3 and 4 are below the threshold value 5.5 °C. Moreover, the difference value for Group 2 is below 2.2 °C which is mentioned as the threshold of histological changes

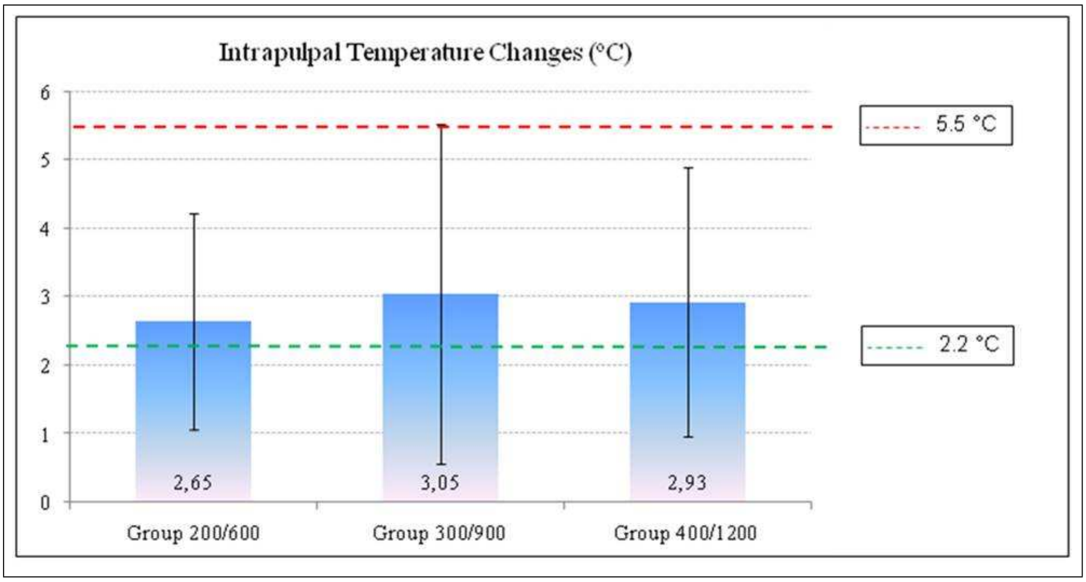


Figure 4.10 Graph of the temperature rise of the pulp chamber wall at lasing in pulse mode. There is no significant difference between irradiated groups. The intrapulpal differences for Group 200/600, Group 300/600 and Group 400/1200 are below the threshold value 5.5°C and none of them is below 2.2 °C which is accepted as the threshold of histological changes.

groups except between Group 3 and Group 4. The intrapulpal differences for Group 2, 3 and 4 are below the threshold value 5.5 °C. Moreover, the difference value for Group 2 is below 2.2 °C which is mentioned as the threshold of histological changes Figure

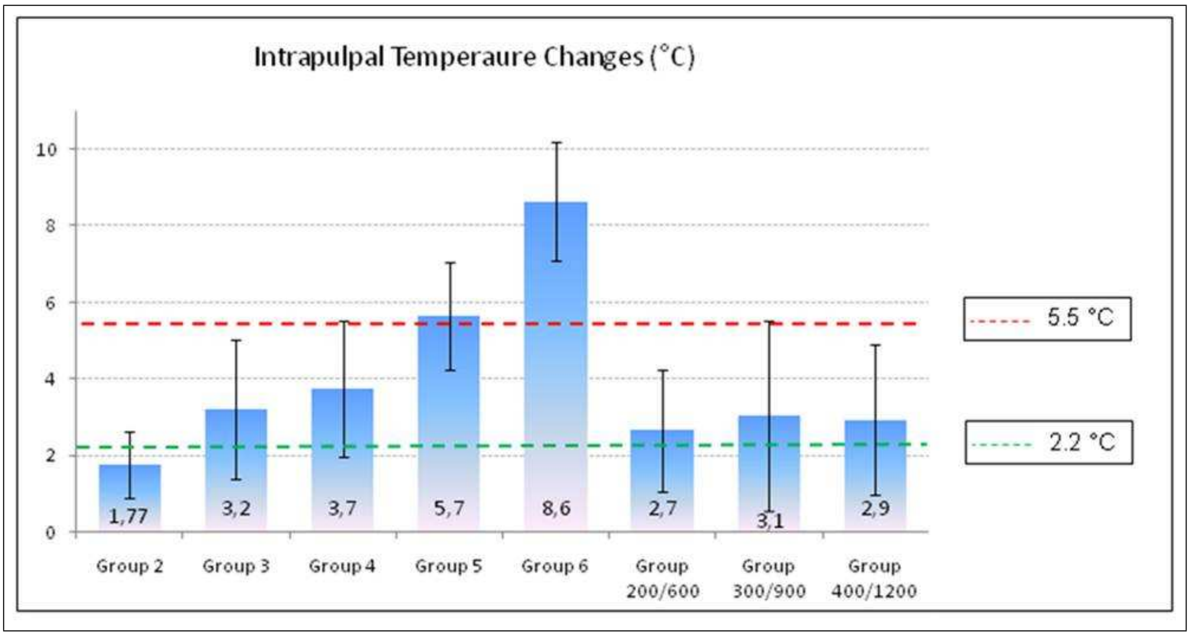


Figure 4.11 As shown in intrapulpal temperature graph for all irradiated groups, Group 5 and 6 have significantly higher intrapulpal temperature increase (*) ($p \leq 0.05$, student t-test).

