Comparison of volumetric changes at implant connector among three different types of abutment after cyclic loading

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A Thesis Submitted in Partial Fulfillment of the Requirements
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การเปรียบเทียบการเปลี่ยนแปลงทางปริมาตรที่รอยต่อของรากเทียมระหว่างหลักยึดสามชนิด
หลังจากการรับแรงกระทำแบบวัฏจักร

น.ส.ปิยวรรณ จิรายุสกมล

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ปิยวรรณ จิรายุสกมล : การเปรียบเทียบการเปลี่ยนแปลงทางปริมาตรที่รอยต่อของรากเทียมระหว่างหลักยึดสามชนิดหลังจากการรับแรงกระทำแบบวัฏจักร. (Comparison of volumetric changes at implant connector among three different types of abutment after cyclic loading) อ.ที่ปรึกษาหลัก : รศ. ทพ.ประเวศ เสรีเชษฐพงษ์, อ.ที่ปรึกษาร่วม : อ. ทพญ.ดร.วรีย์รัตน์ เจิ่งประภากร

งานวิจัยนี้มีจุดประสงค์คือเพื่อเปรียบเทียบปริมาตรของรากฟันเทียมไททาเนียมระหว่างหลักยึดที่แตกต่างกันสามประเภทหลังจากการโหลดแบบไซคลิกโดยใช้การยึดฟันเทียมคร่อมรากเทียมชนิดระดับกระดูก และหลักยึดไททาเนียม(กลุ่ม 1: Titanium abutment) ,หลักยึดทองคำ (กลุ่ม 2: Gold abutment) และหลักยึดเซอร์โคเนีย (กลุ่มที่ 3: Zirconia abutment) มาผ่านการจำลองรับแรงกระทำแบบวัฏจักร โดยเครื่องทดสอบยูนิเวอร์แซล (E1000, อินสตรอน) ทั้งหมด 1 ล้านรอบ, โหลดในแนวแกนดิ่ง ด้วยแรง 100 นิวตัน ความถี่ 15 Hz ปริมาตรของรากฟันเทียมไททาเนียมวัดโดยเครื่องวิเคราะห์ความหนาแน่น (AccuPyc II) และประเมินพื้นผิวโดยการเปรียบเทียบภาพ micro-CT หลังจากการโหลดแบบไซคลิก ความแตกต่างของปริมาตรรากฟันเทียมและเปอร์เซ็นต์การสูญเสียปริมาตรระหว่างหลักค้ำยัน 3 ประเภท ผลการทดลอง แสดงให้เห็นว่าค่าเฉลี่ยเปอร์เซ็นต์การเปลี่ยนแปลงปริมาตรของรากฟันเทียม ไม่มีความแตกต่างอย่างมีนัยสำคัญทางสถิติระหว่างกลุ่มที่ 1 (ไทเทเนียม :mean ± SD;9.3982±0.2128) และกลุ่มที่ 2 (ทอง: ค่าเฉลี่ย ± SD;7.6164±1.9165) แต่มีความแตกต่างกันอย่างมีนัยสำคัญทางสถิติ ระหว่างกลุ่มที่ 1 (ไทเทเนียม :mean ± SD;9.3982±0.2128) เทียบกับกลุ่มที่ 3 (เซอโคเนีย: ค่าเฉลี่ย ± SD; 17.3302±0.5560) และกลุ่มที่ 2 (ทอง: ค่าเฉลี่ย ± SD; 7.6164±1.9165) เทียบกับกลุ่มที่ 3 (เซอร์โคเนีย : ค่าเฉลี่ย ± SD 17.3302±0.5560) ภาพจาก Micro CT ไม่พบความแตกต่างที่เกี่ยวข้องสำหรับแพลตฟอร์มรากฟันเทียมที่สัมพันธ์กับความเสียหายของโครงสร้างบนหลักยึดไทเทเนียมและหลักยึดทองคำในทางกลับกันรากฟันเทียมบนตัวค้ำยันเซอร์โคเนียแสดงความเสียหายเพียงบางส่วนที่ด้านในของแท่นรากฟันเทียม โดยสรุปรากฟันเทียมที่มีตัวค้ำยันเซอร์โคเนียแสดงให้เห็นการเปลี่ยนแปลงเชิงปริมาตรของรากฟันเทียมส่วนใหญ่

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Piyawan Jirayusakamol : Comparison of volumetric changes at implant connector among three different types of abutment after cyclic loading. Advisor: Assoc. Prof. PRAVEJ SERICHETAPHONGSE, D.D.S.,M.S. Co-advisor: WAREERATN CHENGPRAPAKORN, D.D.S.,PH.D.

The aim of this study was to compare the volumetric change of dental implant among three different types of abutments after cyclic loading.Thirty dental implants (4.1x10 mm., Straumann®) were used .All of them were inserted in acrylic resin block with usual surgical protocol to gain primary stability and be able to retrieve for volumetric measurement later. Ten of each dental implant connected with Titanium abutment (Group1: Variobase abutment Straumann®), Gold abutment (Group2: UCLA abutment Straumann®) and Zirconia abutment (Group 3: Care abutment Straumann®) consecutively. Then all specimen was submitted to cyclic loading ($1x10^6$ cycles, axial load, 100N, 15 Hz) at 30°angulated. After loading, the volumetric of the dental implants were measured by true density analyzer (Accu Pyc II).The volumetric changes of dental implants among three abutment types were examined by one-way analysis of variance (ANOVA) followed by Bonferroni post-hoc analysis. A *P* value <0.05 was considered statistically significant. Results showed the dental implant for group 3(mean ± SD; 17.3302±0.5560) was significantly different from Group 1 (mean ± SD 9.3982±0.2128) and group 2 (mean ± SD; 7.6164±1.9165) .In summary, dental implants with zirconia abutments showed the greater of volumetric loss than the other groups.

Field of Study:	Esthetic Restorative and Implant Dentistry	Student's Signature ....
Academic Year:	2022	Advisor's Signature ....
		Co-advisor's Signature ....

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Piyawan Jirayusakamol

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CHARPTER I

INTRODUCTION

A dental implant is composed of three components: the implant body, the abutment, and the prosthetic part. The term "two-piece implants" refers to the abutment and implant body, which are permanently attached through an abutment screw. Two-piece implant systems were developed to overcome the drawbacks of one-piece implants and to allow for adjustment of the prosthesis's angulation following implant insertion. However, due to the two-component nature of the connection, a microgap will exist between the implant and abutment interface (2). The lack of stability at the implant-abutment interface is the primary cause of mechanical complications with dental implants. Occlusal load is the cause of micromotion at the implant-abutment interface. Microleakage and screw loosening are more likely to occur at a higher level of micromotion. The discoloration of the gingiva induced by the titanium implant abutment is one of the drawbacks of placing an implant in the esthetic zone; in certain situations, the thickness of soft tissue may be insufficient to cover the titanium's color, resulting in the presence of gray at the gingiva. This circumstance may come as a surprise to both the dentist and the patient, particularly in the esthetic wwaspect. (3). The usage of cast gold alloy and zirconia abutments is frequently employed in the aesthetic field to address the titanium abutment issue. The cast gold alloy and zirconia abutment, on the other hand, have excellent mechanical qualities. The material interface discrepancies might be caused by the varying amounts of micromotion caused by implant and abutment wear. Numerous research have been conducted on the wear of titanium implants in conjunction with zirconia and titanium abutments. It was discovered that the zirconia abutment influenced titanium implant wear by presenting a black hue at the zirconia abutment following cyclic loading. Previously conducted studies assessed the design, fit precision, material composition, screw shape, degree of friction, preload, and anti-rotational elements. Numerous studies have been conducted to evaluate zirconia abutments in comparison to conventional titanium abutments. There are no studies that assess the volumetric changes at the implant connector between three distinct types of abutment following cyclic loading, which prompted the start of this investigation.

STATE OF THE PROBLEM

RESEARCH QUESTION

Are there any volumetric losses of the internal titanium implant surface among three types of abutment materials after cyclic loading.?

RESERCH OBJECTIVE

A comparison of the volumetric changes of internal surface in dental titanium implant to different material abutment after cyclic loading.

