THE IMPACT OF DIFFERENT CONE-ANGLE IMPLANT - ABUTMENT RELATIONSHIPS ON THE LONG-TERM SUCCESS OF IMPLANT RESTORATIONS

Ph.D. Thesis

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CONTENT

I. PUBLICATIONS

1. Publications related to the subject of the thesis

I. Körtvélyessy Gy, Szabó ÁL, Pelsőczi-Kovács I, Tarjányi T, Tóth Z, Kárpáti K, Matusovits D, Hangyási DB, Baráth ZL: Different Conical Angle Connection of Implant and Abutment Behavior: A Static and Dynamic Load Test and Finite Element Analysis Study. *Materials* 2023; 16(5): e1988.
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II. Körtvélyessy Gy, Hangyási DB, Tarjányi T, Tóth Z, Matusovits D, Pelsőczi-Kovács I, Baráth ZL: Static and Dynamic Compression Load Tests of Conically Connected, Screw Fixed Dental Abutment-Implant Assemblies. *Analecta Technica Szegediensia* 2023; 3.1-12.
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2. Other publications related to the subject of the thesis

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3. Presentations related to the subject of the thesis

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II. A szelíd és az agresszív - avagy a válasz az implantátumot, a fejlesztést, a világmindenséget, meg mindent érintő végső kérdésekre In: Szegedi, Tudományegyetem Fogorvostudományi Kar és a Magyar Fogpótlástani Társaság Szegedi Fogorvosnapok 2019. A Magyar Fogpótlástani Társaság XXIII. Kongresszusa - Implantációs és digitális protetika a 21. században: Szegedi Fogorvostalálkozó és Tudományos Konferencia

II. LIST OF ABBREVIATIONS

AL: Alumnium

AUC: Area under the curve

C: Carbon

Cu: Cuprum

Cr: Chronium

EC: External contact outside the implant platform

FDA: U.S. Food and Drug Administration

FEA: Finite Element Analysis

Fe: Iron, Ferrum

IC: Internal contact within the implant

ICC: Internal conical contact within the implant

INCC: Intra-implant internal non-conical contact

Mn: Mangan

Mo: Molibden

N: Newton

N: Nitrogen

Nb: Niobium

O: Oxygen

PS: Platform Switching

PM: Platform Matching

RCTs: Randomized controlled trials ()

Sn: Stannum

Ti: Conventional grain size titanium base material

V: Vanadium

WHO: World Health Organization

Zn: Zincum

III. INTRODUCTION

3.1. The dental implant, history of implant treatment

Over the last half century, considerable advances have been achieved in human medicine – including dentistry – where many novel treatment modalities were developed rapidly, bringing about significant positive changes in the care of dental patients. However, according to a recent report published by the World Health Organization (WHO) [1], almost half of the world's population (~45% or 3.5 billion people) are affected by some form or oral diseases, which have increased by 1 billion worldwide in the last 30 years. Untreated dental caries is the most common chronic disease worldwide, affecting an estimated 2.5 billion individuals (both children and adults), while severe periodontal disease – a major cause of total tooth loss – is estimated to affect 1 billion people worldwide. Conscious lifestyle choices, the education and promotion of the appropriate preventive oral healthcare habits and the increasing potential for research and development in dentistry are all factors that are working to improve these trends. Long-term, safe treatment of these group of patients could be provided by implant-supported dental prostheses.

Implantology is a constantly evolving discipline of dentistry. In the second half of the 20th century, research focused on implant design, materials and surgical techniques; while later on, surface modification techniques and minimally invasive procedures were developed to achieve osseointegration as quickly as possible. Following the 2000s, research has focused on prosthetics-guided implant planning, digital prosthodontics and factors influencing long-term implant success. Implant-anchored restorations have a survival rate of 89 - 93% over 10 years [2,3], making them a long-term and successful solution.

At the same time as the advent of medicine, replacement of missing teeth was already being considered as a viable medical intervention [4]. The use of dental implants based on titanium (Ti) and titanium alloys is considered to be one of the main treatment options for replacing lost or missing teeth [5]. Such implants are biocompatible, corrosion-resistant, and have favorable mechanical properties [6,7,8]. Furthermore, it has been documented that the success rates of Ti dental implants are high, independently of the implant placement protocol applied [9,10].