4.9.

In second section, there is no significant difference in intrapulpal temperature changes between irradiated groups. The intrapulpal differences for Group 200/600, Group 300/600 and Group 400/1200 are below the threshold value 5.5 °C. None of them is below 2.2 °C which is accepted as the threshold of histological changes Figure 4.10.

Among all lased groups, Group 5 and 6 have significantly higher intrapulpal temperature increase Figure 4.11.

4.5 Work Done

Universal testing machine pulled the bracket with an increasing force during the debonding process. Work done by the universal testing machine against the bonding

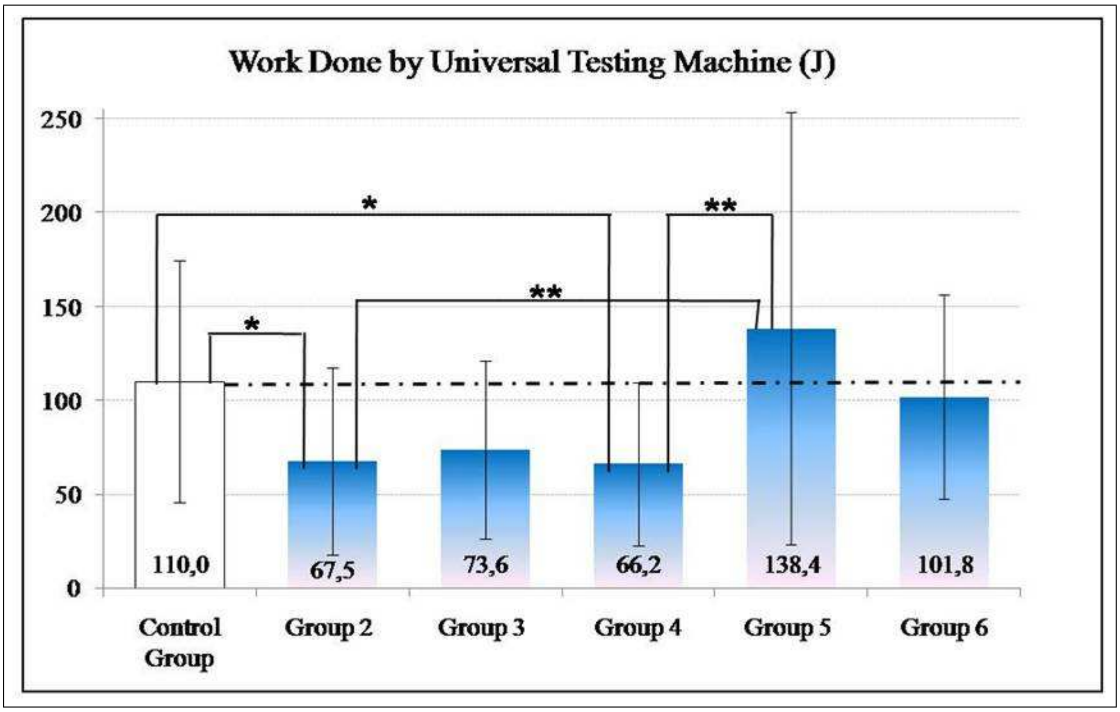


Figure 4.12 Work done by universal testing machine during debonding interval is shown on graph for each group in CW mode. Group 2 and Group 4 are significantly less than Control Group (*) ($p \leq 0.05$, student t-test). Among the irradiated groups, there is just two significant differences that Group 5 spends more energy than Group 2 and Group 4 to debond brackets (**) ($p \leq 0.05$, student t-test).

strength was calculated simply by taking the product of the Load and debonding time. For both sections, Laser irradiation decreased the load applied for debonding; consequently work done by the testing machine was changed due to laser irradiation.

In section 1, Group 2 and Group 4 are significantly less than Control Group. Among the irradiated groups, there are just two significant differences that Group 5 spends more energy than group 2 and group 4 to debond brackets (Figure 4.12).

In section 2, Group 200/600 and Group 300/600 are significantly less than Control Group. Among the irradiated groups, spent energy in Group 400/1200 is significantly more than Group 200/600 and Group 300/600 to debond brackets (Figure 4.13). Also, all groups in CW mode spend more energy than Group 200/600 and 300/900 (Figure 4.14).