HYPOTHESIS

There is no difference in the volumetric changes of internal titanium implant surface and three types of abutment material after cyclic loading.

CONCEPTUAL FRAMWORK

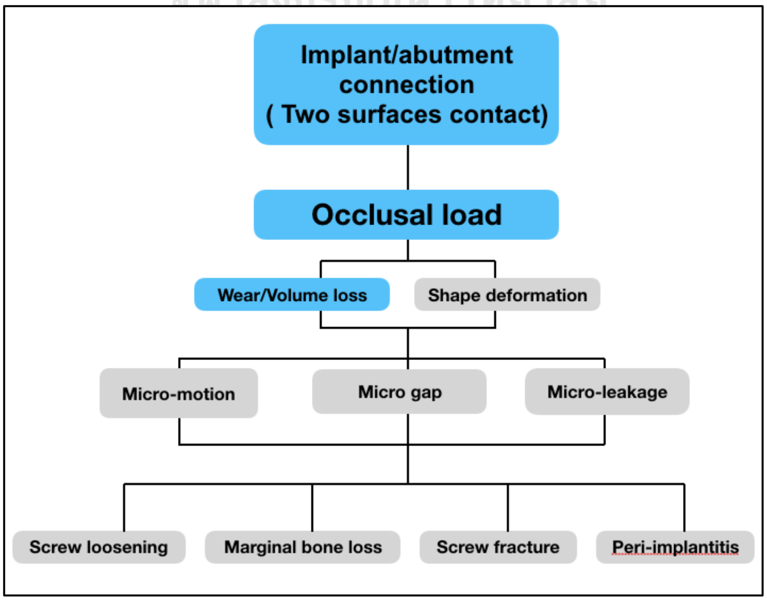


Figure 1. Conceptual framework

(I) Keywords

Dental implant, Gas pycnometry, Titanium, Zirconia ,Gold ,Abutment

EXPECTED BENEFITS OF STUDY

The outcomes of this study may provide useful information regarding the potential wear among different implant abutment and implant connector

CHARPTER II

REVIEW LITERATURE

DENTAL IMPLANT

Dental implants have grown in popularity as a means of replacing missing teeth in modern dentistry. The anterior maxilla is the most crucial location for esthetic success in the oral cavity (4). Due to the great exposure of this area, referred to as the esthetic zone, implant restoration presents complications. Cosmetic problems are frequently seen, particularly in patients with a thin tissue biotype. It has been demonstrated that the underlying titanium casts a shadow through the tissue, leaving the marginal gingiva black and discolored (5). To mitigate titanium's blue-gray hue, the abutment material should be more closely matched to the tooth color. Due to its white look, excellent strength, and biocompatibility, zirconia is a popular material for implant abutment. However, when the zirconia abutment is removed following occlusal function, the black band surrounding the implant-abutment contact is always visible. Titanium tattoos have been described and documented because of the increased wear between zirconia and titanium abutments (6). The increased wear of the titanium implant at the interface with the harder zirconia material has been suggested to increase implant complications ranging from the destruction of the anti-rotational component of the connection, pre-mature screw loosening, and implant failure, to the released metallic particle inducing local soft tissue inflammation and entering the circulatory system, resulting in heavy metal toxicity. (7) Titanium Alloys for Implants Historically, titanium has been used to create endosseous implants (8). Titanium has established itself as the gold standard because to its superior strength-to-weight ratio and biocompatibility (9). Titanium's corrosion resistance is derived from the oxides that form on its outer surface. Oxides inhibit elemental dissolution and promote the deposition of biological molecules, allowing bone to develop close to the surface, a process known as osseointegration. Titanium grade 4 cpTi has the highest oxygen content (0.40 percent). There is a titanium group that is the strongest and has been used for dental implants since 1965. Due to the micro-rough surfaces, they have been reported to promote bone-implant contact, leading in increased osseointegration. The ultimate tensile strength of 550MPa, on the other hand, may be insufficient. Implants with a small diameter and internal connections have a relatively thin interface wall in modern designs. Manufacturers became aware of narrow cpTi implants failing owing to fracture and began developing stronger alloys (10). Due to its enhanced strength-to-weight ratio, titanium grade 5 (TiV, Ti6Al4V) is the most frequently utilized alternative alloy in implant dentistry today. It exists as a dual alpha-beta phase that is strengthened by a 4% addition of the beta stabilizer vanadium, while the alpha phase is stabilized by a 6% addition of aluminum (11). TiV has an ultimate tensile strength of 864 MPa after heat treatment, which along with its corrosion resistance makes it an excellent choice for dental implants. Due of TiV's strength, it is employed in a variety of dental implant components, including abutments and screws. TiV's high strength and increased hardness (296 VH) make it resistant to fracture but also make roughening the surfaces necessary for osseointegration more challenging (10, 11).

TITANIUM ABUTMENT

Titanium is an exceptional metal since it is extremely biocompatible with both hard and soft tissue. Titanium is resistant to corrosion in salt chloride solution, which makes it an ideal material for human implants. It boasts the lowest density of any metal and the highest strength-to-weight ratio of any. Titanium quickly produces oxides, which play a role in the osseointegration process. Titanium has been successfully utilized in implant dentistry for over fifty years and is still the industry standard.

Titanium is found in two distinct atomic crystalline states.

1. Titanium that has not been alloyed or is commercially pure has a hexagonal close-packed (HCP) or alpha atomic structure. Titanium turns into a body-centered cubic (BCC) or beta atomic structure at elevated temperatures, however this structure can be sustained at normal temperature by alloying with molybdenum or vanadium. Commercially pure titanium has a higher resistance to corrosion. Due to titanium's exceptionally high reactivity, commercially pure titanium (alpha) oxidizes instantly when exposed to air. The interstitial oxygen and nitrogen concentration of commercially pure titanium has an effect on its strength. (12) 2. Beta titanium alloys have the highest strength values of all titanium alloys, but they produce less oxides

and are therefore less biocompatible. These alloys are strengthened not only by the cubic crystalline structure centered on the body, but also by the beta-stabilizing elements alloyed with the titanium. Additionally, beta alloys can be heat treated to achieve even higher strengths (6). Due to their great strength and light weight, beta alloys are frequently employed in industrial and technical applications. Due to their high strength to elastic modulus ratio, they are also used in orthodontic applications. Titanium alpha-beta alloys comprise both crystalline and amorphous structures and display intermediate characteristics. One alpha-beta alloy, Ti-6Al-4V, has found limited application in dentistry for implant and abutment manufacturing. The strength values of this alloy are greater than those of commercially pure (alpha) titanium. However, it is less biocompatible and more difficult to generate a micro-textured surface, which is beneficial for dental implant osseointegration. The American Society for Testing and Materials (ASTM) classifies titanium alloys into 38 categories, with grades one through four regarded to be economically pure. Commercially pure titanium (alpha) grade four is frequently utilized in the manufacture of dental implants and abutments due to its excellent flexural strength

(550 MPa) and biocompatibility. The hardness of commercially pure titanium grade four is 258 VH (13).

GOLD ABUTMENT

Since 1932, gold alloys have been utilized in dentistry. These alloys are classified as mild, medium, hard, and super-hard. It was rapidly discovered that gold alloys (containing less than 65 percent gold) rusted excessively quickly. This issue was resolved in 1948 with the addition of palladium. Later in the 1950s, platinum was added to gold alloys. This resulted in a decrease in its growth and an increase in the connection of metals and ceramics. Gold has been a well-known metal for hundreds of years. Due to its qualities, which include a specific weight of 19.3, a melting point of 1062°C, a boiling temperature of 2600°C, ductility and plasticity, and thermal and electrical conductivity, it is often employed in routine dental treatment, mostly as a substructure for prosthetic restorations. The UCLA abutment (14) can be used in a screw-retained configuration. This can be accomplished by casting the abutment into the superstructure (15). When considering the bone resorption pattern and the position of the opposing tooth, the UCLA abutment can be linked directly to the implant fixture. This process is advantageous for producing the emerging profile, and gold alloy was chosen for this application. However, as the price of gold has risen dramatically, other minerals with economic advantages are emerging as viable substitutes. (15, 16, 17)

The design of the abutment allows the fabrication of the restoration directly to the implant fixture, bypassing the transmucosal abutment cylinder. This technique is valuable in overcoming the problems of limited interocclusal distance, interproximal distance, implant angulation, and soft tissue response. Another major advantage of the UCLA abutment is the of improved esthetics.