The development of modern dental implantation dates back to the early 20th century. Initially, subperiosteal implants were placed under the periosteum. G. Dahl patented his implant in 1941 [11]. The metal plate followed the shape of the jawbone, from which the prosthetic heads protruded, extending into the oral cavity and served to support the tooth replacement. As these were not fixed in the jawbone, they could provoke osteomyelitis due to their movement of up to 2 mm during loading and were considered unsuccessful in the majority of cases. Enosseous implants were then used, which were fixed in the bone; Linkow was responsible for the development of blade-shaped implants made of Ti in the 1960s [12]. The blade implants were encapsulated by connective tissue fibers in the bone, which not only fixed the implant, but were also able to absorb masticatory forces. The most important pioneer of modern implant dentistry, however, was Per-Ingvar Brånemark, who was a professor at the University of Gothenburg [13], carrying out the first preclinical and clinical studies in the 1960s. In parallel, André Schroeder, professor of the University of Bern started to study tissue integration mechanisms of different implant materials, and his group was the first to document the direct bone-to-implant contact of Ti implants in histological sections. Additionally, Schroeder et al. was the first to report on soft tissue reactions of Ti implants; their research helped to define the concepts of dental implant and osseointegration, and to describe the role of screw implant shape, material and surface treatment for long-term success [14].

3.2. Ti and its alloys as dental implant materials

Ti (element 22 of the periodic table) belongs to the group of non-ferrous metals, being the ninth most abundant element, accounting for 0.63% of the Earth's crust. The two most important minerals to access Ti are rutile (TiO2) and ilmenite (FeTiO3). Two allotropic modifications of unalloyed Ti are known in the solid phase: alpha and beta Ti, with further allotropic modifications observed at high pressures above 882°C. Ti was first produced in an impure form by Jöns Jakob Berzelius in 1825, while Matthew Hunter succeeded in producing it in a high-purity form in 1910. Because of its wide range of applications, presently, Ti is one of the most important non-ferrous metals [15]. Titanium raw material is used for the production of many medical devices, from orthopedics to dentistry, but it is also used in other areas of industry, such as watchmaking, aircraft manufacturing, and space research.

Pure Ti with an iron (Fe) content of less than 0.1% is the most suitable for biological use. However, alloys of various types have been developed to improve its mechanical properties. Ti alloys can be divided into two groups: firstly, the substances that promote the stability of the alpha-phase: Al, O, C and N. On the other hand, the so-called beta-bonding agents, which can have isomorphic (e.g., Mo, V Nb), eutectoid (e.g., Mn, Fe, Cr, Cu) or neutral (e.g., Zn, Sn) effects [15].

Biocompatibility, biomechanical functionality and biological stability are some of the most important requirements for dental implant materials. As a result of biological and material science research, pure Ti and some of its alloys have been described as the metal most suitable for these requirements and are used for the manufacture of dental implants [16]. Modifications of pure Ti are characterized by good mechanical properties and resistance against forces in the oral cavity, but lower wear resistance and mechanical resistance compared to Ti alloys [17]. Four groups of unalloyed Ti (Grade 1 to 4) may be distinguished, depending on the amount of impurities (oxygen and Fe) present. The higher the Grade number of Ti, the better its mechanical properties, and the worse its biocompatibility [15]. From Grade 1 to Grade 4, the tensile strength and yield strength increase, but Young's modulus and ductility decrease. Grade 1 Ti is relatively soft and has better heat conductivity, while the highest tensile strength and yield strength are observed in the case of Grade 4 Ti [17]. The most commonly used Ti alloy in dental applications is the Grade 5 Ti alloy (TiAl6V4), which has better mechanical properties than pure Ti, due to its 6% Al and 4% V content by weight. Initially there were concerns about the application Grade 5 Ti alloy due to its corrosion resistance and ion emission. De Morais et al. implanted a Grade 5 orthodontic implant into the tibia of a New Zealand rabbit and concluded that the amount of ions released by the material was not significant, did not exceed the average intake through meals and therefore, there was no evidence of toxicity [18]. The properties of the different modified Ti are summarized in **Table 1**.

Table 1: Summary of the physical properties of Ti modalities up to Grade 1-5.

Property	Grade 1	Grade 2	Grade 3	Grade 4	Grade 5
Tensile strength (MPa)	440	490	440	590-735	860
Yield point (MPa)	282	320	357	470	795

European dental implant manufacturers most commonly use pure Ti – typically Grade 4 – for their dental implants, while Middle Eastern and overseas manufacturers prefer to use Grade 5 which is the international standard for implant abutments and prosthetic components [19].

