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Abstract

Title of thesis: The effects of Neodymium: Yttrium-Aluminum-Garnet laser on the Osseotite® surface: An *in vitro* scanning electron microscope evaluation

Arash Mike Rostami, Master of Science 2013

Thesis directed by: Mark A. Reynolds, DDS, PhD, Chairman, Department of Periodontics

Background: Recently Nd:YAG laser has been proposed as an option to decontaminate the implant surface. The aim of this study was to evaluate the effects of laser irradiation on Osseotite implant surfaces using the Nd:YAG laser. Methods: Four 3i Osseotite® implants were used with each implant providing 3 sides for observation. The implants were mounted in jig that moved the surface perpendicularly past a Nd:YAG laser beam. After positioning at 3mm or 10mm a fresh surface was irradiated with either 0.8w, 2w, or 3w delivered through a 360nm fiber optic tip. The specimens were evaluated using SEM microscopy and the surface area of alteration was recorded using Image J software. Results: The results demonstrated that implant surfaces irradiated with Nd:YAG, regardless of distance (3mm, or 10mm) or power setting (0.8-3W), showed alteration of the Osseotite® surface. Repeated ANOVA analysis showed surface alterations were significant with increased power settings, but not to distance (P=0.005). The altered areas on the implants included charring, bluing, loss of surface roughness and in severe cases, melting, blistering and loss of surface layer as viewed by SEM and clinical photos. Conclusion: Regardless of power setting or distance, use of Nd:YAG on the Osseotite® surfaces will produce surface alterations. The implications of these surface alterations on re-osseointegration are not known.

The effects of Neodymium: Yttrium-Aluminum-Garnet laser on the Osseotite® surface:

An *in vitro* scanning electron microscope evaluation

By

Arash Mike Rostami, DDS

Thesis submitted to the faculty of the Graduate school of the
University of Maryland, Baltimore in partial fulfillment
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Master of Science
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DEDICATION

To my mother and father, and my family who have been a constant inspiration for me.

To Drs. Mark A. Reynolds, Mary-Beth Aichelmann-Reidy, and all of my co-residents for their invaluable guidance and input, without which this work would not be possible.

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I. INTRODUCTION

As biomedical sciences progresses, new innovative solutions to treatment of oral diseases are introduced, one such example is the dental implant. Dental implants have afforded patients the ability to replace an edentulous area, improving esthetics, function, and quality of life. Unfortunately, dental implants have been shown to be susceptible to some of the same diseases that affect natural teeth. Clinicians have studied various options of treating an ailing implant afflicted with peri-implant disease. This paper sets out to give some historical information about the tooth replacement option of dental implants, various treatment options in treating implant diseases, and the results of an *in vitro* study of the effects of Nd:YAG laser treatment on Osseotite® implant surfaces.

Tooth replacement options

There are several options available to the clinician and the patient for replacement of missing teeth. In a study of 200 volunteers involving 121 females and 79 males divided into four treatment categories: fixed partial denture (FPD), removable partial denture (RPD), implants, and no treatment; found that the majority of the subjects in the study reported the main reason for replacement of missing teeth was function and esthetics.[1] The most important factor affecting the choice between the restorative options was damage to adjacent teeth, pain, post-operative discomfort, cost, patient awareness of different treatment options, and dental phobia.[1] Greater satisfaction was seen with dental implants and FPD options, while lesser satisfaction was found in the RPD and no treatment groups.[1] Some factors in treatment planning for the implant

restoration of missing teeth include: patient's medical history, amount of bone available to anchor an implant, bone density, anatomy of and proximity to vital structures, addiction, smoking, parafunctional habits, occlusion, and patient expectations[2]. Implants have been shown to have a favorable survival rate (97.3%) as a tooth replacement option[3].

History of dental implants

Humans have been trying to replace missing teeth with implants much longer than originally thought. An archaeological excavation in a Neolithic cemetery in Ramallah, Egypt yielded, amongst other finds, a life sized shell carving of a human tooth.[4] Based on its morphology, it most closely resembled a left maxillary central incisor.[4] There have been several interpretations of its function, including the possibility that it was intended for dental implantation.[4] Archeological dating suggests the tooth or implant is 5,500 years old.[4] Other implant options such as gold have been discovered in other archeological sites. Earliest proof of osseointegrated implants comes from the Mayan civilization when, in 1931, a fragment of a human mandible was found which was thought to belong to a female in her 20s dating back to 600AD.[5] In the 1970s, a radiograph of the mandible was taken, showing bone formation around coral implants, suggesting they were inserted while the patient was alive and not post mortem as originally suggested.[5]

The modern age of dental implants was accidentally discovered by Per Ingvar Branemark in 1952. An orthopedic surgeon by training, Branemark was interested in the study of bone healing, and marrow micro-vascular circulation.[6] He conducted studies in rabbit femur using titanium viewing chambers and observed that the

chambers were inseparably incorporated into the bone and had to be surgically removed, thus leading to the discovery of titanium as a biocompatible material to be used with dental implants. In subsequent studies, Branemark extracted teeth in dogs and replaced them with screw shaped titanium implants, fixed prostheses were then connected after allowing for a healing time of 3-4 months[6].

Branemark further conducted vital microscopic studies on human microcirculation and intravascular behavior of blood cells viewed through an optical titanium chamber in the left upper arm of volunteers. Tissue reaction in these long term studies showed no indication of an inflammatory process. Concluding from his studies that bone anchorage according to principles of osseointegration would work in humans, he treated his first edentulous patient in 1965[6].

Implant shapes have varied considerably and have evolved over time, but currently, the root form implant is the most commonly used implant; but other implant designs such as the blade shaped implant and the staple implant have been used in the past. The blade shaped implant, first described by Linkow, was inserted into the jaw bone after a trench was created for it in the bone using a high speed hand piece. Although osseointegration of up to 82% bone implant contact was reported, due to the traumatic insertion, fibrous encapsulation was a possible complication.[7] Also, due to the shape of the blade implant, if complications were to arise, removal was challenging.

Another type of implant design was the transmandibular or staple implant, consisting of a splint attached to the lower border of the mandible which was inserted through a skin incision extra-orally, and through which two transmandibular screws were

inserted into the mouth.[8] The reported implant survival was 90% after 15 years. However, this design is rarely used today due to the need for general anesthesia.[8]

Osseointegration

Dental implants are considered osseointegrated when there is direct contact between the implant and the supporting bone, lacking periodontal ligaments and therefore the ability to move; the implant is effectively ankylosed. [9]. Branemark describes an implant as osseointegrated when it is "directly connected to living remodeling bone without any intermediate soft tissue component"[10]. Criteria for implant success were described by Albrektsson as[11]:

1. Immobile implant
2. No radiographic radiolucency
3. Vertical bone loss of less than 0.2mm annually following the first year after placement of implant
4. Absence of pain, infection, neuropathy, or paresthesia
5. In the context of above, a successful rate of 85% at the end of 5 years, and 80% at end of 10 year.

Implant materials and surface characteristics

In order to obtain the ideal implant surface, several different materials have been tested (1) metals, (2) polymers and (3) ceramic materials[12].

Metals

Historically, several metals and alloys were considered for fabricating dental implants. Examples include gold, cobalt-chromium and stainless steel. These metals are no longer used as they tend to be surrounded by fibrous tissue (fibrous encapsulation) after insertion and have been replaced with implants that are made of titanium or titanium alloys[12].

Currently, most root form implants are made from commercially pure titanium or titanium alloys. Commercially pure titanium is classified according to its oxide content as grade 1 through 4. Grade 1 contains the least amount of oxygen (0.18%) while grade 4 contains the most (0.4%)[13]. When titanium is exposed to air an oxide layer forms consisting of TiO_3, TiO_2, Ti_2O_3, Ti_3O_4. This layer is 100 angstroms thick, precluding direct contact between the titanium and bone.[14] This layer is passive and stable, and is thought to be the basis for the stable bond between titanium and biologic tissue. Studies to evaluate the biocompatibility of pure titanium were performed using ten cylinders made of polycarbonate covered with a 120-250 nm thick layer of titanium. The cylinder implants were placed in the tibia of rabbits and removed 12 weeks later for observation with TEM. Soft tissue was not present between the implant and the surrounding newly formed lamellar bone. No active inflammation or necrosis was observed, although scattered macrophages were seen. This observation led to the conclusion that titanium is "the material of choice" [15].