The subgingival placement of the restoration not only helps with interocclusal distance limitations but also provides improved esthetics. Beginning the restoration in a more apical position not only allows the emergence profile through the soft tissue to be more gradual and natural in appearance but it could also be in porcelain instead of the usual titanium cylinder. (1)

The literature has indicated that castings consisting of at least 50% gold should not present an adverse galvanic response when connected to titanium. The fit cannot be visually inspected, adequately checked with an explorer, or even verified with radiographs because light discrepancies would not be discernible. If the relationship of the implant fixture analogues in the master cast is slightly incorrect, the improper fit of the casting may go unnoticed. Thus, all UCLA abutment restorations on multiple implants must be made in separate segments, one for each implant, and each checked microscopically against an implant fixture analogue to evaluate the individual fit.

The UCLA abutment initially solved problems of limited interocclusal space by eliminating the use of the transmucosal abutment cylinder and the gold alloy cylinder. This approach was most notably beneficial for partially edentulous patients and also useful in the fabrication of some overdenture tissue bars where it was critical to have a low- profile bar. The UCLA abutment allows contours of the restoration to be altered to compensate for the closeness of the implants. The subgingival portions of the restoration can be made narrower than the manufactured abutment cylinders to allow seating of the restoration and oral hygiene access.

ZIRCONIA ABUTMENT

All-ceramic restorations have been used to replace metal restorations for esthetic reasons. They maintain gingival color more comparable to the natural one than the other metal restoration (9). Zirconia is one type of the oxide ceramic widely used in both anterior and posterior restorations due to its excellent properties, high toughness, strength, fracture toughness and biocompatibility. Zirconia is the crystalline oxide (ZrO_2) ceramic form of metal zirconium. It was first identified in 1789 by Martin Klaproth, but wasn’t investigated for biomedical applications until the 1960’s. In 1969, Helmer and Driskell first described its use in the ball for total hip replacement operations. Since the discovery of transformation toughening, zirconia has been under considerable investigation with aims for further biomedical applications(**2**).

3Y -TZP is the most common form of zirconia used in the dental industry because the yttria stabilizes the highest percentage of the strongest tetragonal form of zirconia. The mechanical properties of 3Y-TZP are strongly correlated to the grain size. It can be categorized into three forms: monoclinic (M), tetragonal (T) and cubical (C). Three crystalized forms of Zirconia are divided by varying temperature. Pure zirconium represents the monoclinic phase at room temperature which is stable up to 1170^{O}C. Between 1170^{O}C-2370^{O}C stable tetragonal phase can be detected and above 2370^{O}C, the cubical phase is exhibited (10). If a cubic or tetragonal zirconia sample is then allowed to cool back down below 1170 °C. The transformation will reverse and it will revert back to the monoclinic structure, which is accompanied by a 4% expansion. The significant increase in volume upon cooling causes pure zirconia breaking apart at room temperature. (2)

Alloying pure zirconia with additional oxides such as CaO, MgO, CeO_2, or Y_2O_3 allows stabilization of both cubic and tetragonal phases at room temperature, preventing the ceramic to crack propagation during cooling. (2)

The high-fracture toughness is unique to zirconia ceramics and is due to a mechanism to prevent cracks known as transformation toughening. As a crack propagates, stresses become concentrated within the matrix causing a stress-induced transformation phase change. The resulting internal 4% expansion generates internal compressive forces oppose the stress form the crack, slowing its growth and increasing the toughness of the material. (**3**)

The introduction of computer aided design and computer aided milling (CAD/CAM) zirconia has been used to fabricate anything from crowns, bridges, and full-arch restorations to posts, implants, and implant abutments(**3**). Due to its high hardness, Zirconia restoration can be fabricated by using computer-aided design/computer aided manufacturing (CAD/CAM) procedures. This process, known as soft machining allows for faster milling with less wear to the milling components. Following the milling, the zirconia is sintered, resulting in a roughly 25% shrinkage that the CAD system must compensate for (**2**).

Custom CAD/CAM zirconia abutments have become increasingly utilized in implant dentistry. (2) Traditionally, implant abutments have been made from titanium (TiV), and until CAD/CAM technology was implemented, custom abutments were fabricated from cast gold. Although cast gold custom abutments are still fabricated, most custom abutment today are manufactured with CAD/CAM technology from either titanium or zirconia (4). Zirconia is often the material of choice when esthetics are of concern. The white material has been documented to reduce the grey shadow of titanium in patients with thin tissue types. (5)

However, zirconia's strength and toughness, its use has come into question after reports of increased complications, such as abutment fracture and screw loosening. (6) Additionally, reports suggest that the mismatch in material properties between zirconia and titanium results in increased wear at the implant abutment interface. (7) It has also been suggested in the literature that these common complications associated zirconia abutments may be due to the design, manufacturing, and sintering processes and not due to the raw material itself.

PROPERTIES OF DENTAL ABUTMENT

In implant restoration therapies, it is necessary to have features such as adequate fracture toughness, suitability for intraoral conditions, and survival in order for the materials to be successfully identified. For proper material selection, the mechanical properties of the material and suitability for the case should be carefully examined.

The fracture of prosthetic components has different clinical consequences, depending on the component fractured and the location of the fracture. The fracture strength of abutment must provide resistance to functional loading(**8**, **9**). Abutment screw fractures are associated with inadequate screw tightening, screw loosening, improper occlusion concept, premature occlusal contacts, parafunctional habits, cervical misfit of the prosthesis and consequent fatigue of the screw material, fatigue character and yielding strength of the screw material, and fabrication failures.

The physical properties and the design of screws and restorative components dictate their fracture strength and failure mode. Abutment screws can be made from a variety of materials The study was reported that mean load of titanium abutments is statically higher than zirconia abutments (**10**). Also, the thickness and angulations of abutment materials can affect the fracture resistances. The literature was comparing fracture resistance of the abutments that different thickness and individual zirconia. They were reported that while thickness of restoration wasn't significantly different, angular individual zirconia abutment was showed lower fracture resistance(**8**).

IMPLANT ABUTMENT CONNECTION

Two-piece implant systems, consisting of the implant and abutment, are

widely used in dental restorations. The advantage over a one-piece implant system is that it allows the implant to be unloaded during the bone healing phase and provides the benefit to adjust the prosthetic angle depending on the abutment selected (angled or straight) for placement on the implant. A disadvantage to a two-piece implant system is the resultant microgaps or spaces that exist along the implant- abutment interface when the abutment is seated on the implant and connected via the abutment screw. Under loading conditions, these spaces permit rotation and micromotion of the abutment and can lead to screw preload reduction, screw loosening, bending, or fracture. Various connections have been machined to improve the fit between the dental abutment and implant.

Biomechanics has long been an interest in implant dentistry. The most common mechanical complication is screw loosening of implant-supported restorations(11). and founded in the single tooth implant. There was 45% of screw loosening in external hexagonal connection. On the other hand, the internal connection designs were found the incidence of screw loosening around 8 %.; however, the problem still remains (12). The screw loosening can be affecting the long- term success rate of implant. Due to the micromotion at the implant abutment interface were increasing dramatically when the screw loosening.

The stability of the implant-abutment interface is required for long-term implant success. The implant-abutment connection is under constant load in the oral cavity. Although it is very difficult to study micromotion clinically, it has been quantified in lab studies and is suggested that it has been suggested occurrence may cause screw loosening and consequently wear and destruction of the implant abutment interface(**13**). The degree of misfit in the connection doesn't appear to have a direct correlation to micromotion and quantitatively, the movements of titanium vs zirconia abutments are documented to be similar(**13**). The force on the screw known as preload is responsible for holding together the implant abutment connection. After a screw is tightened, the threads undergo an elastic deformation that acts to hold the parts together with friction. The torque that a screw is tightened to should correlate with the yield strength of any particular alloy to maintain the clamping force(**14**). Metal fatigue and wear as a result of small masticatory oscillations may cause settling of this joint, which will result in clinical loosening(**14**).