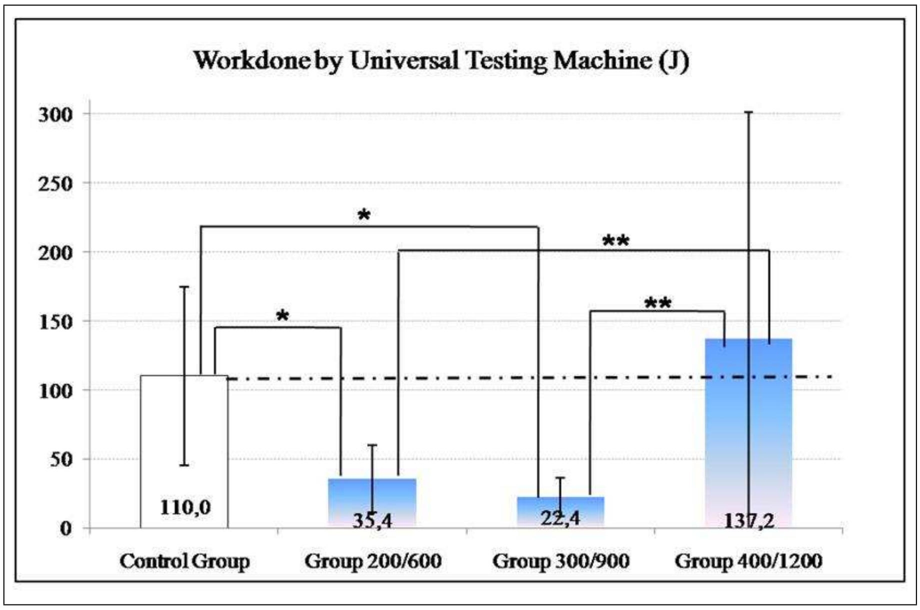


Figure 4.13 Work done by universal testing machine during debonding interval is shown on graph for each group in pulse mode. Group 200/600 and Group 300/600 are significantly less than Control Group (*) ($p \leq 0.05$, student t-test). Among the irradiated groups, spent energy in Group 400/1200 is significantly more than Group 200/600 and Group 300/600 to debond brackets (**) ($p \leq 0.05$, student t-test).

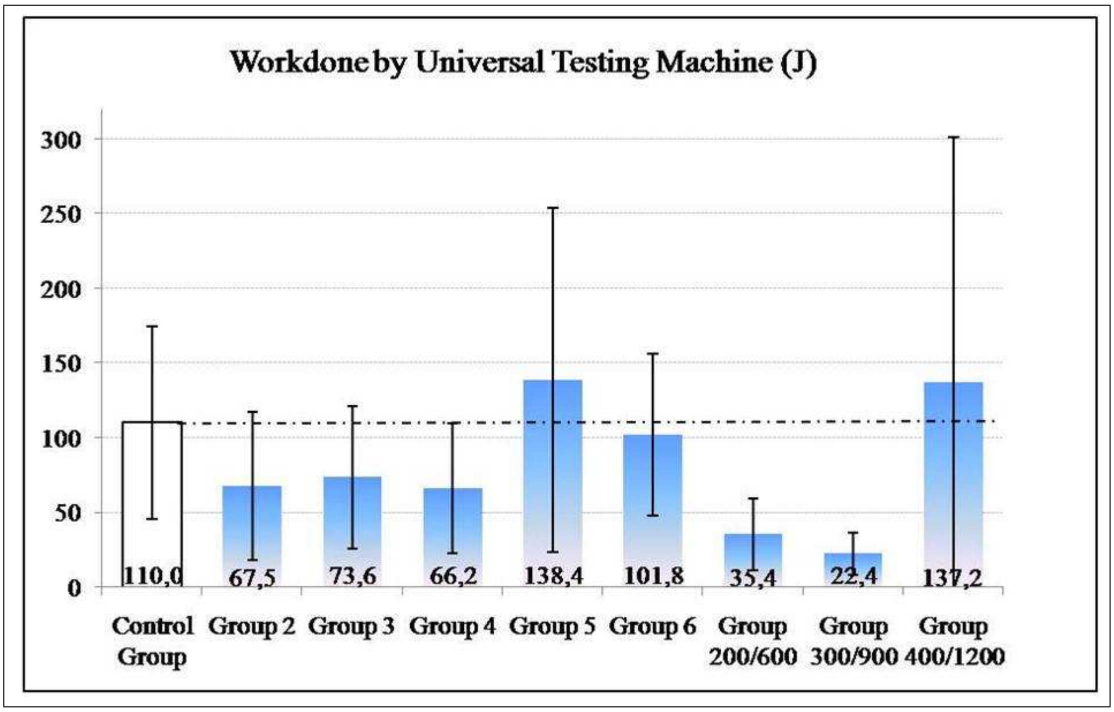


Figure 4.14 For all groups, work done by universal testing machine during debonding interval is shown on graph. All groups in CW mode spend more energy than Group 200/600 and 300/900 (*) ($p \leq 0.05$, student t-test).