Titanium implants can also be classified based on their average surface roughness (S_a) into smooth ($S_a \leq 1\mu m$) and rough ($S_a \geq 1\mu m$)[16]. Smooth surface implants are machine turned and have an S_a of about 0.5-0.9 μm. Different methods for creating surface roughness are available and usually proprietary to the implant company. Bone

to implant contact (BIC) is higher for rough surface implants versus machined implants; rough surface implants also have higher torque removal values and better survival when using shorter implants[17-19]. Implants can have porous sintered surfaces, be blasted with aluminum or titanium oxide, acid etched, or plasma coated with titanium/ceramics. Combinations of the aforementioned procedures have also been utilized[20]. Alloying titanium with other metals allows better mechanical properties to be achieved. Titanium zirconium alloy with 13-17% zirconium shows significantly better mechanical properties such as increased elongation and fatigue strength than pure titanium.[21]

The Osseotite® implant surface is manufactured by Implant Innovations, Inc. (3i) (Palm Beach Gardens, FL). A rough surface is achieved by treating the machined titanium (TI) surface with duel thermo-etching[22]. This method involves immersion in 15% Hydrogen fluoride (HF) and then etching in a mixture of H_2SO_4/HCl (ratio of 6:1) while being heated at 60°-80°C for 3-10 minutes[22]. Examination of the surface under SEM reveals presence of many sharp pits 2μm in diameter[22]. Atomic force microscopy (AFM) examination of the surface shows presence of discrete surface depression areas[22]. Maximum vertical distance in a section analysis was 1070nm[22]. The roughness average (Ra) value of the AFM scanned surface was 226nm and the Root mean square roughness (Rq) value was 286nm[22].

Davies discussed the mechanism of bone formation around Osseotite implant surfaces[23]. He described the process of contact and distance osteogenesis. In distance osteogenesis, new bone is formed around the implant. The surrounding bone provides a source for osteogenic cells that lay down a new matrix. He describes the

blood clot that forms around the implant immediately after insertion, and micro topographic implant surface feature is thought to be responsible for this retention between the blood and surface. This close adaptation results in adaptation of the fibrin clot to the implant surface[23]. The surface roughness parameter increases as follows: machined<Nanotite<Osseotite[24]. A histological evaluation of the Osseotite® implant surface 6 months after placement in the posterior maxilla showed bone to implant contact of 72.38% versus 33.98% for machined surfaces. Lazzara evaluated placement of 429 Osseotite implants in 155 patients in a multicenter study[25]. The implants were surgically placed and subsequently loaded 2.1 ± 0.7 months. The mean time from prosthetic loading to study follow up was 10 months. At the ten month follow up, seven of the 429 implants had not integrated. Six of the non integrated implants were identified prior to loading. The cumulative post-loading implant survival was 99.8%[25]. This study suggested that successful functional loading of Osseotite dental implants was possible at less than 3 months[25]. Another study showed implant survival at 97.1% in platform switched Osseotite implants placed in the anterior maxilla[26].

Polymers

Polymers such as polyethylene and Teflon (polytetrafluoroethylene), have also been studied and are generally stable in the tissues[27]. However, these polymers are hydrophobic and have little adhesion to living cells in spite of their high stability. Therefore, they are not suitable materials for the construction of implants.[27]

Ceramics

Dental implants with hydroxyapatite (HA) surfaces have also been studied. HA is biocompatible and is similar in composition to natural bone.[28] HA-coated implants show an earlier bone response than commercial pure titanium and titanium-alloy implants[28]. A plasma-spraying method has been used to apply HA to the surface of titanium implants[28] . Despite the simplicity and versatility of this method, it has several problems: ease of fracture of the coating layer, vulnerability to bacterial infection, non-uniform coating thickness, and easy degradation in the human body[28]. These inherent problems result in the rapid breakdown of osseointegration and progression of peri-implantitis.

More recent studies have evaluated zirconium as an implant material. Typical titanium implants have a gray color, and in thin gingival biotype, can show through the gingival tissue, leading to poor esthetic outcomes; lack of this gray color is an advantage of zirconium[29]. Multiple studies have evaluated the survival rate of implants ranging from 87.5% to 98%[29]. One such study showed survival rate of 95.4%, but had a high risk of crestal bone loss greater than 2mm during the first year after placement[29]. This material appears promising, but further long term studies are needed to document the performance of such devices in the oral environment[29].

Peri-implantitis

Classification

Peri-implantitis was defined by Albrektsson at the First European Workshop on Periodontology as an inflammatory process affecting tissues around an implant causing pocket formation and the loss of supporting bone; this is a separate entity from peri-implant mucositis, the reversible inflammatory changes in the soft tissues

surrounding a functional implant without any bone loss.[30] Froum and colleagues classified peri-implantitis as early, moderate, or advanced. Early peri-implantitis is described as an implant with probing depths ≥4mm, bone loss <25%, moderate is an implant with probing depths ≥6mm and bone loss of 25-50%; an implant with probing depths ≥8mm and bone loss >50% is classified as advanced. All of these classifications include bleeding and/or suppuration on probing[30].

Etiology/diagnosis

The Sixth European Workshop on Periodontology evaluated peri-implant disease.[31] It was noted that the two most common peri-implant diseases are peri-implantitis and mucositis, both of which are caused by the inflammatory response to bacteria.[31] Peri-implant mucositis occurs in 50% of sites restored with implants, and peri-implantitis in as high as 40% of sites.[31] The following risk indicators were identified: poor oral hygiene, history of periodontitis, smoking, and diabetes. It was concluded that the outcome of non surgical treatment of peri-implantitis was unpredictable, and that the goal of the surgical treatment is access to the implant surface for debridement and decontamination[31]. Due to the inflammatory nature of the process, surface decontamination is paramount. Lack of evidence for additional benefits of regenerative surgery to repair lost bone height around infected implants was also noted[31].

Implant type, abutment type, prosthetic components, implant angulation, proximity of implants, and position of the implants can cause changes in the dimension of the tissue surrounding the implant. Furthermore, they can affect bone resorption and establishment of the biologic width on the implant. This phenomenon is unrelated to

the pathologic disease process of peri-implantitis; thus, not all probing depths greater than 3mm are indicative of peri-implantitis[32].

Microbiology

Surface modification of the machined implants leads to increased surface area for bone contact. However, it may also serve as a reservoir for increased formation of plaque and bio-film on the implant surface. Colonization of the implant sulcus is similar to the natural tooth sulcus, whereby healthy dental bacteria are seen around healthy implants[33]. Facultative, gram positive rods and cocci are seen around healthy implants, however peri-implant microbial flora similar to dental periodontal pockets harbors bacteria such as P. gingivalis, P. intermedia, P. nigrescens, T. forsythia, C. rectus and A. actinomycetemcomitans (especially serotype b)[34]. It was noted that beneficial species and complexes were reduced in diseased versus healthy implants[35, 36].

Management of peri-implantitis

Several authors have evaluated the treatment for peri-implantitis. The treatment should include removal of the biofilm surrounding the implant, decontamination of the implant surface, and correction of the bony defect through either surgical recontouring of the bone or regeneration of the defect and re-osseointegration.[34]. Re-osseointegration was found to be more favorable on roughened surface implants than machined implants[31]. In a systematic review evaluating surgical versus non surgical treatment of peri-implantitis, it was found that both groups had similar clinical

outcomes as far as implant stability, but histological and radiographic outcomes were superior in the surgical group[37].

Subramani conducted a review and categorized surgical peri-implantitis treatment as resective or regenerative. Resective procedures involve removal of the intrabony defect and establishment of positive architecture; this is a favorable approach when there is little bone loss. Regenerative approaches are recommended for moderate bone loss, and much like regeneration around teeth, involves placement of a graft at the site for space maintenance and use of a barrier to contain the bone while preventing epithelial down growth. In both treatments thorough surface decontamination is paramount. In advanced cases of bone loss where surface decontamination is difficult, it is recommended to remove the implant, regenerate the deficient ridge, and then replace the implant [38].

Implant surface decontamination

Albouy evaluated the influence of implant surface characteristics on the outcome of treatment of peri-implantitis[39]. This dog study model created peri-implantitis using ligature ties around the implants. The ligatures were removed once 40-50% bone loss was obtained. Four weeks after the removal of the ligatures, surgical therapy including mechanical cleaning of the implants was performed. Five months after surgical therapy, the dogs were sacrificed and block biopsies obtained. The results showed that the implants with machined surface, titanium grit blasted (TiOblast®) and Sandblasted and acid-etched (SLA®) surfaces, had radiographic bone gain, while the spark anoded (TiUnite®) surface showed additional bone loss after therapy. Resolution of the peri-implantitis was achieved in tissue surrounding the implants

with machined and TiOblast surface. The author concluded that resolution of peri-implantitis is dependent on the surface of the implant[39].