3.3. Osseointegration and its influencing factors

Brånemark et al., while investigating mycorrhization of bone tissue, accidentally discovered the fixation of implants in bone. In their animal experiments, they observed that when Ti-based metal cells were used, they anchored into the bone after a short time. This process was termed osseointegration by Brånemark, which refers to the direct contact of the implant with the bone tissue without a connective tissue layer at the light microscopic level. This osseointegration refers to the process by which living bone tissue fuses and integrates with an artificial implant, creating a stable and functional connection. The development of osseointegration has been shown to depend on the following parameters: 1. implant material; 2. implant design; 3. implant surface; 4. condition and quality of the bone; 5. surgical technique; 6, implant loading conditions [20].

While in the 1960s and 1970s, the focus of studies was on the implantability and osseointegration of implants, currently, the main aim of studies is to determine the factors affecting the long-term success of implant replacements. An important factor for the long-term success of osseointegrated implants is the relationship between implant and abutment.

In my PhD research we were interested in the implant material and design influencing factors which have a significant role in the osseointegration process. There are several types of implant designs, one of the main difference is in the superstructure. The mechanical stresses caused by masticatory forces may vary for different types of implant-superstructure relationships. The loosening of the fixation screw connecting the implant to the abutment and the irreversible deformation of the implant in the long term are both characteristic of loading. Both phenomenons lead to failure of implant restorations [21-23]. There are several publications investigating the mechanical behaviors of the implants with the Finite Element Analysis (FEA). In the study of Lin et al., a FEA method was to investigate the type of contact between implant and peri-implant hard tissues for which the distribution of masticatory forces is more uniform. They found that the von Miss stress distribution is most uniform for the tapered connection implant - abutment [22].

3.4. Implant-abutment connection

Implant-based dentures have no harmful effects on neighboring teeth and provide an aesthetic prosthesis that is similar to natural teeth [24]. The load transmission mechanism on osseointegrated dental implants considerably differs from that of natural teeth: in the case of dental implants – that are fixed directly in the cortical and cancellous bone – there is no stress reduction (i.e. stress absorption) as with the case of periodontal ligaments in natural teeth, therefore occlusal forces are transmitted directly to the surrounding bone [25]. As a result of reduced stress-bearing capacity, increased bone resorption rates and consequential peri-implant bone defects may develop more easily. An implant-supported dental restoration is a complex system, where the implant-abutment connection has a fundamental role in the long-term stability of the whole unit [26]. Due to occlusal forces, micro cracks and fractures may develop in the implant or in the connected elements [27]. Recurring mechanical forces may lead to reversible or irreversible changes to implant geometry, in addition, they may lead to vertical and horizontal micro-movements between the implant and the abutment, which may result in screw loosening or screw fracture [26]. The forces acting on the dental prosthesis are distributed, they act on the superstructure, the implant (neck, wall thickness, body), the implant connection and subsequently on the adjacent bone through different mechanisms and at different heights, depending on implant connection design and implant geometry [28]. Different implant connections may considerably affect the aforementioned force distribution.

The relationship between the implant and the abutment is usually described as an internal or external relationship. The distinguishing factor separating the two groups is the presence or absence of a geometric element extending above the coronal surface of the implant. The connection may also be described as a *sliding fit*, where there is a small space between the mating parts and the connection is passive, or a *friction fit*, where there is no space between the mating parts and the parts are literally forced together. The mating surfaces may include rotational resistance and indexing and/or lateral stabilizing geometry. This geometry is hereafter described as octagonal, hexagonal, tapered, tapered hexagonal, cylindrical hexagonal, spindle, cam, cam tube and pin/spigot [29]. The shortcoming of the earlier connections was originally noted by Brånemark, who suggested that the external hexagonal connection should be at least 1.2 mm high to provide lateral and rotational stability, especially in single-tooth applications. However, the original 0.7 mm design and its numerous variations remained unchanged until recently. In the literature, hexagonal screw joint complications, mainly due to screw loosening, have been reported, with a prevalence ranging between 6% to 48% [30]. To overcome some of the inherent design limitations of the external hexagonal screw joint, a number of alternative joints have been developed; the most prominent are the tapered screw, tapered hexagon, internal octagon, internal hexagon, cylindrical hexagon, Morse taper, ridge, internal ridge and flexible connection. The new designs were established to improve connection stability during function and insertion, and to simplify the clinician's toolbox for completing the restoration. For dental implants approved for marketing by the Food and Drug Administration (FDA) as early as the early 2000s, there are at least 20 different implant/abutment interface variants [31]. The idea of an internal connection arose due to the unwanted complications of the external connection. Internal implant-abutment relationships have the following advantages:

- Lower platform height for superstructures
- Improved lateral load distribution within the implant
- Protected fixing screw
- Long internal walls provide additional rigidity to the connection
- Long wall lengths for extra longitudinal support
- Potential for antimicrobial sealing

The interface between the implant and the abutment determines the strength and stability of the joint, as well as lateral and rotational stability. An implant with an internal hexagon is usually provided with a hexagon 1.5-2 mm depth. This connection is characterized by a deeper distribution of intraoral forces within the implant to protect the retention screw from excessive loading. Internally connected implants also provide excellent strength for the implant-superstructure connection [32, 33 34]. In order to improve the connection between the implant and the abutment, numerous variations have been developed within the concept of the internal connection [35]. Among these, the Morse taper connection should be highlighted, where a tapered abutment column is inserted into the threadless stem of a dental implant with the same taper. Implants designed with a Morse taper interface engage their abutments by using a fivedegree angulated friction fit internal wall into which an abutment with a rounded male extension is placed. The abutments achieve anti-torsion properties owing to the phenomenon of cold welding, which occurs after the abutment has been inserted and tightened. Cold welding or contact welding is a solid-state welding process, in which the joint is formed at the interface of the two parts to be welded without fusion [36, 37].

Minimising crestal bone loss is essential for the long-term success and survival of implants [38]. In implant dentistry, the concept of platform switch (PS) is based on the placement of a narrow diameter abutment on a wider diameter implant. This concept can basically only be realized with an internal tapered implant-abutment connection. For implants placed according to the PS concept, the implant-abutment interface is closer to the center of the implant (horizontal misalignment) [39]. Studies [40- 42] have reported that implants placed according to this concept are affected by minimal peri-implant bone loss compared to platform-matching (PM) implants (i.e. implants with matching abutment and implant body diameters).

The relationship between implant and abutment can be either of an external or internal nature. According to Leutert [43], the external implant-superstructure relationship has the highest risk of fracture, while the internal relationship has a significantly higher long-term success rate. The risk of fracture of the abutment through-fixing screw is also lower for the internal implant-abutment connection, according to Khraisat et al. [44]. The most common complication for both external and internal connections is loosening of the screw that fixes the implant to the abutment, which is significantly more common for the external connection than for the internal connection [45]. When comparing external hexagonal, internal hexagonal and Morse tapered connections against forces in the same axial direction and oblique forces, the external hexagonal connection was found to be the least resistant against forces in the oblique direction; additionally, this was also where the greatest stress was observed on the surrounding cortical bone. This stress was lower for the inner hexagonal connection, and biomechanically, the Morse tapered connection proved to be the most resistant to stresses [46]. The vertical and horizontal mismatch between implant and abutment alters the load on the entire prosthesis and may lead to screw loosening, screw fracture, bone microcracking, partial ischaemia, crestal bone loss and failed osseointegration [47]. Research by Alikhasi [48] also supports the strong correlation between free rotation of the abutment and screw loosening of the implant-abutment interface. Morse taper connections are able to resist large lateral forces, but the inhibition of superstructure rotation is controversial, as the connection cross-section is circular, so only friction stands in the way of rotation. Based on the results of Kuang-Ta et al. [49], it was concluded that despite the use of octagonal indices, all groups experienced superstructure rotation under high loading. However, their study did not address the question of whether there is a difference between different taper angle connections. For dental implants with a tapered connection currently on the market, there are different versions ranging from a 14-degree Morse taper connection to a 90-degree connection.

The transmission of forces and stress distribution are influenced by numerous factors, such as occlusion, quantity and quality of the bone, implant body design, number and location of implants, implant inclination, osseointegration, abutment design, fit and the micromovements of the abutment, implant platform, thread design at the implant neck and implant-abutment connection, which maybe external (EC) or internal (IC) [50,51]. One of the most common complications is loosening and fracture of the fixing screw. Based on the literature data, screw fracture risk is higher in the case of EC, while the rate of long-term success is considerably higher in the case of IC [52,53]. Stable internal connection allows the implant body to be loaded, and by reducing the micro-movements, loosening and breaking may be avoided [54,55]. Biomechanically, the Morse taper connection has proven to be the most load-resistant when measuring emerging stresses in the implant [56]. Morse connection consists of two interlocking cones in which the friction (cold welding) of the abutment and the body walls of the implant (self-closing connection) ensure that the load is distributed over a large surface [57]. Although Morse taper connections may withstand large lateral forces, inhibiting the rotation of suprastructures is questionable, as the cross-section of the connection is circular, therefore only friction inhibits rotation. Even with the use of octagonal indices, superstructure rotation may occur under heavy loads [57-59]. Lack of anti-rotation may result in subsequent loosening. Numerous studies have demonstrated a strong correlation between free rotation of the abutment and the loosening of the implant-abutment interface [60]. Most implants used in the recent years have an internal conical connection (ICC), although the angle of taper varies widely, and is often combined with an internal hexagonal or an octagonal connection [61]. From a clinical perspective, it is critical to how these combinations impact implant stability. In many cases, different implant manufacturers use the same design of implant-abutment connection, as this simplifies the transition between different systems for the clinicians in everyday practice