4.6 Applied Laser Energy

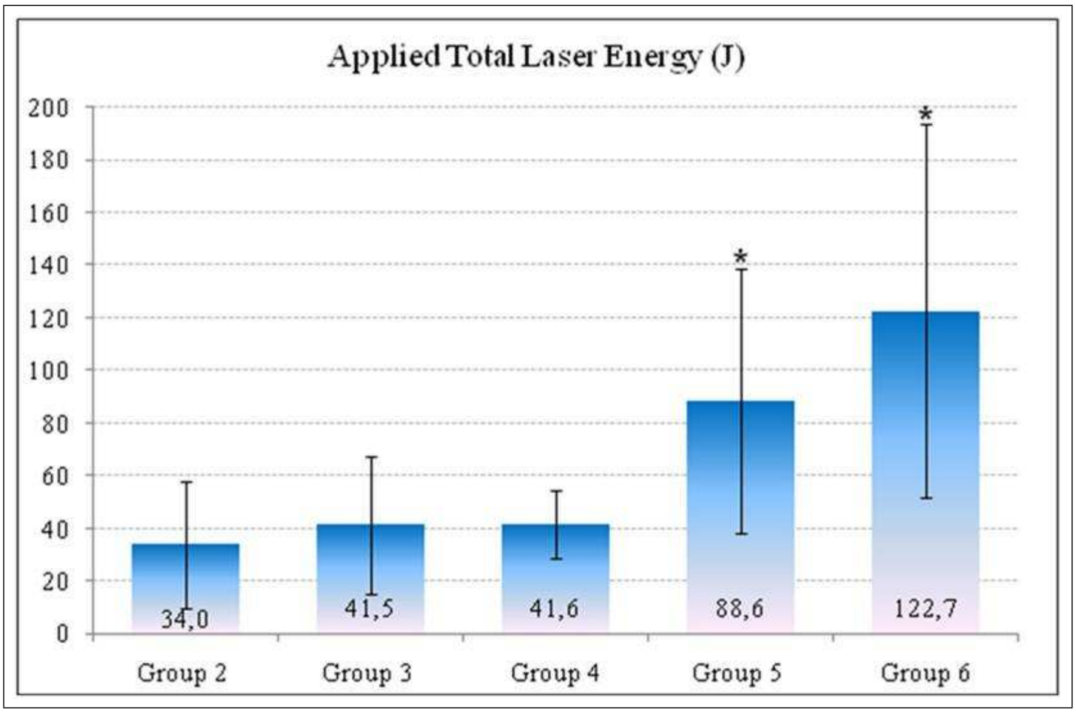


Figure 4.15 Averages and standard deviations of applied total laser energies for each group in CW mode are shown on graph. Applied Laser energies for Group 5 and 6 are significantly more than the other lased groups (*) ($p \leq 0.05$, student t-test). No significant difference was seen between the other groups.

The Laser energy applied on to the samples is calculated by the measured output power level of laser and irradiation time. Averages and standard deviations for each group are shown in Figure 4.15, Figure 4.16 and Figure 4.17.

In first section, applied Laser energies for Group 5 and 6 are significantly more than the other lased groups. No significant difference was seen between the other groups (Figure 4.15).

In second section, applied Laser energy for Group 300/900 is significantly less than Group 400/1200 (Figure 4.16).

While comparing two sections, total Laser energies applied in Group 2, 3 and 4 are significantly less than Group 400/1200. Group 5 and 6 have significantly more total laser energy applied than Group 200/600 and 300/900 (Figure 4.17).

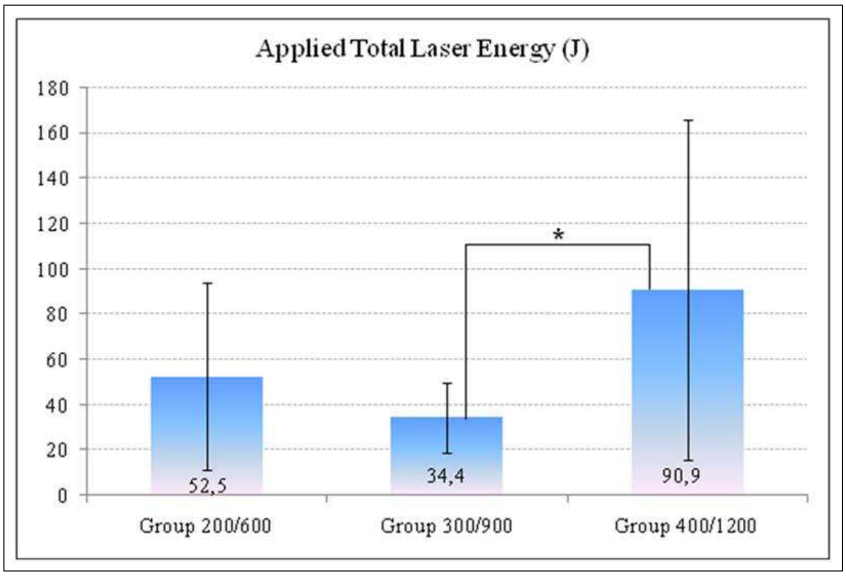


Figure 4.16 Averages and standard deviations of applied total laser energies for each group in pulse mode are shown on graph. Applied Laser energies applied Laser energy for Group 300/900 is significantly less than Group 400/1200 (*) ($p \leq 0.05$, student t-test).