Mechanical decontamination of implant surfaces treated surgically or non surgically have been studied; they include the use of ultrasonic scalers with plastic or carbon fiber tips, stainless steel, titanium or expanded polytetrafluoroethylene (ePTFE) curettes and scalers, air abrasive powders, and pumice polishing cups[38]. Various chemical decontamination methods using local antimicrobials such as chlorhexidine, tetracycline, citric acid, hydrogen peroxide and phosphoric acid have been evaluated and compared in studies and systematic reviews[38, 40-42]. A study comparing decontamination of machined, plasma-sprayed (TPS) and hydroxyapaptite (HA) implant surfaces contaminated with radioactive endotoxin from P. gingivalis was assessed by burnishing implant surfaces with cotton pellets soaked in water, citric acid (CA) or chlorhexidine 0.12%, or treated with an air abrasion system. It was found that CA was equally effective in decontaminating all the implant surfaces studied, and that machined implants due to the lack of surface roughness were more effectively decontaminated than the other systems with the materials studied. It was also noted that air abrasion was consistently the most effective method for removal of endotoxin[40].

Schmage and colleagues looked at options for surface decontamination using 10 different accepted methods for surface decontamination, including use of plastic curettes, carbon curettes, Sonic-Flex clean/prophylaxis brushes, rubber cups, Sonic-Flex clean implant set, Piezon master 400, Prophy Max with Periosoft curettes, Vector system with periodontal curettes, Cavitron Prophyjet, and Er: YAG (key

laser)[43]. The lowest damage to the implant surfaces was found with plastic curettes and Sonic-Flex clean/prophylaxis brushes. Excessive damage occurred with use of Prophy Max with the Periosoft curettes. The best cleaning scores were found with Sonic-Flex clean implant set and the Cavitron Prophyjet with ClinPro Prophy powder on all implant surfaces, while the worst cleaning was noted with carbon curettes. Although several studies have shown superior results using Er:YAG, this study was not able to confirm the results because of the surface alterations and inefficient surface cleaning by the Key laser.[43]

Esposito conducted a systematic review of the literature to evaluate if any single treatment was better in preventing peri-implantitis and the best option for treatment.[44] Esposito concluded that insufficient reliable evidence was available concerning preferred plaque control prevention methods around peri-implant soft tissue or for therapeutic regimens in the treatment of failing implants.[44] Because bacterial contamination of the implant surface is considered the main cause of peri-implantitis, recently, the use of lasers to decontaminate the implant surface has been proposed.[45]

Lasers

The theory of stimulated emission was first proposed by Albert Einstein.[46] In 1960, Theodore Mainman invented the synthetic laser ruby, which he used to create the first laser.[46] The emission occurs when the atom moves from the excited state to the more stable form by emission of a quantum of energy. Laser light is special in that it is monochromatic (one wave length), directional (low divergence from the main beam), and coherent (same phase relationship). The wavelength of light can be delivered in three different modes: continuous, gated pulse, and free running pulse

mode. Continuous refers to the emission at one power level continuously as long as the switch is pressed.[46] Gated pulse has on and off periods, usually in microseconds. Free running pulse mode refers to a very large laser energy being emitted in a short period of time, followed by a relatively long time during which the laser is off. The laser light can be delivered through an articulating arm with mirrors at each end, hollow wave guides, or fiber optics.[46]

Lasers react with oral tissue through photothermal ablation (vaporize, coagulate), photomechanical ablation (disruption of tissue due to shockwave formation), or photochemical effect (use light sensitive dyes).[46] Use of lasers in periodontal treatment has several advantages which include: blocking pain signal transmission, increased production and secretion of endorphins, inflammation reduction due to vasodilatation, accelerated tissue repair and cell growth, improved blood flow, and reduced formation of scar tissue.[46] However, lasers have the potential for increased temperature change at the target site.[47] The threshold of 7°C is the highest thermal change which is biologically acceptable.[47]

Currently, several different types of lasers exist with various wavelengths of light output. CO_2 laser has a typical wavelength (λ) of 960-1,064nm, Nd: YAG has λ=1064nm, Gallium-aluminum-arsenide (GaAlAs) λ=809nm, diode laser with λ=810nm, Erbium: yttrium-aluminum-garnet (Er:YAG) with λ=2,940nm, and Holmium:YAG (Ho:YAG) with λ=2,090nm.[48, 49] Lasers of shorter wavelength 100-300nm (ultraviolet) are able to ionize tissue through a process known as photochemical desorption. Lasers of longer wavelength, 700-10,000nm (infrared)

can produce significant tissue heating; most surgical lasers are in this category, and are considered thermal lasers.[48]

Laser decontamination uses around teeth

Photodynamic therapy uses a dye and light source/laser to activate the dye causing bacterial death. Photoactivated dye technique can be used to disinfect periodontal pockets, root canals, cavity preparations and peri-implantitis.[50] Cobb investigated the use of pulsed Nd:YAG laser either alone or in combination with scaling and root planing.[51] In this in vivo study, 18 teeth with associated periodontal pockets from 8 patients were treated.[51] 5 specimens were root planed and exposed to laser for 3 minutes using the energy setting of 3.0 watt (W) at 20 pulses per second (pps), 2 specimens were root planed, then exposed to 3 minutes of 2.25W and 20pps, 4 specimens were treated by laser for 1 minute at 1.75W and 20pps, and 4 specimens were treated by laser for 1 minute at 1.75W and 20pps.[51] Prior to and after laser treatment, pockets were sampled for periodontal pathogens; then, teeth were extracted and examined under SEM.[51] They found that all treated specimens, regardless of treatment sequence, exhibited some degree of laser induced root surface alteration.

Coffelt suggests use of Nd:YAG at energy density of less than 200J/cm2 in order to prevent irreparable root damage[52]. Although the laser failed to remove all plaque and calculus deposits, the treated calculus deposits were free of the typical surface layer of plaque.[51] A systematic analysis of the use of Nd:YAG laser in the treatment of patients with periodontitis.[53] It was concluded that there is no evidence to support the superiority of the Nd:YAG laser over traditional periodontal therapy.[53]

Laser irradiation on titanium implants

Photo dynamic decontamination was evaluated using toluidine blue on machined surface, TPS, SLA and HA coated implants. It was found that the dye and laser treatment led to destruction of A. actinomycetemcomitans, P. gingivalis and P. intermedia on all implant surfaces studied[54]. Park evaluated the effects of laser irradiation on machine and anodized titanium disks.[55] For the study, 3 types of lasers were used: erbium chromium-doped yttrium-scandium-gallium-garnet (Er,Cr:YSGG), erbium yttrium-aluminum-garnet (Er:YAG), and carbon dioxide (CO_2). Surface changes were noted with both types of titanium surfaces when Er,Cr:YSGG laser was used with more than 3W of power, while surfaces changes were noted with the Er:YAG laser when more than 2W of power was used, but no changes were noted with CO2 laser.[55] Kreisler evaluated CO_2 laser treatment of implant surfaces *in vitro* and concluded that energy setting of 30J/cm did not affect machined implants, and the surface of TPS and SA specimens looked glazed.[49] For HA coated implants, the coating was affected by laser irradiation at 15.2 J/cm.[49] Another effect of laser treatment is the temperature change associated with the procedure. Gemiani evaluated the temperature increase during CO_2 and Er:YAG irradiation on implant surfaces and concluded that after 60 seconds of continuous irradiation with the CO2 laser, a temperature change of 13.9°C was seen, while 60 seconds of irradiation with the Er:YAG produced a 50.6°C temperature increase.[56] CO_2 laser in continuous mode produced a 10°C temperature increase after 36 seconds of use, but the Er:YAG laser produced a much more rapid temperature increase of 10°C in only 10 seconds.

The same study evaluated the use of the pulsed mode versus the continuous mode and concluded that a much lower temperature rise could be achieved in pulsed mode; furthermore, the temperature change could be drastically reduced by using air and water spray. When using the air water spray and pulse mode, only a 5.5°C temperature increase was noted with the CO_2 laser and 1.3°C with Er:YAG laser.[56] A similar study was conducted with ErCr:YSGG laser and CO_2 using sandblasted, hydroxyapatite and acid-etched surfaced implants.[57] Upon applying the CO_2 laser for 60 seconds at 4W continuously, the mean thermal increase on all surfaces was 8.56°C. On sand-blasted acid-etched implant (SBAE) and TPS surfaces, the temprature increase exceeded 10°C after 60 seconds of use. HA surfaces had the least thermal increase at 4.3°C. ErCr:YSGG (pulsed mode at 1.5W, 20HZ, 60 seconds) showed a mean thermal change of 5.02°C with the greatest temperature increase corresponding to the HA (7.50°C) and SBAE (6.7°C).[57]

Nd:YAG Laser use on implants

A study by Kreisler showed that using Nd:YAG on implant surfaces produced alterations on all implant surfaces after 5 seconds of laser use.[49] The scale of damage was proportional to the power setting and was discernible even at the lowest setting possible in this experiment (3.6J/cm). Cracks, melting, and crater formation were the predominant alterations noted.[49] In the case of HA coated implants, the HA coating of the implants was destroyed on a large scale. An increased O/TI ratio on the smooth, SLA, and TPS surfaces was found.[49] Kreisler concluded that Nd:YAG laser

was not suitable for use in decontamination of implant surface irrespective of the power output.[49]

Contrary to Kreisler, Vassalli evaluated the use of Nd:YAG on treatment of peri-implantitis and found that when the pulse energy was kept below 30mJ/cm, there was no change in surface morphology and temperature on all implants when temperature was kept below 37° C using constant air flow. Another study from the same group showed that Nd:YAG laser with similar settings (i.e. 1-1.4 W mean power, 20mJ/cm pulse energy, 50-70 Hz) could induce bacterial ablation. The bacteria tested included A. actinomycetemcomitans, and E. coli.[58] The same group also tested whether the use of photosensitizing agent methylene blue was able to kill bacteria in suspension or agar media. Their results suggest that the use of Nd:YAG in photodynamic therapy is viable in treating bacteria in periodontal pockets or around implants with reduced edema and pus.[59]

II. PURPOSE

The aim of this study was to evaluate the effect of Nd:YAG laser treatment on the Osseotite surface of 3i Biomet implants in order to determine any surface changes due to the laser treatment, and to evaluate if Nd:YAG is a suitable means of decontaminating the Osseotite® surface without damaging the surface.