There are two types of abutment and implant connection: external connection and internal connection. The connection between abutment and implant are via a screw. The problems associated with abutment screw as loosening or fracture are the most common complications on implant rehabilitation. Khraisat., et al. found that the complication of abutment screw was lower with an internal connection(**15**).

During occlusal loading in implant restorations, the area around head of the abutment screw is the area with highest torque stress. Similar occlusal forces create screw fractures in metal and ceramic abutments, but occur screw deformation in metal abutments (**16**).

Internal connection implant designs have moved away from external butt joint connections that placed significant force on the screw itself. Newer tapered internal connections reduce the stress on the screw and improve the mechanics at the implant abutment connection by instead transferring the lateral forces to the walls of the implant(**17**). Based on the principles of Morse taper, 8 degree tapered internal connections also aim to create a so-called cold weld between abutment and implant alloys(**18**).

These principles work very well for titanium implant-titanium abutment connections because the metallic properties for elastic deformation, however principles of Morse taper create internal stresses in ceramics that may be responsible for the increase fracture rate of zirconia abutments(19).

Internal connections may reduce screw loosening and improve mechanics, but it Even with precision fit components, inevitable microgaps are still present between implants and abutments(20). The microgap while present prior to loading then becomes larger with cyclic loading as wear occurs. (21). These gaps are present with both zirconia and titanium abutments and have been shown to harbor bacteria. The resultant microleakage in and out of the microgaps have been proposed to induce inflammation leading to peri-implant pathology In addition to becoming a reservoir for bacteria, corresponding with micromotion, these microgaps allow a constant flow of saliva and wear products into and out of the implant abutment interface that may potentially increase the wear at the junction(20).

Titanium implant and zirconia abutment interface

Ceramic implant abutments have been used successfully as support for anterior single crown since the 1990's(**22**). The use of zirconia as an abutment material was introduced in 1997 to improve esthetics but maintain the strength of titanium. Since then, zirconia abutments have become increasingly more prevalent along with the progression of CAD/CAM technology, which made design and fabrication assessable and achievable without great effort(**23**). The interaction between zirconia abutments and titanium implants has been investigated previously, but the literature is extremely limited. Studies have focused singularly on either the mechanical or corrosion aspects of the interaction.

The first mechanical wear related article was published in 2003 and highlights the visual observation of displaced titanium debris from an implant that was noted on a white zirconia abutment after cyclic loading(**24**) The visualization of similar phenomenon occurring clinically has sparked interest due to the potential biological and mechanical implications. In 2011 a pilot study directly investigated the wear at the implant abutment interface associated with zirconia abutments(**25**). Zirconia abutments were placed in titanium abutments and cyclically loaded for 1,000,000 cycles. The specimens were examined at periodic intervals during the testing in an effort to track wear progression. During the examinations the interface was evaluated and the surface area of the wear was quantified using digital photography and scanning electron microscopy (SEM). Based on the surface area, it was determined that zirconia abutments generated 8.3 times greater wear than titanium abutments. Additionally, the rate of wear of zirconia abutments showed a peak at 250,000 cycles, after which it continually decreased.

In 2012, a similar study was performed except micro-CT was added to better visualize the specific patterns of wear that occur at the implant-abutment interface(**26**). Based on the surface area, this study also concluded that zirconia abutments cause significantly more wear than titanium abutments. Although insightful, these studies only report surface area and not the volumetric loss of material. In addition, these studies did not consider the electrochemical interaction that occurs in the oral environment.

Titanium, as described above, has been used as the gold standard for dental implants due to its biocompatibility, however, it is not completely inert to corrosive attack(27). Titanium is a passive metal primarily due to the oxide layer that spontaneously forms on its surface. When this passive film is disrupted, the underlying raw metal becomes exposed and susceptible to corrosion(27). This process typically occurs in cycles: the oxide layer is damaged and removed, corrosive attack occurs, and the surface recovers, reforming a new protective passivation layer(28) (29). Many studies have demonstrated the corrosive potential of titanium in biological systems(29). In fact, there are numerous reports that show the electrochemical degradation of titanium in the presence of saliva(30).

Microgap, Micromotion and Microleakage

1. **Microgap**

Most common complication after implantation is marginal bone loss around the connection on implant and abutment. There are many factors relating to the marginal bone loss includes surgical trauma, peri-implantitis, occlusal overload, microleakage, biological width and implant anatomy on the crest area(**31**). The influences of microgap and micromotion existing between the implant and the abutment-interface on marginal bone loss.

The implant abutment interface microgap, defined as the microscopic space between implant and corresponding abutment. The microgap between the titanium abutment and the titanium implant is smaller than that between the zirconia abutment and the titanium implant. Moreover, the implant abutment interface microgaps of zirconia abutments increase significantly when torque values less than those of manufacturer recommended values are applied.

Additionally, the implant abutment interface under cyclic loading increases and becomes close with time, thereby achieving a metal–to–metal cold welding (**32**). Currently, except for one type of taper connection that is totally fixed by 1.5° Morse taper and a large contact surface of the implant and the respective abutment (**33**). All other connections need a certain preloaded screw to achieve and maintain the close connection of Implant abutment interface.

MICROMOTION

Even though accurate construction of the implant and abutment reduces micromotion in the implant abutment connections, the present manufacturing procedure cannot eliminate micromotion during chewing between the abutment and the implant. The micromotion of the implant abutment interface consists of micro-abrasion, micro-shift, and micro-rotation of the abutment in relation to the implant. In general, the micromotion size spans from 1.52 m to 94.00 m. (14). According to the implant abutment connection design point, butt joint connections have a tendency to fret, while taper connections are prone to spin. Additionally, the oral environment and fluids impact the wear process to some degree. Three primary sources of microgap creation have been identified: occlusal stress during physiological function, manufacturing tolerance, and micromotion between the implant–abutment link. According to reports, various kinds of abutment connections cause varying degrees of micromotion. Two prominent kinds of abutment connections are the conical and the butt-joint, with the butt-joint accessible in at least three distinct shapes: hexagonal, octagonal, and trilobed. (34)

MICROLEAKAGE

After connection of the two components and before to loading, the microgap size varies between 0.1 and 10 m. This size may expand during cyclic loading. However, the majority of oral bacteria have a breadth of 0.2–1.5 m and a length of 2–10 m. (35). Therefore, bacteria and endotoxin may readily enter the implant internal cavity via the microgap, resulting in biomaterial exchange between the implant internal cavity and the peri-implant oral environment. Microgaps interact with micromotion to produce mechanical damage. Mechanical damages include fretting wear, adhesive wear, and screw loosening (36). Fretting wear refers to microfracture and chipping between the implant-abutment contact, while adhesive

wear is the plastic deformation of the implant-abutment interface (21). In general, for the majority of two-piece implants, the abutments should be attached through a screw with the required torque value. Through the implant-abutment contact, both implants and abutments will transmit occlusal stresses from a prosthetic suprastructure to the surrounding bone tissue. Nonetheless, an implant-abutment interface connection with inadequate margin fitness might create unintended fast stress, resulting to screw loosening during mastication. In addition, Sahin et al. (37) revealed that a significant implant abutment interface microgap led to a high degree of microleakage and a low removal torque value. As long as practicable, the removal torque value should be equal to or greater than the tightening torque value. Reduced removal torque values indicate that screws are susceptible to loosening; specifically, the implant abutment contact microgap will favor screw loosening by producing microleakage. During chewing, the implant-abutment interfaces of all two-piece implants demonstrate chipping and plastic deformation, indicating that fretting wear and adhesive wear occur. Blum and colleagues (21) demonstrated that particles were either lodged in the layer linking the surfaces of the implant-abutment contact or suspended inside the microgap. Depending on the location of the implant-abutment contact and the implant system, the size and shape of wear particles varied. The sizes generally varied from 2 to 30 m. They are given in a variety of shapes, including flat and spherical. In the meanwhile, plastic deformation was seen to varying degrees in all tested implant systems. When a zirconia abutment is connected to a titanium implant and the two components work together, the deformation energy tends to further distribute to the component with a low Young's modulus, namely the implant (34)

SCREW LOOSENING

Inadequate biomechanical design of the prosthetic reconstruction or occlusal overloading is caused by screw loosening. It has also been reported that, with screw-retained abutments, the abutment loosening occurs frequently. Loosened screws may cause costly complications, such as screw fractures and fracturing of the framework (34) .