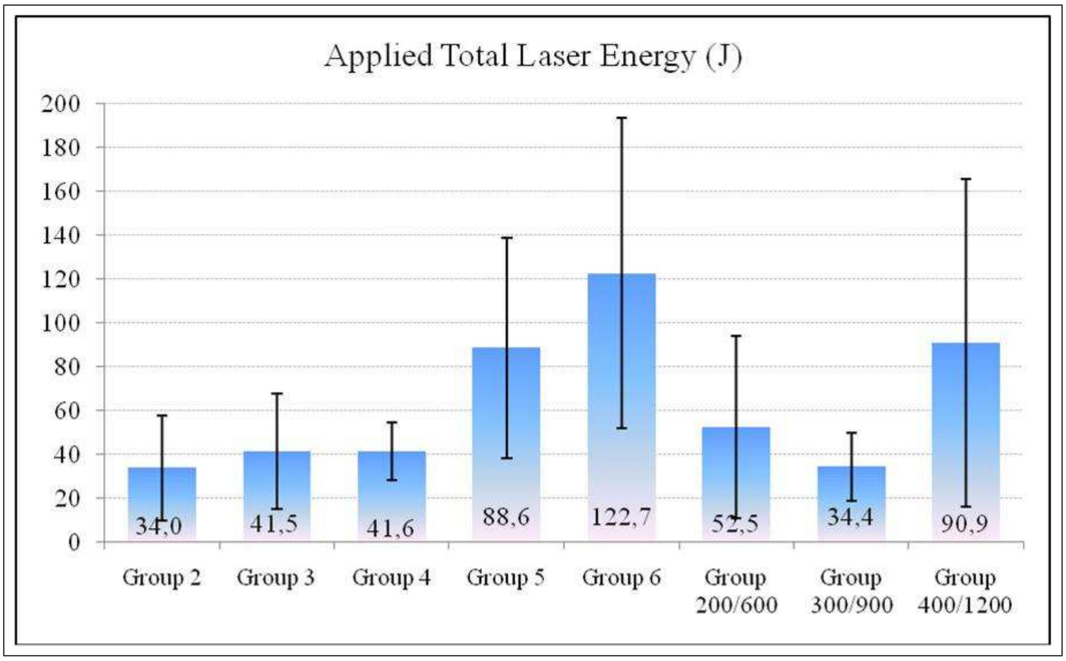


Figure 4.17 Averages and standard deviations of applied total laser energies for each group are shown on graph. Total Laser energies applied in Group 2, 3 and 4 are significantly less than Group 400/1200 (*) ($p \leq 0.05$, student t-test). Group 5 and 6 have significantly more total laser energy applied than Group200/600 and 300/900 (*) ($p \leq 0.05$, student t-test).

4.7 Examination of Dental Surface by a Stereomicroscope

Post-lasing surfaces of teeth and ceramic brackets were examined by a stereomicroscope for evidence of residual adhesive and post-lasing photos were taken. There are three kinds of results. In most of the samples residual adhesive was totally observed on the enamel. Most of the bases were clean and out of remnant adhesive of adhesive but small hollows appeared on the bracket bases, the remnant resin appeared to be burned out directly under the corners of brackets. Laser powers applied with these time intervals were not able to debond without remnant adhesive on the enamel surfaces. In another group it was observed that the adhesive was debonded on the bracket‘s base. There was no remnant adhesive on the enamel. And also rarely observed that the bracket‘s wing was broken during the debonding procedure.

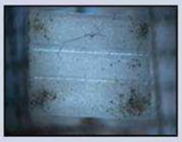
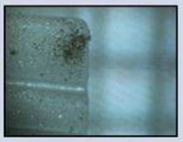
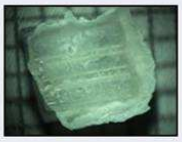
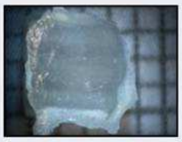
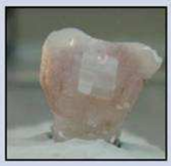
Three kinds of observations by microscope	Enamel Surface	Bracket Base	Percentage of Samples in CW Mode	Percentage of Samples in Pulse Mode
Adhesive remnant on the enamel		 	**83.36 %**	**79.27 %**
Debonded adhesive on the bracket		 	**11.36 %**	**18.67 %**
Broken bracket wing			**2.27 %**	**2.13 %**

Figure 4.18 Example for three kinds of results observed by stereomicroscope, either in lased groups or in non- lased group, are shown in the figure above.

4.8 Summary

1. In CW mode, Group 2 and Group 6 require significantly less force than the control Group (11.7 N). All in all, load at breaking point of Group 2 is significantly less than Group 4 and Group 5. In second section, in experiments in pulse mode, Group 200/600 and Group 300/900 have significant decreases in force according

to Control group. Among all irradiated groups there is no significant difference between Group 2 (5.9 N) and Group 300/900 (6.0 N) which have best results according to load values at breaking point.

2. Irradiation of the bracket by the 1070-nm Ytterbium fiber Laser had no effect on time spent till debonding in CW mode. In pulse mode Group 200/600 (11.7 sec) and Group 300/600 (7.6 sec) requires significantly less time than the control group (21.1 sec). According to the time spent during debonding procedure, the best values is Group 300/900 which has also significantly less debonding time than all groups have in CW mode.