III. MATERIALS & METHODS

Pilot study:

A pilot study was undertaken to determine the power settings and distance to the implant surface to be used for the main study. Data from the pilot study showed extensive damage to the implant surface when the laser beam was contacting the implant, so for the main study only two distance categories were tested A) 3mm B) 10mm with laser beam perpendicular to the implant surface.

Implants:

A total of 9 Osseotite® implants were used (Biomet 3i implants Palm Beach Gardens, Florida 33410), 4 for the pilot study and 4 for the main study with one control implant. Each implant provided three test surfaces along the long axis of the implant 120 degrees from one another which was marked at the implant collar using a sharpie marker. Twelve implant surfaces were used for the pilot study and 12 implant surfaces for the final study. In the final study, each implant surface also provided 3 threads as separate units of measurement. Table 1 shows the implant surface and

corresponding energy delivered to each implant in the pilot study and table 2 shows the study design from our final study.

Energy Level	Distance	N
0.8 W	Contacting	1
0. 8 W	3 mm	1
0.8 W	10 mm	1
2 W	Contacting	1
2 W	3 mm	1
2 W	10 mm	1
3 W	Contacting	1
3 W	3 mm	1
3 W	10 mm	1
4 W	3 mm	1
4 W	10 mm	1
6 W	Contacting	1

Table 1: Study design for the pilot group.

	Distance	Energy Level	N(Threads)
Implant surface 1,2	3 mm	0.8 W	6
Implant surface 3,4	10 mm	0.8 W	6
Implant surface 5,6	3 mm	2 W	6
Implant surface 7,8	10 mm	2 W	6
Implant surface 9,10	3 mm	3 W	6
Implant surface 11, 12	10 mm	3 W	6

Table 2: Study design for the treatment groups

Procedure

A jig was fabricated to hold and move the implants along a constant path at a constant speed (figures 1,2). Implants were sequentially mounted on the positioning device (jig) using the cover screw or impression post. An electric implant motor (W&H Dentalwerk Bürmoos Austria) with 20:1 ratio head connected to pulley system

attached to a rail was used to move the implants parallel to (contacting) and at a 90 degree angle (3mm,10mm) to the laser beam at the rate of 1.4mm/second. The laser was delivered through 400 micron fiber optic cables; in addition, a 635nm 5mW red diode aiming beam was also emitted through the fiber optic cable to provide operator orientation prior to Nd:YAG delivery via a foot pedal. The laser beam was activated simultaneously as the motor moving the implants. Since the attached implant travelled at standardized rate of 1.42mm/sec past the laser tip, the laser output power (Watt) was to be divided by a factor of 1.42 to provide actual power delivered per mm of implant surface/second. For the purpose of calculating the delivered Watt (W) or Joules (J) the following formula was used: W=(Joules x Hertz)/1.42, total of 0.56J/mm was delivered at the .8W output, 1.42J/mm for 2 watts, and 2.11J/mm for 3watts.

A control implant was mounted in the jig; one side was run past the laser fiber optic tip (contacting) without the activation of the laser to test any damage caused by of the fiber optic tip contacting against the implant surface, and the second side served as the no treatment control. Implants were handled using sterile gloves and sterile forceps, avoiding any contact with the implant surface and only contacting the carrier mount supplied with the implant, or cover screw. Implants were removed from the jig, transported in the manufacture supplied sterile container, photographed and examined under SEM microscope.

Implants were irradiated with a Nd:YAG laser (PerioLase® MVP-7™, Millenium Dental Technologies, Inc.) at power settings of 0.8W, 2.0W, and 3.0W, corresponding to energy frequencies of 40mJ, 100mJ, 150mJ, respectively at 20KHz

frequency and pulse duration of 150μs. The fiberoptic tip of the laser was cleaved after each implant, and the output power was verified. After mounting at 3mm or 10mm a fresh surface was irradiated with either 0.8w, 2w, or 3w delivered through a 360nm fiber optic tip positioned perpendicular to the surface.

Surface analysis

SEM imaging (FEI Quanta 200, Hillsboro, Oregon 97124 USA) and clinical photos taken with a Canon SLR using a macro lens (Canon U.S.A., Inc.) were used to determine surface alterations to the implant surface. Magnifications of 1:50, 1:100, 1:350 and 1:400x was used for SEM evaluation of implant surfaces. The unit selected for measurement was the thread peak; measurements were made using ImageJ software (National Institute of Health. Bethesda, Maryland). For the purpose of calculating altered surface area magnification of 350x was used.

Statistical analysis

The data analyzed from the SEM images was surface area (μm^2) of effect per implant thread; the mean and standard deviation were calculated. Two implant surfaces with 3 representative threads were analyzed using a 3 (power) x 2 (distance) x 3 (thread) repeated measures analysis of variance (ANOVA) with repeated measures (JMP Software, Statistical Discovery™, Cary, NC, USA).