[62]. Ideally, the reverse torque value does not change under the effect of load forces, nonetheless, screw loosening is a very common problem. It is expected from the implant-connection that the value of the reverse torque does not change due to the occlusal forces, but in reality, the loosening of the screws is also a very common problem [63,64]. For this reason, there is no clear and ideal decision in terms of the ideal taper angle for the implant-abutment connection. Additionally, these is also a limited understanding whether a change in the value of the taper angle affects the loosening of the screws. During conical fitting, the flexibility of Ti results in a vertical displacement (compression), which may be followed by an irreversible dimensional change (plasticity) in the material of the implant neck [65,66]. It is important to clarify for clinicians how the degree of conicity affects this material deformation. In addition, this deformation – due to the wedge effect – may cause the microcracking of the surrounding bone, as well as result in crestal bone loss and the loosening of the implant. Increasing the inner wall thickness of the implant body or reducing the diameter of the implant-abutment connection reduces the tension in the peri-implant bone. However, due to the internal geometry, this internal wall thickness may only be increased up to a certain limit [67,68]; the most commonly used narrow implants are in a diameter range of 3.3-3.8 mm, the implant wall width is less in case of smaller cone angle, which may increase the risk of irreversible dimensional change mentioned in the implant body [69].

3.5. Impact of implant-abutment relationships on implant success

The failure rate of implants due to static and dynamic loads is relatively high (32%) for implants with inadequate primary stability [70-74]. It is therefore critical to estimate the potential for failure in any given dental implant design. Experimental mechanical testing of dental implants provides useful data for engineers, physicists (involved in implant design) and clinicians [50-52]. In order to avoid failures in implant systems, it is important to have a detailed understanding about the mechanical behavior of the dental implants prior to their clinical application, which may be assessed via mechanical testing of the connection between the implant and the abutment. Parameters such as maximum allowable mechanical stress and reversible deformation, elastic limit and fracture toughness are key indicators for determining the long-term durability of dental implant systems; thus, static and dynamic mechanical measurements should be performed [27,28,50,51,52].

There are several publications on the impact of the implant-abutment connections on implant success. Comparative studies of EC and IC were summarized in a systematic review and meta-analysis by Camps-Font et al. [75]. Similarly, Rodrigues et al. [76] published a systematic review and meta-analysis of randomized controlled trials (RCTs), where the results of ICC were compared to internal non-conical connections (INCC); overall, no significant differences were found between the implant-abutment connections in terms of survival rate and biological complications. ICC showed a greater preservation of peri-implant bone tissue and a lower probability of prosthetic complications than INCC and EC [75,76]. However, most studies have considered implants with different conical angle connections as one group, thus, the effects of different angles on clinical performance are poorly documented in the literature. In addition to peri-implant marginal bone loss and prosthetic complications, more data on implant survival and biological complications could be provided, if more future studies would focus on these angle deviations. Nonetheless, numerous studies indicate that implant diameter and length may considerably affect the stress distribution and occlusal load transfer at bone-implant interfaces [77,78]; for example, based clinical reports, the use of shorter implants presents with considerable disadvantages (i.e. lower success rate, survival), in contrast, there are considerable benefits of increasing implant length (while simultaneously keeping implant diameter constant) in enhancing bone-implant contact area and primary stability, up to a cut-off point of around 12-15 mm [79].

In terms of the mechanical stability of the implant-superstructure connection, in addition to the differences in the raw material and the cone angles, the manufacturing parameters and the dimensional accuracy of the manufactured products also play an important role [80].