These are 2 main mechanisms of screw loosening for implant-supported restorations, as follows:

1. Excessive bending on the screw joint and settling effects. When the load is larger than the yield strength of the screw, there are created the bending forces created on a single-tooth restoration and a plastic permanent deformation of the screw will happen when there is loss of tensile force in the screw stem. Reduced contact forces have been shown to occur between the abutment and implant then the screw joint loosens more easily.

2. The other mechanism of screw loosening is based on the smooth of 2 surface contacts. Even a carefully machined implant surface is slightly rough when viewed microscopically. Because of this microroughness, no 2 surfaces are completely in contact with one another (35).

GAS PYCNOMETRY

Pycnometry is a traditional method for estimating the density of porous (excluding closed pores) and nonporous (regular or irregular shaped) substances. A pycnometer is used to measure the volume of the solid phase displacement fluid without taking into account the void volume (porosity) of the sample. (36) The displacement fluid might be a gas or a liquid, such as water (water pycnometer) (gas pycnometry). Helium is the most often employed gas owing to its inertness and tiny atomic size, which allows it to readily permeate the (open) pores of the sample; hence, the solid volume can be precisely calculated. A gas pycnometer consists of a sample chamber and an expansion chamber. The true volume of a porous or nonporous substance is measured by altering the gas pressure in a chamber with a known volume. Typically, the helium gas pycnometer consists of two chambers with predetermined volumes (determined by earlier calibration) that are joined by an expansion valve. The sample (with a preset mass) is placed in one chamber, while the gas expansion takes place in the other room (the reference chamber). Gas pycnometry is a common analytical technique. It has been used on a wide range of substances, including cereal seed meteorites, polymeric fibers, coal samples, dental composites, granules, plastic films, and silica aerogels. A helium pycnometer was used by Xu Yang et al. to quantify the absolute volume change of cement pastes during the early-age hydration process. On the basis of the trends of absolute volume change and heat flow, the absolute volume change of cement pastes in the early-age hydration process was separated into four stages (dissolution period, induction period, quick response period, and stable period) and coupled with conventional techniques. (37) Dental composite polymerization shrinkage was determined by W.D. Cook. A gas pycnometer was used to estimate the volumes of specimens before and after photopolymerization, allowing for the calculation of the total volumetric shrinkage. (38). Combining X-ray computed micro-tomography (CT) and He-gas pycnometry, Valentin Robin (39) designed and evaluated a technique for measuring linked porosity in unconsolidated subsurface sands.

CHARPTER III

MATERIALS AND METHODS

RESEARCH DESIGN

This study was an in vitro experimental study. The intervention of this study was found the volumetric change of titanium implant and three difference abutment material after cyclic loading.

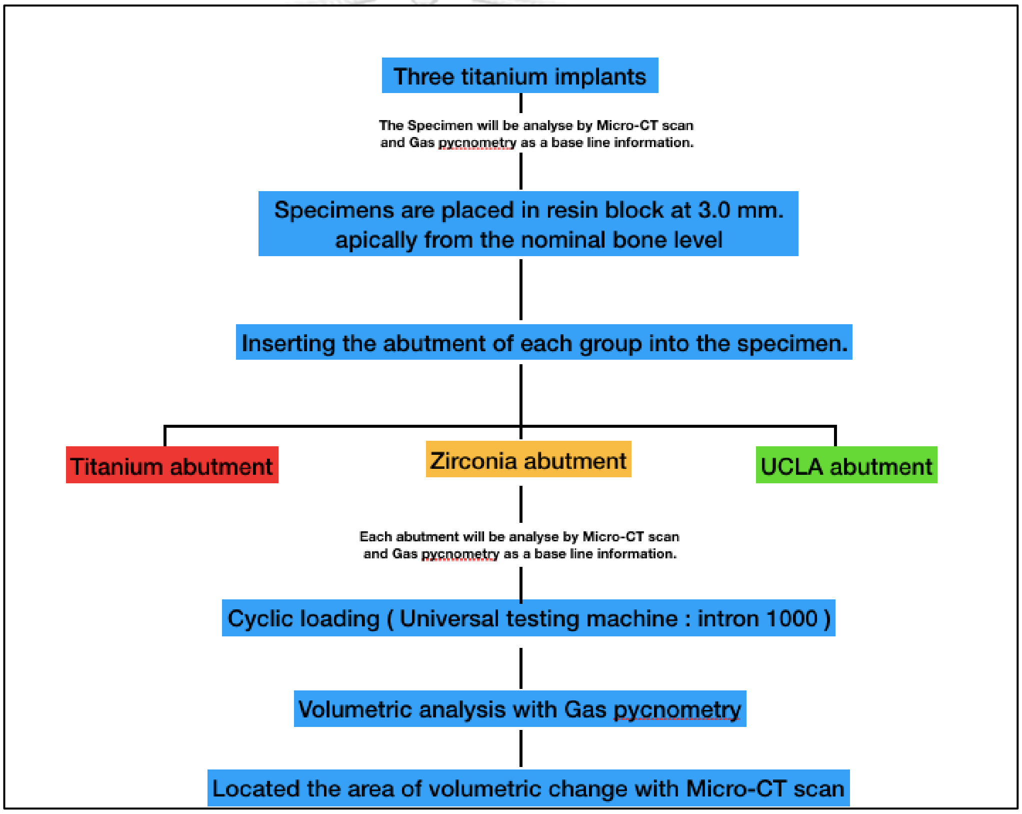


Figure 2. Diagram of the research design

SAMPLE SIZE DESCRIPTION

The sample size was determined using the means and standard deviations derived from previously published publications. Using the G power application, the computation was completed. Based on 5% Type I Error and 80% research power, the calculated sample size was thirty specimens. Due to the sample size calculation requiring a minimum of 10 specimens per category. Therefore, 10 specimens each group should be used in this investigation (n=10). In this investigation, there were three experimental groups, therefore the total number of specimens was thirty.(36).

MATERIALS

- An internal connection implant (BL Ø4.1 x 10mm SLA, Straumann)
- Titanium abutment, Zirconia abutment, UCLA abutment (Gold alloy type 3)
- Epoxy resin block (5 x 10 mm.)
- Testing base 30 °

METHODS

Baseline examination

- All titanium implants and abutments were analyzed using X-Ray computer Tomography scanning and True density analyzer AccuPyc II.

Specimens' preparation

An internal connection implant (BL Ø4.1 x 10mm SLA, Straumann) was placed into the resin block for testing. Each implant was embedded into the resin block, in accordance with DIN EN ISO 14801:2016 standards. Three types of abutments were secured to the implants, titanium abutments (Variobase abutment, Straumann), gold abutment (UCLA abutment, Straumann) and zirconia abutments (Care abutment, Straumann). Ten samples of each abutment type were tested. The abutments were torqued to 35 N/cm as recommended by the manufacturer.

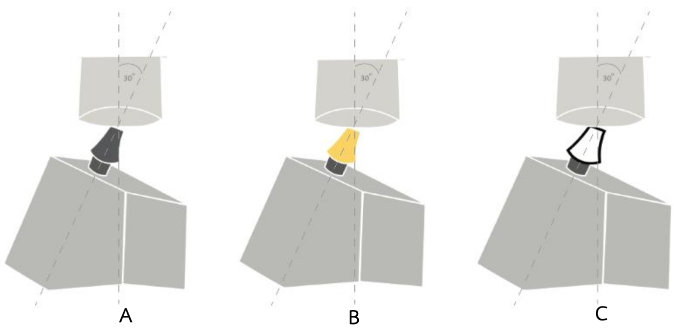


Figure 3. Group A: Ti/Ti , Group B : Ti/Gold Group C Ti/Zr

Mechanical testing of specimens

Figure 4: sample preparation after cyclic load

The specimens are loaded into the testing base using a loading jig, which maintains their position throughout the process. On the loading platform of the compressive machine, the loading jig and the specimen were positioned (Universal testing machine, Instron) The loading platform was sloped at an angle of thirty degrees with respect to the plane that was horizontal. This inclination was utilized to do off-axis loading in order to imitate the way that forces operate on an implant, and it was set at an angle of 45 degrees. In a universal testing machine, the specimens were loaded with a force of cyclically loaded forces with 1x 106 cycles at frequencies of 15 Hz (ISO 14801:2016), 100 N. (UTM). Following the completion of the cyclic loading, the samples were taken away from the testing platform, and the abutment was then unscrewed from the implant.