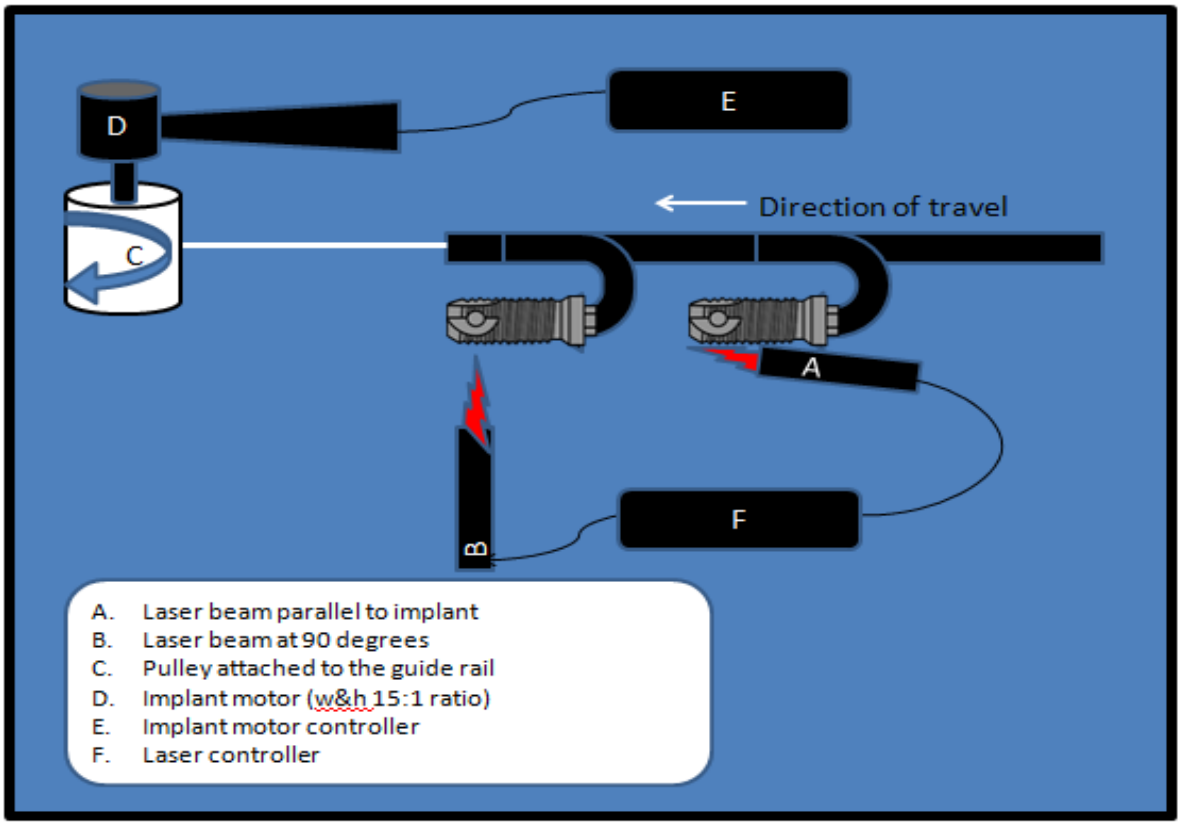

Figure 1: Schematic of the jig fabricated to hold and irradiate the implants

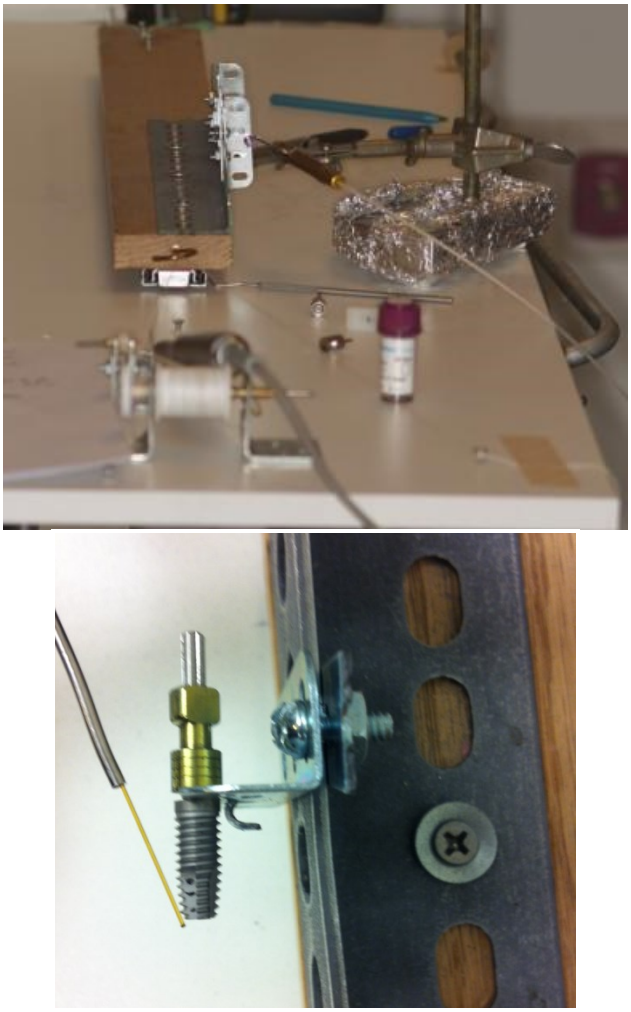

Figure 2: Photograph of the jig fabricated to move the implants in linear motion at a constant speed

Control implant

Clinical photos taken of the control implant shows an intact implant surface without any sign of melting, surface changes, or physical distortion (Figure 2).

Under SEM evaluation, the control implant shows a highly roughened surface with machining marks visible at 350x magnification. There is no evidence of charring, melting, or cracking on these implants (Figure 3). Examination of the surface under SEM reveals presence of many sharp pits 2μm in diameter.

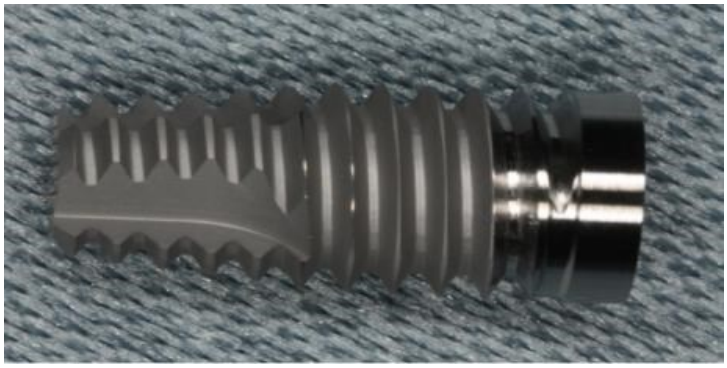

Figure 3: Clinical photo of control implant showing no visible damage

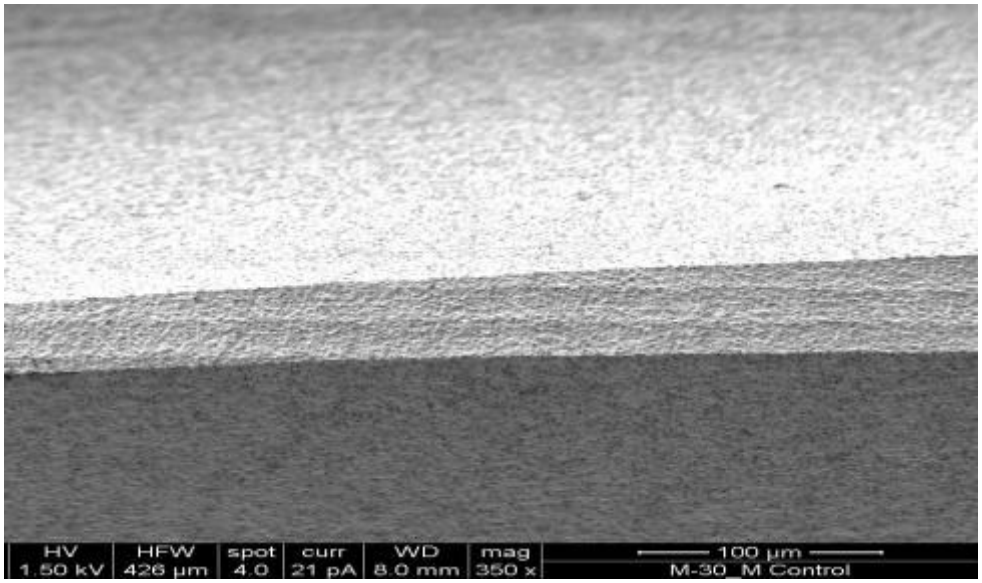

Figure 4: SEM image of control implant at 350x magnification

Pilot study

Clinical and SEM evaluation of the pilot study with laser fiber contacting (parallel) and power setting of 0.4 or 6W (table 1) shows extensive damage, melting, cracking, and surface alteration on implant surface while contacting the laser beam fiber optic tip at 0.4W or 6W (Figure 4,5,6,7, and 8).

During irradiation with the laser, there was visible sparking, expression of smoke, and audible sizzling while the laser beam was activated and contacting the implant surface.

Figure 5: Clinical photo of 0.4W laser while contacting the implant

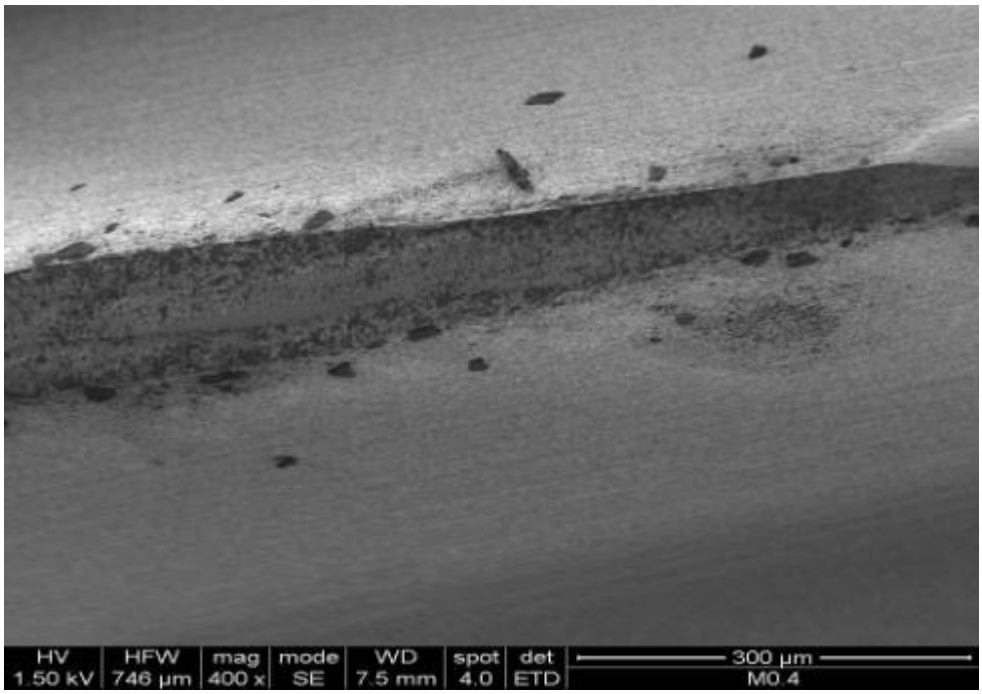

Figure 6: SEM image at 400x magnification after 0.8W delivery contacting (parallel) the implant thread peak