Dental implant finite element analysis (FEA) is a computational method used to analyze the biomechanical behavior of dental implants. It is a powerful tool that allows dentists and dental engineers to simulate and analyze the stresses and strains that are exerted on the implant and the surrounding bone tissue. Dental implant FEA is used for a variety of purposes, such as optimizing implant design, predicting implant failure, and determining the optimal loading conditions for dental implants [81].

IV. AIMS OF THE STUDY

The mechanical stability of the implant-abutment connection is one of the most important factors for long-term successful implant restorations. In the literature, it has been reported that a conical connection is the most reliable connection type for dental implants. However, there is limited evidence on how the mechanical properties of the implant-abutment connection are influenced by the small or large taper angle of the connection and the quality of the Ti material used for the abutment. Therefore, the aim of our mechanical studies was to simulate the effect of chewing forces on implant-abutment models with different taper angles and to investigate their mechanical stability in order to assess which taper connection represents a long-term, successful and safe solution for small diameter implants. The loosening of the fixig screw between implant and abutment leads to short term dentures and longer term implant failure, and therefore we investigated how the extension torque values of the fixing screws change under load for implant-abutment models with different taper angles. For which taper-angle connection the highest torque value is retained, i.e. for which one long-term success is likely.

The specific objectives of the study were:

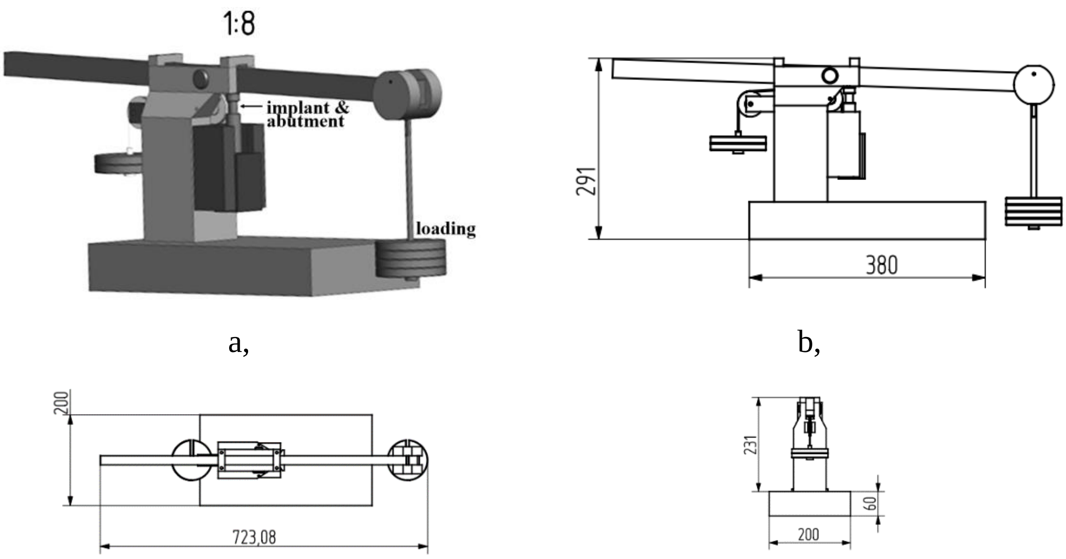
1. Determination of both horizontal and vertical **deformation** of **titanium implants** of **different material** quality (Grade 4 and 5) using a **static mechanical testing protocol**.
2. Determination of whether the **implant material influences** the degree of **screw loosening.** To determine the loosening of fixing screw of implants of different material quality (Grade 4 and 5) using a **static mechanical testing protocol**.
3. Determination of whether **reversible or irreversible deformation** occurs in implant - abutment models with **different conical angle relationships under vertical loading** using a **static mechanical testing protocol**.
4. Determination of whether there is a difference in the **reverse torque values** under **vertical loading** for the fixation screws of **different contact cone angle** models using a **static mechanical testing protocol**.

5. Determination of whether **reversible or irreversible deformation** occurs in implant - abutment models with **different conical angle relationships under vertical loading** using a **dynamic mechanical fatigue protocol**.
6. Determination of whether there is a difference in the **reverse torque values** under **vertical loading** for the fixation screws of **different contact cone angle** models using a **dynamic fatigue testing protocol**.
7. Determination of the **mechanical behavior** of the implant during a **static load with FEA** computational methods, difference of the **mechanical stress** distribution incase of 30 degree and 90 degree along the wall of the implant and at the conical connection
8. Determination of the influence of **design/manufacturing accuracy** on the implant - abutment relationship using **software analytics**.