3. All groups except Group 3 and Group 4 in CW mode are significantly different from each other but there is no significant difference between groups in pulse mode. Group 2,3 and 4 in CW mode and all groups in pulse mode are below the threshold value 5.5 °C. Just Group 2 is below 2.2 °C which is accepted as the threshold of histological changes. All groups are below the threshold value in pulse mode. Among those accepted results which are below the threshold value, it is seen that just Group 2 is significantly less than Group 5, Group 6 and Group 400/1200.

4. Work done by universal testing machine is significantly less in Group 2 (67.5 J), Group 473.6 j), Group 200/400 (35.4 J) and Group 300/900 (22.4 J) than Control group (110.0 J). Also, Group 200/400 and 300/900 have less work done values than all other groups.

5. Group 2, 3, 4, 200/400 and 300/900 show no significance among each other but totally more laser energy applied on samples in Group 5, 6 and 400/1200 during debonding.

6. The number of samples that has adhesive remnant on the enamel surface in CW mode applications is more than in pulse mode applications.

5. DISCUSSION

Traditional bracket debonding is achieved by applying a sufficiently large force to break the bond. These forces may tear out enamel. As a gentler procedure the lasers reduces the incidence of damage caused by debonding while permitting the maximum possible bond strengths to be used. In previous laser debonding studies, CO_2 (10600 nm), Nd:YAG (1060 nm), KrF (248 nm), XeCl (308 nm) Tm: YAP (1980 nm) and GaAlAs (808 nm) lasers were used experimentally for debonding ceramic brackets. In those studies, purposes like the laser-tissue interactions and transmissibility of the wavelength was considered during the selection of the laser for debonding ceramic brackets experiments. A carbon dioxide laser, whose wavelength is more easily absorbed by the ceramic brackets, has been usually the predominant choice for debonding experiments. On the other hand, Hayakawa used a Nd:YAG laser because of its higher degree of enamel transmissibility than a carbon dioxide laser. In the present study, 1070-nm Ytterbium fiber Laser which was not used before for debonding ceramic brackets, is chosen because of the advantages of fiber lasers over other types and properties of that wavelength. As mentioned before fiber lasers are easy to use and economical because of their compact size and extended lifetime. Light is already coupled into a flexible fiber so it can be easily delivered to a movable focusing element. Also a fiber laser has an high output power that can provide very high optical gain. High optical quality of this type laser is also another advantage because the fiber's waveguiding properties reduce or eliminate thermal distortion of the optical path. It is also acceptable according to the properties of this wavelength, because it is known that the minimal value for absorption coefficient of hydroxapatite is observed in visible and near infra-red region.

Any process that degrades the bonding resin will facilitate debonding. Laser energy can degrade the adhesive resin by thermal softening, thermal ablation and photoablation [50]. Thermal softening occurs when the laser heats the bonding agent until it softens. So the bracket succumb to gravity and slide off the tooth surface.

Thermal ablation occurs when heating is fast enough to raise the temperature of resin into vaporization its range before debonding by thermal softening occurs. This results in the bracket is being blown off the tooth surface and photo ablation also results in the bracket's being blown off the tooth surface. It occurs when very high energy laser light interacts with the adhesive material and the energy level of the bonds between the adhesive resin atoms rapidly rises above their dissociation energy levels resulting in decomposition of the material.

Neither thermal ablation nor photo ablation was examined during in these experiments. All the brackets debonded by sliding down off the tooth under the influence of the applied load and fibrous form of the softened bonding agent were observed on the base of the bracket after debonding procedure so thermal softening is accepted as responsible for debonding mechanism in this study. The result of this research agree with those of Strobl et al.[51], Tocchio et al.[50], and Rickabaugh et al. [53] in that the laser can effectively thermally soften the adhesive to permit ceramic bracket removal. Tocchio et al. [50] reported debonding of monocrystalline brackets with a 1060-nm wavelength occurred by either photo ablation or thermal ablation at power densities greater than 26W/cm2 and power densities greater than 32 W/cm^2 for polycrystalline brackets, also they observed thermal softening at lower powers. Per contra, Hayakawa [56] 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 in the 2.0J and 3.0J groups debonded immediately after laser irradiation without mechanical effects. Also, Hayakawa [56] explained explosive "blow off" during lasing as the result of thermal ablation or photoablation, rather than thermal softening. Furthermore, Mimura et al. [18] and Obata et al. [52] suggested both thermal softening and resin contraction from ceramic brackets could be responsible for the debonding mechanism.

According to the all collected and recorded data in this search, results are performed for load at breaking points, debonding time intervals, intrapulpal temperature changes, work done and applied laser energies.