Sample preparation for Gas pycnometer testing

After removing the sample from the acrylic resin, it was cleaned by rinsing it with water and then being subjected to steam in order to remove any remaining acrylic residue. Finally, the sample was dried using air blowers before being sent to the laboratory for the gas pycnometry test and the titanium implant was analyzed by using the X-ray computer Tomography Scan and True density Analyzer AccuPyc II .

Methodology

Using data from a prior study(37) on dental implants with an alpha of 0.05 and test power of 0.8, the sample size was computed. There were three sets of thirty implants with a diameter of 4.1mm and a length of 11mm. Group1 concerning Titanium abutment. Ten dental implants were fitted with titanium abutments (Group1). Ten implants were fitted with gold abutments (Group2). Ten implants have Zirconia abutments attached (Group3). All specimens were secured to a loading jig within the testing platform. The loading jig and specimen are put on the compressive machine's loading platform (Universal testing machine, Instron). In a Universal Testing Machine, specimens were loaded with a force of cyclically loaded forces with 1x 106 cycles at 15 Hz (ISO 14801:2016), 100 N. (UTM). Each specimen was treated to a total of one million load cycles to mimic occlusion movement over a period of five years. (38) Following the conclusion of the loading test, the specimens were removed from the base. After cyclic loading, samples are extracted from the testing platform and the abutment is unscrewed from the implant.

Prior to affixing the abutment to the platform, the implant volume (in cm3) was determined. The True Density Analyzer (AccuPyc II) was utilized for the volumetric analysis of dental implants. After the mechanical cycle, the volume of the implant was analyzed using the same method.

The specimens were scanned using a high-resolution micro-CT equipment (Bruker Skyscan 1173, Kontich, Belgium) with the following parameters: voxel size of 13.73 m, 100 kV, 100 A, 360 of rotation, 0.5 mm of Al filter, 0.7 of rotation step, 250 ms exposure, and ImageJ software for image reconstruction.**DATA ANALYSIS**

Using the Shapiro-Wilk test, we determined whether or not the data were normally distributed. Levene's test was used to determine whether or not the variances were homogenous. The F-test and one-way analysis of variance (ANOVA), followed by Bonferroni post-hoc analysis, were used to compare the three different kinds of abutments in terms of implant volume and the percentage of volume loss. The statistical analyses were carried out using SPSS Statistics for Windows, version 22.0 (IBM, Armonk, New York, United States), and a P value of less than 0.05 was regarded as being statistically significant.

CHARPTER IV

RESULT

DENTAL IMPLANT VOLUME

Group (N=10/Group)	Mean ± SD (Preload volume /cm^3)	Mean ± SD (Post-load volume /cm^3)	Mean ± SD (volume loss /cm^3)	Mean ± SD (% volume loss /cm^3)
1. Dental implant/Titanium	0.1785±0.0007[a]	0.1617±0.0004[a]	0.0168±0.0044[a]	9.3982±0.2128[a]

abutment				
2. Dental implant / Gold Abutment	0.1758-0.0008^{a}	0.1624±0.0005^{ab}	0.0134±0.0005^{ab}	7.6164±1.9165^{ab}
3. Dental implant/Zirconia Abutment	0.1760±0.0011^{a}	0.1455±0.0007^{c}	0.0305±0.0011^{c}	17.3302±0.5560^{c}

Same lowercase letter indicates no statistically significant difference between the groups ($p>0.05$)

Table 1. Mean (Standard Deviation) of preload volumes, post-load volumes, volume loss and percent volume loss according to the three different abutment types.

Results of one-way ANOVA followed by Bonferroni post-hoc analysis comparing volumetric of dental implant at the various abutment types are shown in Table 1. The mean percent volume loss of dental implant in all groups are shown in Figure 4. At the mean preload volume of dental implants for all groups were not significantly different.

At the mean volume loss of dental implant for group 3 (Zirconia: mean ± SD; 0.0305±0.0011) was significantly higher than group 1 (Titanium: mean ± SD; 0.0168±0.0044) and group 2 (Gold: mean ± SD; 0.0134±0.0005).

At the mean percent volume loss of dental implant for group 3 (Zirconia: mean ± SD; 17.3302±0.5560) was significantly higher than group 1 (Titanium: mean ± SD 9.3982±0.2128) and group 2 (Gold: mean ± SD; 7.6164±1.9165).

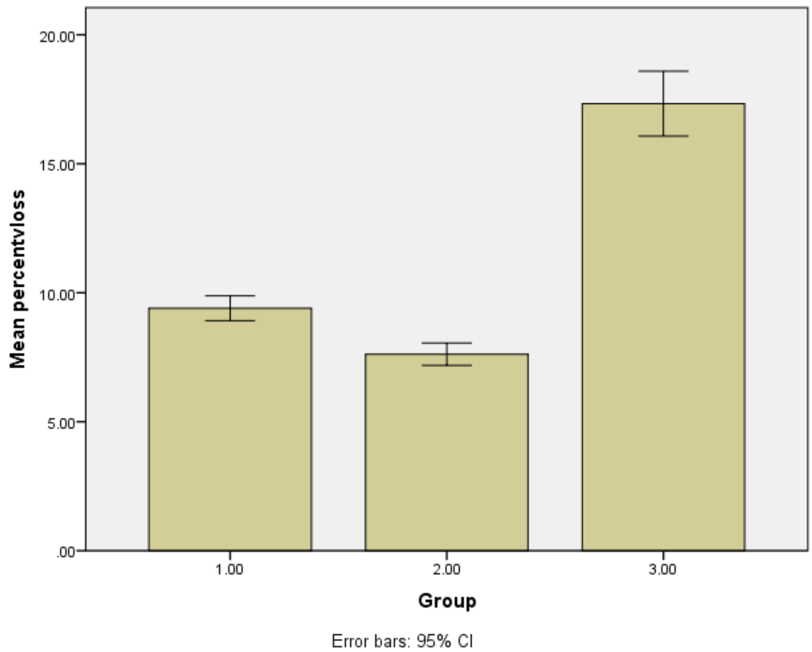
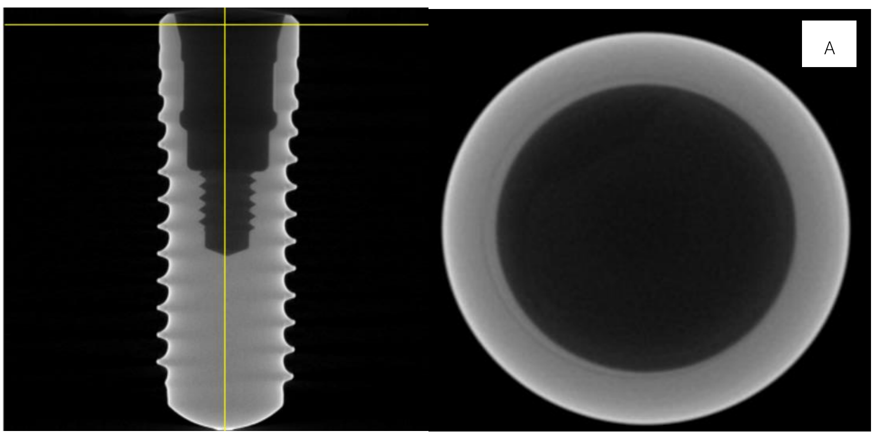