V. MATERIALS AND METHODS

5.1. Instruments

The first static load test was performed with a self-developed loading machine (**Figure 1**). The one-arm lifter with a lever ratio of 1:8 is loaded by cylinders with a weight of 1.25 kg each. The resulting compression force can be set from 0 N to 500 N.



a, b,

c, d,

Figure 1. Self-developed loading machine

A second-round static load tests and the dynamic load tests were performed with a fatigue machine (Instron ElectroPuls E3000, Norwood, MA, USA). To measure extension torque, a BMS MS150 electric torque screwdriver (BMS Torque Solutions, Ireland) was used. All load tests were performed at the University of Szeged, Faculty of Dentistry.

5.2. Test modells

For the first static load test, the abutments and implant body models were made from Grade 4 and Grade 5 Ti materials, 3.4 and 3.8 mm in diameter, with the following cone angles: 35°, 55°, 75°, and 90° (**Figure 2**). A total of n=84 abutment-implant assemblies were used for the first static load tests. At least 3 samples were tested using the same parameter set.

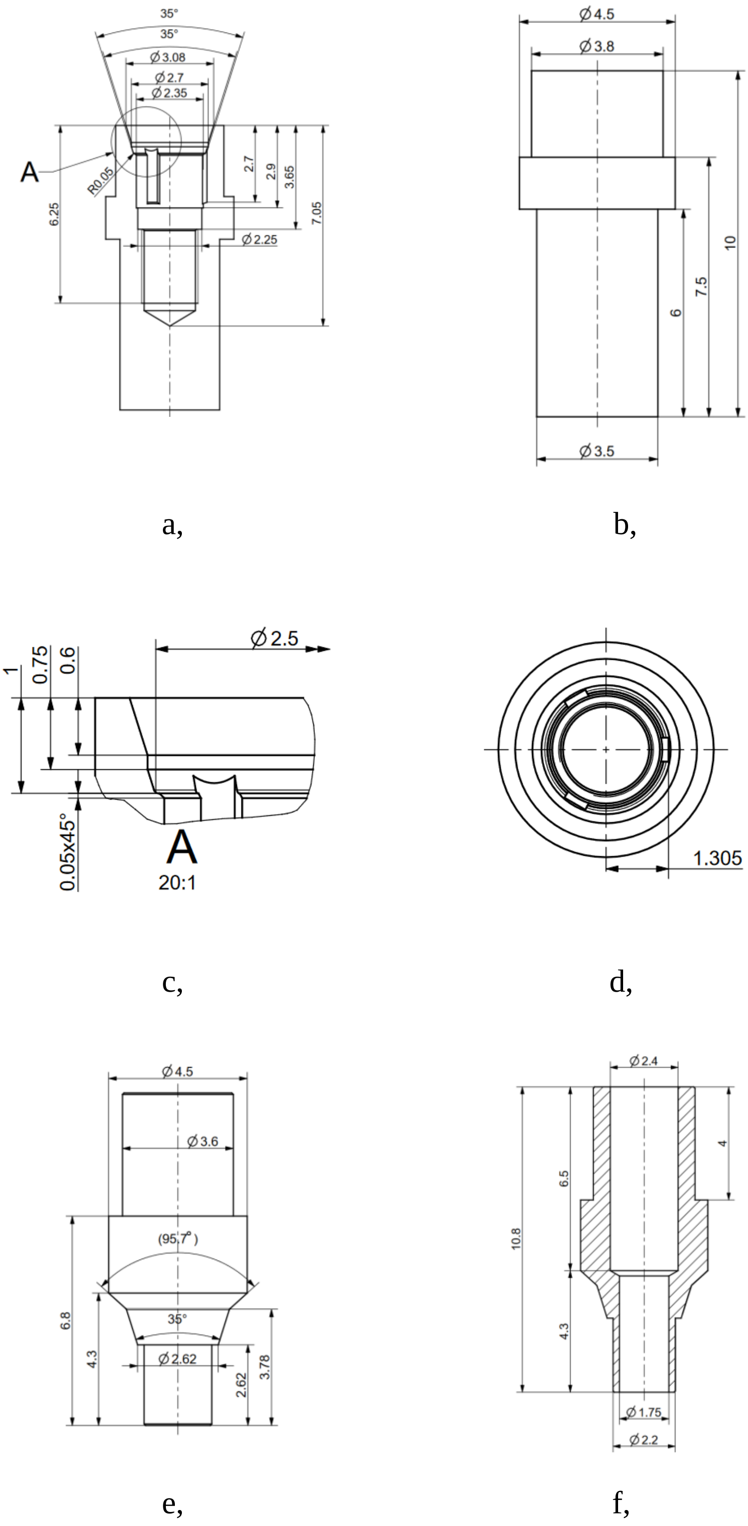


Figure 2. Technical drawings and parameters of a test implant (top two rows, a-d) and the corresponding abutment (bottom row, e-f)

For the first dynamic load test abutments and implant bodies were prepared from Grade 4 Ti in 3.4 mm diameter with the following cone angles: 30°, 45° and 60°. A total of n=21 implant samples were used for dynamic load tests.