Figure 7: Clinical photo of the effect of 6W laser beam contacting the implant

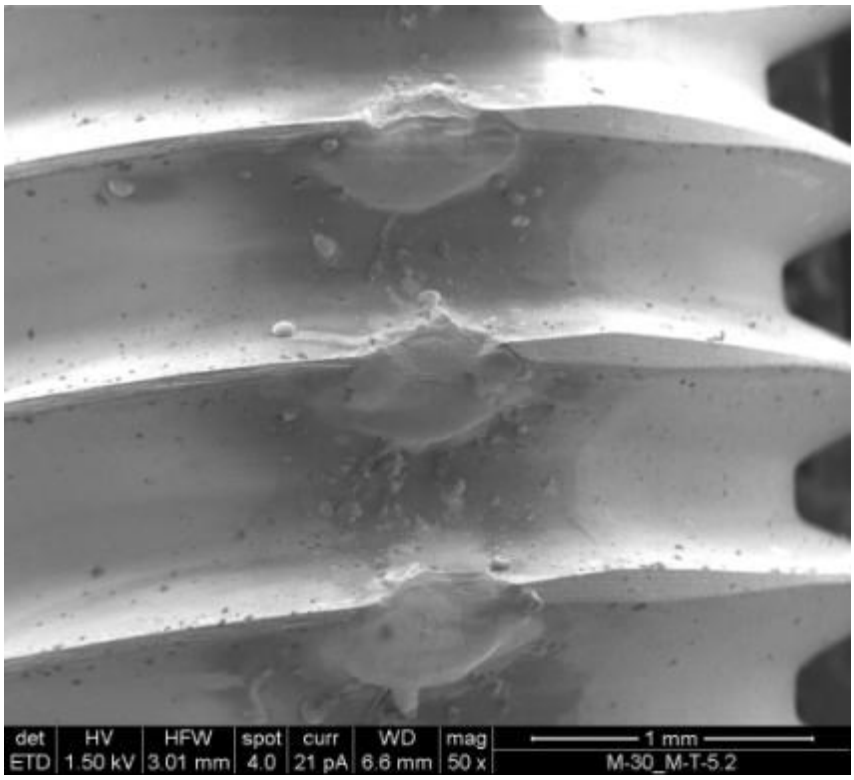

Figure 8: SEM photograph at 50x magnification after treatment with 6W and contacting mode

Study groups
0.08 Energy level
3mm distance

At this distance and energy level, there was no visible damage to the implant surface (Figure 9). Evaluation of the implant surface under SEM microscope at 350x magnification shows melting and charring of the implant surface, mainly at the tips of the threads (Figure 10). Table 3 shows the altered surface area of the implant as recorded at 350x magnification with mean area of 3430 ± 1300 μm^2.

Power	Distance	Thread1	Thread2	Thread3	Mean	st dev
Watt	mm	altered surface area μm^2			μm^2	μm^2
0.8	3	2184	4909	4967	3340.3	1300.3
0.8	3	3052	2909	2021		

Table 3: Area of surface alteration 0.8W 3mm away

Figure 9: Clinical photo of 0.8W at delivered at 3mm distance.

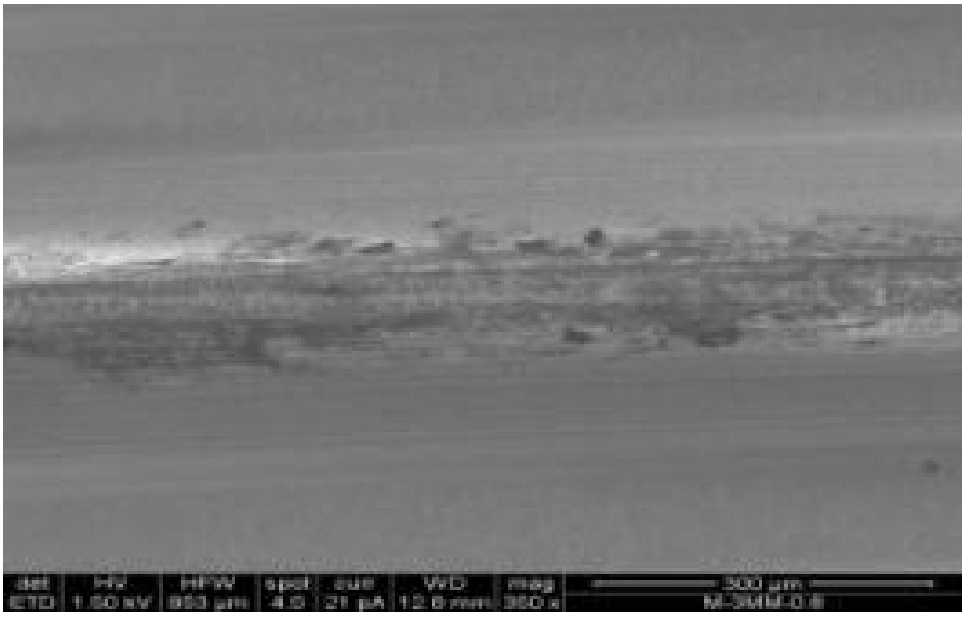

Figure 10: SEM image of 0.8W surface 3mm away at 350x magnification

0.8 W Energy level
10mm distance

At this distance and energy level, there was no visible damage to the implant surface (Figure 11). Evaluation of the implant surface under SEM microscope at 350x

magnification shows melting and charring of the implant surface, mainly at the tips of the threads (Figure 12). Damage to the implant was more diffuse, and there was a wider area of charring. Mean area of alteration was 3765 ± 1096 μm^2. Table 4 shows the altered surface area of the implant as recorded at 350x magnification.

Power	Distance	Thread1	Thread2	Thread3	Mean	st dev
Watt	mm	altered surface area μm^2			μm^2	μm^2
0.8	10	4816	3758	4703	3765.2	1096.7
0.8	10	2008	2976	4330		

Table 4: Area of alteration 0.8W 10mm

Figure 11: Clinical photo of 0.8W delivered at 10mm distance.

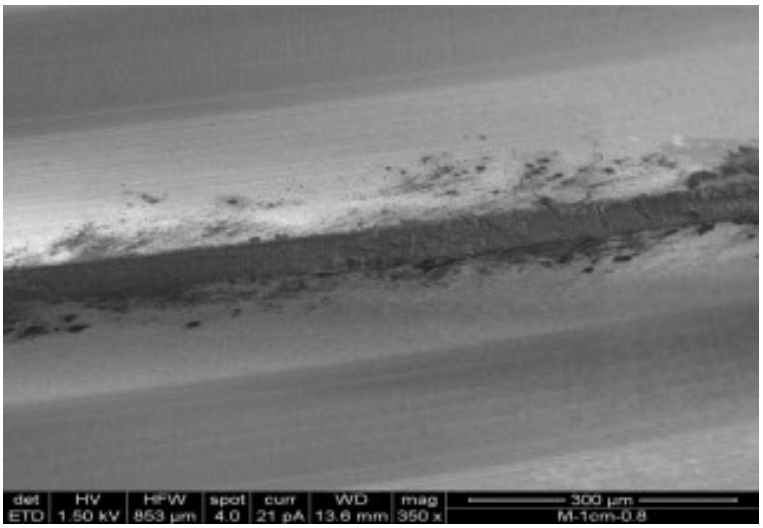

Figure 12: SEM image of 0.8W at 10mm distance viewed at 350x magnification

2.0 Watt energy level
3mm distance

At this distance and energy level, there was visible damage to the implant surface, showing a clear path of the travel of the laser beam marked by surface alteration and bluing of the implant surface (Figure 13). Evaluation of the implant surface under SEM microscope at 350x magnification shows melting and charring of the implant surface. Damage is mainly at the tips of the threads, but faint signs of surface alteration are evident deeper in the grooves. Figure 14 shows the interface between altered area and area of no damage. Table 5 shows the altered surface area of the implant as recorded at 350x magnification.