Figure 5 Mean percent volume loss after cyclic load for each group



Base line: Titanium implant before cyclic loading

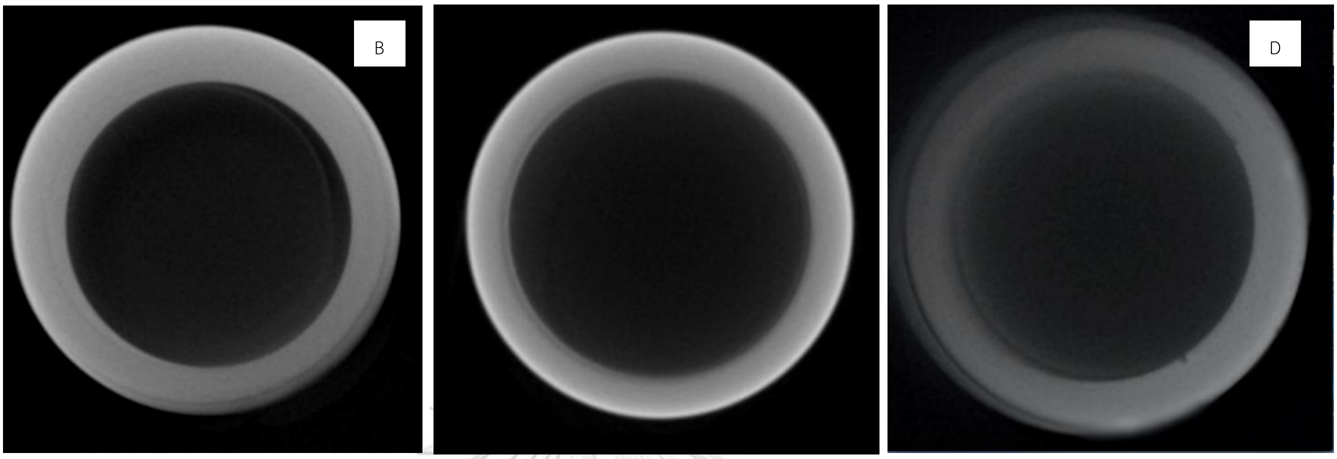


Figure 6 Micro CT photograph, A: Dental implant, B: Group 1 dental implant with titanium abutment, C: Group 2 dental implant with UCLA abutment, Group D: dental implant with Zirconia abutment.

The qualitative analysis performed with Micro CT showed no relevant differences for the dental implant platforms in relation to damage of the structures on Titanium abutment and gold abutment. On the other hand, dental implant on Zirconia abutment shows some damage on the inner of dental implant platform.

CHARPTER V

DISCUSSION AND CONCLUSION

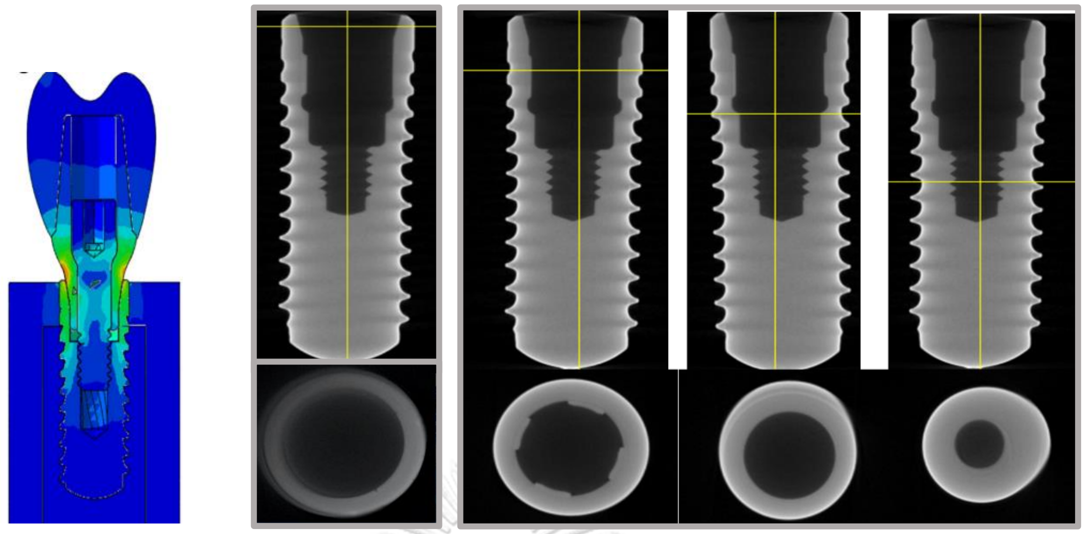
This in vitro study compared the volumetric change of dental implants composed of Titanium, Gold, and Zirconia abutments. Because the percentage of volumetric loss of a dental implant attached to a zirconia abutment was greater than that connected to a titanium or gold abutment following cyclic loading, the null hypothesis can be accepted.

According to earlier research on the wear pattern of dental implants and abutments made of various materials, zirconia abutments cause more wear when attached to titanium implants because of their higher Mohs hardness. (37) The coupling contact has been observed to be harmed by mechanical overload or a high insertion torque when zirconia abutments are seated on an external hexagon implant, in part because zirconia has a much higher hardness than titanium. (38)

However, neither a higher implant wear rate nor a decreased mechanical failure load of the Zirconia abutments was seen in the current study. Uncertain factors may have contributed to the superior zirconia result, and more research on the fabrication process is needed to confirm the geometric precision of the engaged abutment connection.(8)

Our study's findings are in line with those discovered in earlier research, which analyzed the wear that occurred at titanium-on-titanium and titanium-on-zirconia interfaces after implant/abutment assemblies were subjected to thermocycling and mechanical stress. Our findings are consistent with those discovered in earlier research. This was done so that a conclusion could be drawn regarding which material was the most wear resistant. After using zirconia abutments, titanium implants began to show signs of wear. Scans performed using a scanning electron microscope (SEM) revealed that zirconia particles had been transported to the implants.(39) In addition, Klotz et al.(**25**) indicated in their study that a titanium implant that was attached to a titanium abutment showed a lower wear rate when compared to a zirconia abutment. This was due to the fact that the interface materials shared similar qualities. The findings of the current investigation are consistent with the findings of them, who observed that titanium abutments had demonstrated a lower rate of wear.

The geometry of the internal connection has an effect on the amount of movement that the abutment experiences. The connections that utilized Friction-Fit and CrossFit displayed the least amount of horizontal and rotational movement respectively. CrossFit required a significantly lower pull force to dislodge the abutment from the implant when it was subjected to a vertical force, whereas Friction-Fit demanded a significantly higher pull power (50). The fact that we conducted our experiment using the identical cross-fit design for the abutment connection raises the possibility that the design did not have an impact on how the stress was distributed across the different groups.



A B

Figure 7. A. Finite element showed the stress concentrated in abutment and implant connection (Straumann bone level) B: Micro ct of dental implant in group 3. from our study

Concentration of Internal Surface Stress on Fixtures. Typically, the inside surface of the fittings is subjected to a greater concentration of stress than other areas. In the previous study, the bone level implant was found to transfer loads to the internal abutment-implant connection, whereas stress concentrations at the abutment-implant junction may have increased the likelihood of component loosening or implant fracture(40). In relation to our investigation, the micro-CT reveals surface wear at the abutment-implant connection's higher portion. While other areas of the dental implant were not found to be worn, the area of wear was detected by micro-ct. (Figure 6)

Cook et al. utilized a noncontact approach called a gas displacement pycnometer to determine the volume changes in composite materials during polymerization in the dry state in 1999.(41) Pycnometry is a traditional method for estimating the density of porous (excluding closed pores) and nonporous (with regular or irregular forms) substances. Pycnometer is employed measuring the solid phase displacement volume fluid without considering the sample's empty volume (porosity). It is possible for the displacement fluid to be a liquid, such as water (water), or a gas flow meter (gas pycnometry). Helium is the most abundant element gas because it is inert and easily permeates (open) pores. Due to the small size of the sample's atoms, the solid volume can be precisely measured.(42) Our study chose this method because it is precise and does not destroy samples after testing.