5.1 Load at Breaking Point

According to these results, Group 2, Group 6, Group 200/600 and Group 300/900 require significantly less force than the control group. Laser energy significantly decreased the required debonding load for ceramic brackets. In this study with an average force of 11.73 ± 6.39 N was required to debond the ceramic brackets without lasing. Among 8 irradited groups, Group 2 (5.88 ± 2.34 N) and Group 300/900 (5.97 ± 1.87 N) showed the best significant decreasing in load at breaking point. In other words, lasing caused an 50 % of reduction in load at breaking point in these groups. Assisting the results of this study, Xianglong Han [57] had same percentage of reduction in debonding load by applying Nd:YAG laser at 1060-nm, pulse width with of 0.2 ms and 3W for 3 seconds. And for MMA resin Mimura reported that debonding force was decreased at 3 watts output by using CO_2 laser (From a mean value of 122.40 N to 35.57 N). However, Strobl et al. [51] did not produce any significant additional reduction in debonding force in laser aided bracket removal with a 5 second exposure with a total energy of 120 joules (24 watts) while using a near wavelength,1060-nm Nd:YAG laser. In that same study, a CO_2 laser with a power of 14.1 W for 2 seconds was lased and polycrystalline ceramic brackets showed 1.3-fold decrease in the total energy required for debonding.

5.2 Debonding Time

Although there is no significant difference between CW mode groups and control group, debonding time interval significantly decrased in pulse mode. Group 200/600 and Group 300/900 requires significantly less time than the control group. Whereas debonding time interval for control group was 21.07 ± 13.69 seconds, it was reduced to 7.64 ± 3.45 seconds in Group 300/900. It means these significant decrease in debonding time for Group 300/900 is 63.74 % of the control group. Agreeing to this study, also Mimura et al. [18]found debonding times for both MMA and Bis-GMA resins were significantly less when using CO_2 lasers compared with the unlased control groups with a 3 watts output of power. Comparing a near wavelength, in the study of Tocchio [50], a rapid increase in debonding times was seen when they were observed the power level decreased for 1060 nm. The polycrystalline bracket debonding times averaged less than 4 seconds when using 248 nm, less than 5 seconds when using 308 nm radiation and more than 20 seconds when using 1060 nm radiation. Also they observed debonding types of monocrystalline brackets debonded with an applied load were always less than one second at all wavelengths. But, Obata et al. [52] reported a better result that debonding with the super-pulse CO_2 laser occurs at 2 W within a period of 2.9 ± 0.9 seconds.

5.3 Intrapulpal Temperature Changes

Supporting the results of Hayakawa [56], the temperature of the pulp wall started to increase to its maximum point immediately after lasing in this study. On the other hand, Obata [52] reported that the temperature rise in the pulp chamber starts 3 seconds after lasing.

Thermal ablation and photo ablation proceed rapidly and very little heat diffusion occurs therefore the tooth and the bracket stay near physiologic temperatures but thermal softening is a relatively slow processes which causes a large rise in both tooth and bracket temperature. Like most of the previous observations on intrapulpal temperature increases, 5.5 °C is accepted as threshold value for pulpal damage in this study. Also, difference of 2.2 °C which was named as no histological changes is discernible, is taken in account in the results.

Although results of this study shows that intrapulpal temperature changes increase directly proportional to the applied Laser powers in CW mode irradiation, there is no significant difference in intrapulpal temperature changes between irradiated groups in pulse mode. The intrapulpal differences for Group 2, Group 3, Group 4, Group 200/600, Group 300/600 and Group 400/1200 are below the threshold value 5.5 °C. Moreover, the difference value for Group 2 is below 2.2 °C which is mentioned as the threshold of histological changes. In other words, over a measured power of 3.5 W in CW mode causes that intrapulpal temperature passing the threshold value. Another type of comparision is that Ma et al. attracted attention to that there is a linear relationship between lasing time and an increase in intrapulpal temperature. They lased polycrystalline brackets using a power level of 18 W CO_2 laser in continuous mode. A mean intra pulpal increase of 0.91 °C was observed after one second lasing, 1.74 °C after 2 seconds and 2.67 °C after 3 seconds. In addition to that, they showed debonding with 1.48 MPa of tensile load using a CO_2 laser at 18 W for 2 seconds caused an intrapulpal temperature increase of 1.1 °C. Abdul-Kader and Ibrahim [55] used a CO_2 laser at a power of 50 W and used same lasing time with Ma. They observed that bracket temperature increased up to 93.63 °C, the enamel below the bracket was 23.13 °C and to sum up the intrapulpal temperature difference rose to 0.7 ± 0.4 °C. While Fraunhofer [?]was studying on thermal effects associated with the Nd:YAG dental laser during etching at the power settings of 0.8 W, 1 W, 2W and 3W for 12 seconds, they reported that the temperatures measured at power levels 1-3 W caused pulpal inflammation and possible irreversible damage to pulp tissue. 0.8 W showed an increase of 5.40 ± 1.34 °C, 1 W showed an increase of 7.80 ± 1.40 °C, 2 W showed an increase of 9.80 ± 2.17 °C and 3W showed an increase of 20.60 ± 1.67 °C on the buccal surface of teeth. But that study was for etching before bonding so laser was directly irradiated on the enamel. Suleiman [47] et al. measured pulp chamber by using a diode laser with outputs between 1W and 3W. For an identical comparison with our study the results of upper central teeth were got from that research. The intrapulpal increase was 4.5 ± 0.34 °C for 1 W, 7.5 ± 0.23 °C for 2 W and 10.7 ± 0.42 °C for 3 W. Albert Mehl [?], observed the rise in temperature in the pulp was above 8 °C while using Nd:YAG laser applied with a total energy of 80 J. Rickabaugh [53] used a CO_2 laser with 20 W output power and observed an increase of 1.80 ± 1.73 °C applying 3 pounds in 1.64 ± 0.89 seconds, 1.80 ± 1.73 °C applying 1.5 pounds in 1.83 ± 0.69 seconds and 4.47 ± 3.57 °C applying 0.75 pounds in 3.42 ± 1.32 seconds.