Power	Distance	Thread1	Thread2	Thread3	Mean	st dev
Watt	mm	altered surface area μm^2			μm^2	μm^2
2	3	4759	6612	4868	4401.8	1316.9
2	3	3142	3140	3890		

Table 5: Area of surface alteration at 2.0w 3mm away

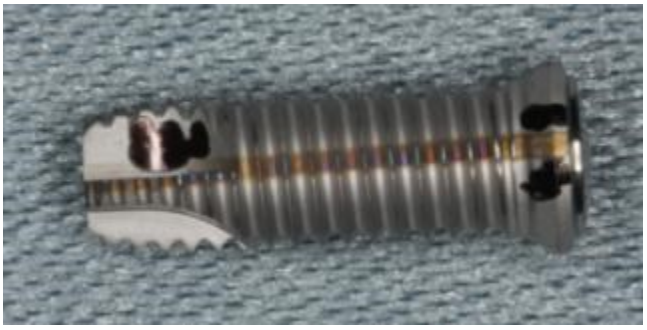

Figure 13: Clinical photo of 2W delivered at 3mm distance

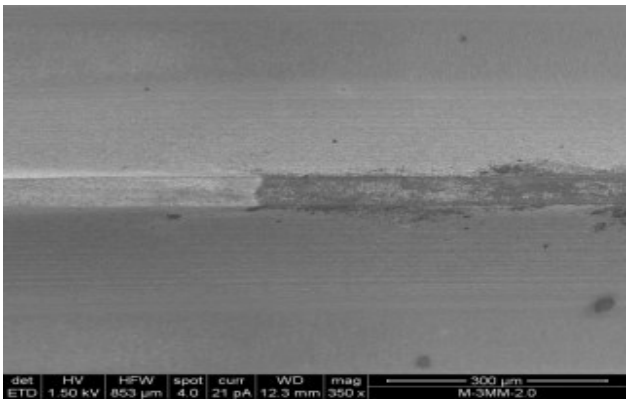

Figure 14: SEM of 2W delivered at 3mm distance at 350x magnification

2Watts energy level
10mm distance

At this distance and energy level, there was minor visible damage to the implant surface as visualized by the naked eye (Figure 15). Evaluation of the implant surface under SEM microscope at 350x magnification shows melting and charring of the implant surface, mainly at the tips of the threads (Figure 16). Table 6 shows the altered surface area of the implant as recorded at 350x magnification.

Power	Distance	Thread1	Thread2	Thread3	Mean	st dev
Watt	mm	altered surface area μm^2			μm^2	μm^2
2	10	3048	2682	4104	3904.3	1028.2
2	10	4773	5346	3473		

Table 6 Area of surface alteration 2W, 10mm away

Figure 15: Clinical photo of 2w delivered at 10mm distance

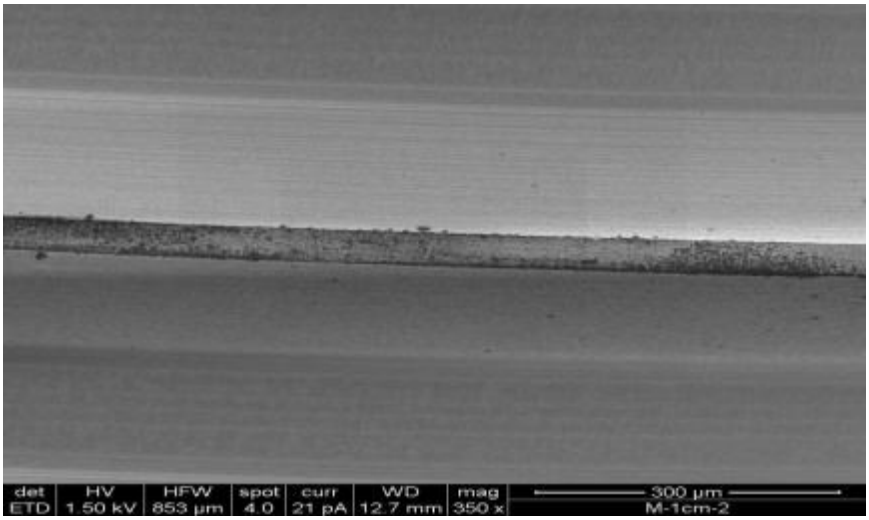

Figure 16: SEM evaluation of 2W delivered at 10mm distance

3 Watt energy level
3mm distance

At this distance and energy level, there was visible damage to the implant surface as visualized by the naked eye, showing a clear path of the travel of the laser beam marked by surface alteration and bluing of the implant surface (Figure 17). Evaluation of the surface under SEM microscope at 350x magnification shows melting and charring of the implant, with damage extending deeper into the implant grooves. Figure 18 shows the interface between altered area and area of less damage

further away from the beam. Table 7 shows the altered surface area of the implant as recorded at 350x magnification.

Power	Distance	Thread1	Thread2	Thread3	Mean	st dev
Watt	mm	altered surface area μm^2			μm^2	μm^2
3	3	4273	3890	3869	3883.2	1080.6
3	3	2519	3081	5667		

Table 7: Area of surface alteration 3W, 3mm away

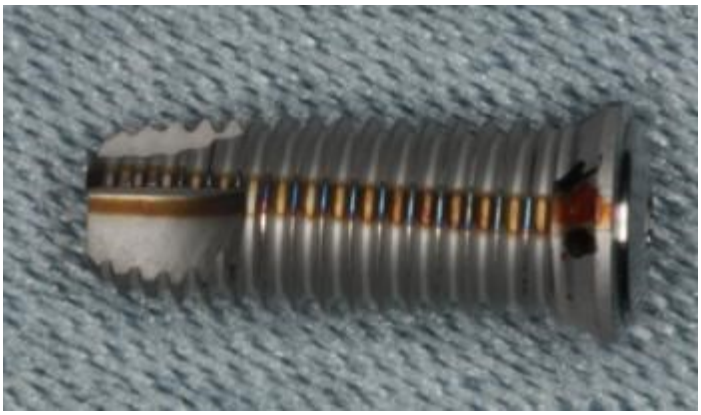

Figure 17: clinical photo of 3W delivered at 3mm distance

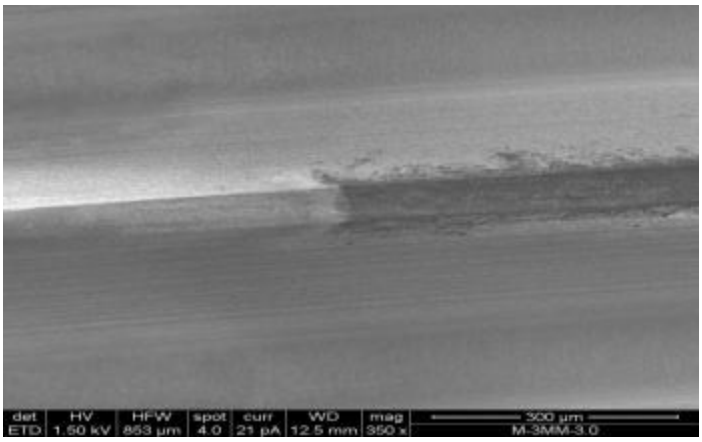

Figure 18: SEM evaluation of 3Wdelivered at 3mm distance at 350x magnification

3W energy level
10mm distance

At this distance and energy level, there was no visible damage to the implant surface (Figure 19). Evaluation of the implant surface under SEM microscope at 350x magnification shows melting and charring of the implant surface, mainly at the tips of

the threads (Figure 20). Table 8 shows the altered surface area of the implant as recorded at 350x magnification.

Power	Distance	Thread1	Thread2	Thread3	Mean	st dev
Watt	mm	altered surface area μm^2			μm^2	μm^2
3	10	2325	2123	1910	3994.2	2123.2
3	10	4990	6628	5989		

Table 8: Area of surface alteration 3w, 10mm away

Figure 19: clinical photo of 3W delivered at 10mm distance

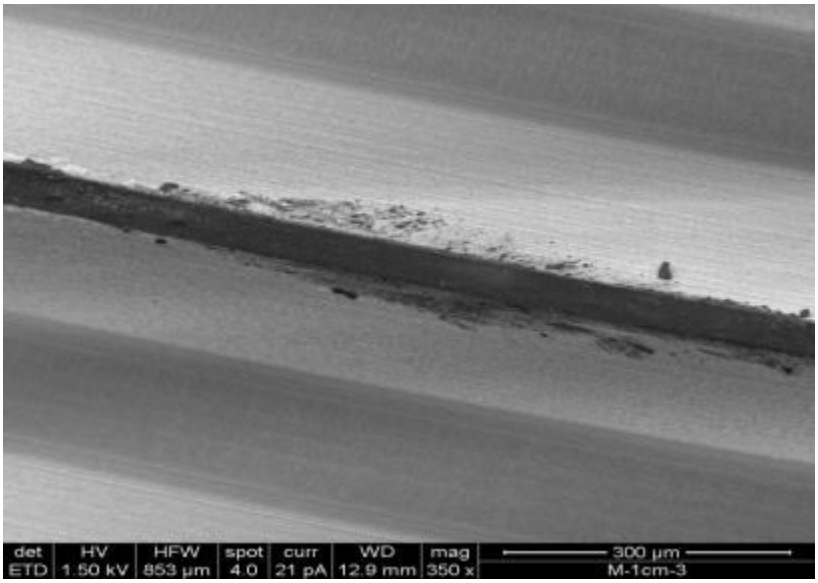

Figure 20: SEM image at 350x magnification 3W delivered at 10mm distance

All implants, regardless of the distance (up to 10mm), and power setting from 0.8watts to 6 watts, showed surface alteration. In the samples in which the fiber optic

touched the implant surface, heavy melting and surface erosion was noted even at the lowest power setting (0.4W). The area of primary damage was closest to the contact area with the laser beam, and showed melting, cracking, and spatter formation. When a non contacting mode of 3mm or 10mm was used, there was a proportional decrease to the amount of physical damage to the implant surface, but regardless of distance, there was visible sparking and interaction between the implant surface and laser beam.