Because there was no attempt made in this investigation to imitate the mouth cavity. It is also possible for the long-term function of dental implant systems to be negatively affected by caustic chemicals and mechanical stress, both of which can cause the pH of the oral environment to drop. (53)

The gas pycnometer concept

Gas pycnometry has been used for decades to determine the volume of solid seals in a gas-tight system. This is achieved by observing variations in forced pressure and solving the ideal gas law. The ideal gas law proposes a constant connection between a confined gas's pressure, volume, and temperature. When temperature and pressure are recorded, the gas contained in the pycnometer exists in two distinct states: an initial state and a final state.

The initial state a occurs when the connection valve is closed, the reservoir chamber pressure is raised to 2 atmospheres (absolute), and the sample chamber pressure is left at atmospheric.

The ultimate condition was achieved by opening the connecting valve and allowing the two chambers to reach equal pressure. The gas in V1 extends into V2, creating V2. To apply the ideal gas law to this system, the gas's internal energy must remain constant during the forced pressure change.

Monitoring the internal energy of a gas by monitoring its temperature. The criteria of the first law are likewise satisfied if the temperature is same at the starting and end states (in a constant volume system). However, when the valve is opened, the expanding gas causes the system temperature to decrease.**Clinical implication**

Based on our findings that zirconia abutments can accelerate dental implant wear, there may be further cause for concern. Depending on the diameter of the dental implant and the zirconia abutment, the risk of dental implant fracture can be increased. When dental implants are placed too deeply, they may increase the risk of abutment fracture and implant failure.

Parafunctional habits, such as bruxism, must also be evaluated for the long-term effectiveness of dental implants. The function of chewing was not considered. According to various research, simulated mastication has little effect on retention.

Intraorally, saliva has been shown to offer lubrication and reduce attachment wear. For this reason, both attachments may have longer-term clinically superior retention stability.

Further studies

1. This study examined only the effects of cyclic loading on a single implant system and implant-abutment connection design. Further research is required to establish the effectiveness of various implant methods, connectors, and implant - abutment connection design.
2. This study did not investigate the fraction of volume loss responsible for dental implant complications. It could be the next area of research.

Limitation of this study

Due to the experiment being in vitro, no patient-specific factors, intraoral circumstances, temperature changes, or non-axial forces on loading were replicated during the physiological function.

Conclusion

1.Under cyclic loading conditions, Only the titanium implant connected to a zirconia abutment demonstrates wear and volume loss.

2.Some titanium particle transfer seen on the zirconia abutment after cyclic load. This discovery has not yet had any clinical implications, but it is possible that a significant volumetric loss will increase risk of component loosening and eventual fracture.

In addition, the release of titanium particle debris may be a source of concern.

APPENDICES

DENTAL IMPLANT

Descriptives

	Group			Statistic	Std. Error
Preload	1.00	Mean		.1785	.00072
		95% Confidence Interval for Mean	Lower Bound	.1769	
			Upper Bound	.1802	
		5% Trimmed Mean		.1786	
		Median		.1786	
		Variance		.000	
		Std. Deviation		.00227	
		Minimum		.17	
		Maximum		.18	
		Range		.01	
		Interquartile Range		.00	
		Skewness		-.228	.687
		Kurtosis		-1.228	1.334
	2.00	Mean		.1758	.00085
		95% Confidence Interval for Mean	Lower Bound	.1738	
			Upper Bound	.1777	
		5% Trimmed Mean		.1758	
		Median		.1765	
		Variance		.000	
		Std. Deviation		.00268	
		Minimum		.17	
		Maximum		.18	
		Range		.01	
		Interquartile Range		.00	
		Skewness		-.477	.687
		Kurtosis		-1.051	1.334
	3.00	Mean		.1760	.00111
		95% Confidence Interval for Mean	Lower Bound	.1735	
			Upper Bound	.1785	

		5% Trimmed Mean		.1760	
		Median		.1750	
		Variance		.000	
		Std. Deviation		.00351	
		Minimum		.17	
		Maximum		.18	
		Range		.01	
		Interquartile Range		.01	
		Skewness		.194	.687
		Kurtosis		-1.631	1.334
Postload	1.00	Mean		.1617	.00040
		95% Confidence Interval for Mean	Lower Bound	.1608	
			Upper Bound	.1626	
		5% Trimmed Mean		.1618	
		Median		.1618	
		Variance		.000	
		Std. Deviation		.00125	
		Minimum		.16	
		Maximum		.16	
		Range		.00	
		Interquartile Range		.00	
		Skewness		-.961	.687
		Kurtosis		1.570	1.334
	2.00	Mean		.1624	.00046
		95% Confidence Interval for Mean	Lower Bound	.1613	
			Upper Bound	.1634	
		5% Trimmed Mean		.1624	
		Median		.1626	
		Variance		.000	
		Std. Deviation		.00146	
		Minimum		.16	
		Maximum		.16	
		Range		.00	
		Interquartile Range		.00	
		Skewness		-.414	.687
		Kurtosis		-1.366	1.334
	3.00	Mean		.1455	.00073
		95% Confidence Interval for	Lower Bound	.1438	

		Mean	Upper Bound	.1471	
		5% Trimmed Mean		.1454	
		Median		.1443	
		Variance		.000	
		Std. Deviation		.00230	
		Minimum		.14	
		Maximum		.15	
		Range		.01	
		Interquartile Range		.01	
		Skewness		.809	.687
		Kurtosis		-1.372	1.334
percentvloss	1.00	Mean		9.3982	.21284
		95% Confidence Interval for Mean	Lower Bound	8.9167	
			Upper Bound	9.8797	
		5% Trimmed Mean		9.4254	
		Median		9.5478	
		Variance		.453	
		Std. Deviation		.67307	
		Minimum		8.27	
		Maximum		10.04	
		Range		1.77	
		Interquartile Range		1.11	
		Skewness		-.770	.687
		Kurtosis		-.745	1.334
	2.00	Mean		7.6164	.19165
		95% Confidence Interval for Mean	Lower Bound	7.1828	
			Upper Bound	8.0499	
		5% Trimmed Mean		7.6317	
		Median		7.7810	
		Variance		.367	
		Std. Deviation		.60605	
		Minimum		6.48	
		Maximum		8.48	
		Range		2.00	
		Interquartile Range		1.05	
		Skewness		-.630	.687
		Kurtosis		-.195	1.334
	3.00	Mean		17.3302	.55602

		95% Confidence Interval for Mean	Lower Bound	16.0723	
			Upper Bound	18.5880	
		5% Trimmed Mean		17.4378	
		Median		17.3921	
		Variance		3.092	
		Std. Deviation		1.75830	
		Minimum		13.16	
		Maximum		19.57	
		Range		6.41	
		Interquartile Range		1.68	
		Skewness		-1.392	.687
		Kurtosis		3.460	1.334
volumeloss	1.00	Mean		.0168	.00044
		95% Confidence Interval for Mean	Lower Bound	.0158	
			Upper Bound	.0178	
		5% Trimmed Mean		.0168	
		Median		.0172	
		Variance		.000	
		Std. Deviation		.00138	
		Minimum		.01	
		Maximum		.02	
		Range		.00	
		Interquartile Range		.00	
		Skewness		-.624	.687
		Kurtosis		-1.090	1.334
	2.00	Mean		.0134	.00040
		95% Confidence Interval for Mean	Lower Bound	.0125	
			Upper Bound	.0143	
		5% Trimmed Mean		.0134	
		Median		.0137	
		Variance		.000	
		Std. Deviation		.00126	
		Minimum		.01	
		Maximum		.02	
		Range		.00	
		Interquartile Range		.00	
		Skewness		-.566	.687

		Kurtosis		-.387	1.334
	3.00	Mean		.0305	.00111
		95% Confidence Interval for Mean	Lower Bound	.0280	
			Upper Bound	.0331	
		5% Trimmed Mean		.0307	
		Median		.0308	
		Variance		.000	
		Std. Deviation		.00351	
		Minimum		.02	
		Maximum		.04	
		Range		.01	
		Interquartile Range		.00	
		Skewness		-1.190	.687
		Kurtosis		2.605	1.334

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