By way of addition, Tocchio reported that polycrystalline brackets were hot to touch whereas the sapphire brackets still felt cold as hot polycrystalline brackets were observed after debonding procedure in present tests.

5.4 Work Done

Work done by the universal testing machine against the bonding strength was related with the load and debonding time. In this study, Laser irradiation decreased the load applied for debonding; consequently work done by the testing machine was changed due to laser irradiation but the both type of the results are not paralel because of the differences in debonding time intervals. As a summary, Group 2, Group 4, Group 200/600 and Group 300/600 are significantly less than the control group. According to student t-test, Group 200/600 and Group 300/900 have the least values among the irradiated groups. This study showes that the work done in control group is 3 times greater than Group 200/600 is and 5 times greater than Group 300/900.

5.5 Applied Laser Energy

The Laser energy applied on to the samples is calculated by the measured output power level of laser and irradiation time. Group 400/1200, Group 5 and 6 are significantly more than the other lased groups. No significant difference was seen between the other groups. Applying more than 3.5 watts of laser power in CW mode showed a sensible rise in total applied laser energy because of the rapid increase in debonding time interval. The same situation is viable for Group 400/1200 in pulse mode, too. As a result, there is a observable striding after 88.6 joules of total energy including both modes. While Hayakawa [56] used an Nd:YAG with three different energy levels : 1.0, 2.0, or 3.0 J. The 3.0 J lasing group, when compared with the nonlasing and the 1.0 J lasing groups, showed a significant decrease in bond strength. All in all, it was reported that the 3.0 J group showed no significant differences among the brackets used. Hayakawa [56] explained debonding was achieved more effectively in the 3.0 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. Therefore, no significant differences were examined among bracket types. Although previous studies suggested that the Bis-GMA bond group had a higher degree of debonding than did the 4-META MMA bond group. For 2.0 J and higher lasing, however, Hayakawa [56]found no statistically significant differences among adhesive resins under identical lasing conditions.

5.6 Examination of Dental Surfaces by a Stereomicroscope

Strobl et al. informed that generally conventional debonding techniques have been used bracket fractures occured 10 % to 35 % of the time. Due to reduction in force, time and energy lasing neither in CW mode nor in pulse mode is effective enough for debonding without any remnant on the enamel surface. However, post-lasing surfaces of teeth and ceramic brackets were examined by a stereomicroscope and post-lasing photos were taken. Residual adhesive was totally observed on the enamel in the 80 % of the samples. The bases were clean and out of remnant adhesive of adhesive but small hollows appeared on the bracket bases, the remnant resin appeared to be burned out directly under the corners of brackets for each lased sample. It is supposed that because of the thickness of the adhesive is the thinnest under the corners of brackets. The area of black hollows are proportionally increased by the level of the power. All specimens had identifiable black deposits at 2.0 J and at 3.0 J, small hollows appeared on the bracket bases, and eruptions of dissolved ceramic were observed around the peripheries in the study of Hayakawa. Conversely to this study, Hayakawa reported the remnant resin directly under the lasing points. And also broken wings were rarely observed during these debonding procedures so there is not enough samples to compare or generalize.

6. Conclusion and Further Works

As a conclusion, significantly decreased bond strength, debonding time and work done were observed while debonding ceramic brackets with a 1070- nm Ytterbium fiber Laser. Lasing caused a 50 % of reduction in required load for debonding and showed a 3- fold decrease in time. Intrapulpal temperature changes are below the accepted threshold value until the level of 3.5 watts of laser power in continuous wave mode. Also applying more than 3.5 watts of laser power showed a rapid increase in total applied laser energy. It can be reported that a sensible striding is observed after 88.6 joules of total energy applied on the ceramic brackets in both modes. Moreover, during debonding, the work done by universal testing machine is diminished up to 5 times by irradiation. All in all, Group 300/900 significantly has best results over control group among the other lased groups. However, in-vitro debonding could now be done on extracted human teeth. In addition to this, temperature increases and tooth strength should be tested under in-vivo conditions because blood flow by pulp circulation may cause a cooling effect on intrapulpal temperature increase. Thus, these issues must be studied further before the clinical use of this technique.

Laser applications in debonding require further improvement because Laser could mean very rapid and painless debonding without the risk of either enamel tear outs or bracket fractures. If debonding can be achieved with lasing alone, mechanical operations during bracket removal become unnecessary, alleviating patient discomfort at bracket removal. The increasing application of lasers to dentistry and the rapidly falling prices of these instruments, this unusual debonding procedure provide further study.

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