Statistical analysis

The ANOVA showed significant main effect for Watt, reflecting comparable surface alterations at all energy levels, when compared to no treatment, independent of distance (Table 9). Figure 21 shows the plot of the area of surface alterations versus power (W) applied from 3mm or 10mm.

Variable	F value	p value
Watt	0.949	0.005*
Distance	1.23e-5	0.995
Watt*Distance	9.1e-3	0.918

Table 9: Statistical analysis: *indicates significance

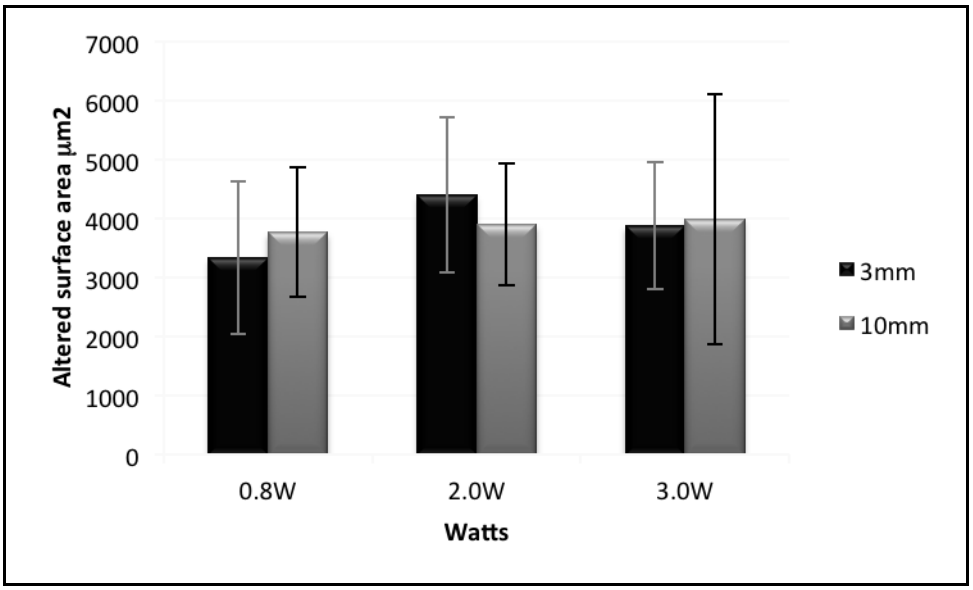

Figure 21: Mean altered surface area of implants at 3mm and 10mm at each energy output.

V. DISCUSSION

As implant designs have evolved, so have the implant surfaces. When the original Branemark implants were introduced, the surface was a smooth machined surface, but as research into osseointegration progressed, it was shown that rougher implant surfaces provided a quicker and higher bone to implant contact. Peri-implantitis and its progression is based on several factors, one of which being the surface topography of the implant. A recent Meta analysis shows the summary estimates for the frequency of peri-implant mucositis were 63.4% of participants and 30.7% of implants, while those of peri-implantitis were estimated at 18.8% of participants and 9.6% of implants. A higher frequency of occurrence of peri-implant diseases was recorded for smokers with a summary estimate of 36.3%[60]. Contrary to previous reports and reviews[61], this systematic review showed that previous history of periodontitis did not significantly increase the occurrence of peri-implantitis[60]. However, with such a high prevalence of peri-implant disease, options to treat the disease have to be explored.

Use of lasers has been proposed by several authors as a viable option to decontaminate the implant surface. Previous authors have evaluated several different types of lasers with different wave lengths and power settings. The aim of this study was to evaluate the effect of Nd:YAG laser treatment on Osseotite surface of 3i Biomet implants in order to determine any surface modification due to the laser treatment, and to evaluate if Nd:YAG is a suitable means of decontaminating the implant surface without damaging the surface.

Our results contradict the claims of Vassalli that use of Nd:YAG with a low power setting of less than 30mJ did not have any morphological changes to the implant surface[45]. Giannelli evaluated the power output setting of 1.4 W, with 20 mJ pulse energy and 70 Hz pulse frequency. Laser irradiation was carried out in non-contact mode (2 mm away from the irradiated surface) for 1 minute and concluded that laser irradiation of LPS-coated titanium disks significantly reduced LPS-induced nitric oxide production and cell activation by the macrophages and strongly attenuated intercellular adhesion molecule-1 and vascular cell adhesion molecule expression, as well as interleukin-8 production by the endothelial cells without any reported damage to the titanium surface[62]. In this study, using the lowest power setting of 0.4W and the laser beam contacting the implant surface produced visible changes to the implant surface at a clinical level, and under SEM evaluation at 400X magnification, clear melting and charring of the implant surface was visible. At higher power settings of 6W and contacting mode, clear melting, cracking and implant damage was visible which could be appreciated in the SEM examination at 350X magnification.

Our preliminary examination with the laser beam contacting parallel to the implant surface showed surface damage at the lowest power setting that we were able to achieve, leading to the conclusion that Nd:YAG will produce surface changes. This observation is in agreement with Kreisler, who noted cracks, melting, and crater formation at any setting (as low as 14.4J/cm^2)[49]. Kreisler noted a finding of appearance of high silicon peaks under SEM examination. Romanos performed SEM examination showing extensive melting in all Nd:YAG laser irradiated titanium discs, with surface damage (loss of porosity and microfractures) at even the lowest power

setting. The studies by Kreisler along with our own demonstrate clear damage to the implant surface, but it is unclear if this surface change has a negative effect on re-osseointegration of the contaminated implant surface.

In order to determine if the Nd:YAG laser can be used in an open flap surgery setting, the laser energy was directed at the implant surface at a 90 degree angle, but not touching the implant. After bench testing, distances of 3mm and 10mm were selected for the remainder of our testing; the selected distances were thought to be achievable in a surgical setting without possibility of touching the implant. Following Giannini's studies on bacterial ablation power settings of 0.8, 2,3 watts were also selected for our main study. Our, results at distances of 3mm and 10mm, showed less damage to the implant surface compared to contacting mode. The damage was proportional to the power (Watts) used, and as the power increased, so did the level of damage to the implant surface. As the distance from laser beam to implant was increased from 3mm to 10mm, less damage could be visualized clinically, but under SEM magnification, melting and charring of the implant surface was still visible. As the distance from the implant was increased, the area of damage was limited more to the peaks of the threads of the implant, suggesting that the metallic implant surface acts like a lightning rod, with the tips of the threads absorbing most of the laser energy. Even at a distance of 10mm, a black raised deposit could be seen on the implant surface. Our study did not evaluate the chemical properties of the deposit, but other authors have suggested this to be a silicone deposit as viewed using energy dispersive spectroscopy[49].

There are several limitations to our current study which might affect the clinical outcome. Our study was in vitro and was performed on dry sterile implant surfaces, whereas the use of the laser in a surgical setting will be around infected implants that are contaminated with saliva and blood. This is important since Nd:YAG laser with the λ=1064nm has an affinity for blood, and pigmented tissue. In our study we did not evaluate any temperature changes to the implant surface, although when handling our pilot implant, we did note a heating of the implant when the laser beam was passed in contacting mode. This thermal change was significant enough to be felt when the implant was handled. Some studies have used air or water cooling during laser treatment around titanium surfaces in order to maintain an acceptable temperature increase. Although we did note a physical change to the implant surface at distances up to 10mm we are not able to determine within the confines of this study if this change has a positive or negative effect on re-osseointegration of the implant surface. There are several studies that evaluate the antimicrobial effects of the Nd:YAG laser, but in many of these studies, power settings and treatment times are not at clinically acceptable levels. For example, in one study, laser treatment was performed at 6watts for 1 minute[63]. There is a scarcity of knowledge on the bactericidal settings for Nd:YAG on titanium surfaces. This is made more difficult because the interaction of the laser with the metallic surface of the implant is very different than with natural teeth, thereby making many of the studies on Nd:YAG and natural teeth irrelevant to its effects on implants[64-67].

The use of photodynamic therapy using a diode laser or Er:YAG have also been documented, and are thought to have a positive effect on decontamination of the

implant surface[49]; thus, further studies are necessary to evaluate if Nd:YAG provides any additional benefits not afforded by diode, or Er:YAG which outweigh the physical damage to the implant surface. Several authors have advised against the use of the Nd:YAG laser to treat peri-implantitis, or for second stage surgery of submerged implants[49, 68, 69] but further research is needed to determine if the surface alterations are a negative or positive change.

VI. CONCLUSION

Within the limits of this in vitro study, the Osseotite® surface appears particularly susceptible to damage when exposed to the Nd:YAG laser, even at comparatively low energy (0.8W, 40mJ, 20 Hz/ mm of implant surface/second); however, it remains unclear whether such surface alterations adversely impact osseointegration.

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