Grade 4 Ti implants with 3.4 mm diameter were selected for the second-round static and dynamic tests, with the following cone angles: 24°, 35°, 55°, 75°, and 90°. A total of n=35 implant samples were used for both static and dynamic load tests.

The implant models and abutments were manufactured by Denti System Ltd. (Szentes, Hungary), an example is shown in (**Figure 3**).

Figure 3. Pictures of a test implant and abutment used in the second-round static and dynamic tests

5.3. Static load test protocol

5.3.1. The first static load test protocol

At the beginning of our firsth static load tests, the assembly height and implant diameter of an implant abutment were measured before tightening the fixing screw, and the total height was then measured again after tightening the fixing screw to 35 Ncm. The measurement values were recorded. Static loads were then applied. The choice of the load rate was based on the amount of masticatory force applied to a tooth, 100-200-300-400-450-500 N (100 N corresponds to 10 kg weight). The specimens were subjected to successive static loads of different rates for 60 seconds. Total lengths were measured after the loads. After 50 kg load, implant diameter was measured again. The fixing screw connecting the implant to the superstructure was untwisted after the static loads and the extension torque values were measured. The implants and superstructure were reassembled without tightening the fixing screw and the total length and diameter were measured. This connection was then also fixed by tightening the fixing screw to 35 Ncm torque and the total length was measured again. The bolt was tightened again after 60 seconds, and the reverse torque value was measured after 24 hours.

5.3.2. The second static test protocol

The implant and the abutment head were tightened with the fixing screw, with a torque of 35 Ncm. The specimens were then placed in a special box, fabricated for this experiment, that held them under the load head during loading. The samples were then pressed by the machine, the force being perpendicular to the surface of the implant abutment, as seen on **Figure 4**. The compression value reading (in N) was obtained by the machine from the position of the loading head. During the load test, the load was gradually increased to 500 N over 20 seconds, and after reaching this peak force, the load was decreased back to 0 N over another 20 seconds. After each load test, the extension torque of the fixing screw were measured by an electric torque screwdriver. From the obtained load curves, resilience and the energy dissipation were calculated from the area under the curve (AUC), with a numerical method.

Figure 4. Fatigue machine used during the experiments and the setup of the static and the dynamic load tests. The loading head was perpendicular to the top surface of the implant head

5.4. Dynamic load test protocols

5.4.1. The first dynamic load test protocol

On each implant our protocol started with tightening the fixing screw between the implant abutment and implant with 35 Ncm torque. After this the assembled implant model samples were put under loads. In the first phase 250 N (equivalent to 25 kg) load over 10 seconds was applied on the implant abutment. The dynamic load test started after this first phase. During the dynamic load test a periodic force with 150 N (equivalent to 15 kg) amplitude sine wave was applied with 15 Hz frequency. This results in a force that varies dynamically over time between 0.1 kN and 0.4 kN. The fatigue test lasted 30000 cycles. After the fatigue testing the 250 N loading force was released over another 10 seconds until zero. After this process each implant and implant head were disassembled by untwisting the fixing screw and during this process the reverse torque was measured.

5.4.2. The second dynamic load test protocol

After the second static load test, the same samples were used for further fatigue tests. On each implant, the first step of our measurement was the tightening of the fixing screw that holds together the implant abutment and the implant, with 35 Ncm torque [82]. Following this, the implants were put under loading. In the first phase, the fatigue machine loaded with 0.25 kN (equivalent to 25 kg) force over 10 seconds on the abutment. The dynamic load test started following the first phase; during the test, a 0.15 kN (equivalent to 15 kg) amplitude sine wave was applied with 10 Hz frequency. This resulted in a dynamically changing force between 0.1 kN and 0.4 kN over time. The fatigue test lasted 15,000 cycles. After the fatigue test was done, the force was released over another 10 seconds. Following this process, each implant and abutment were unscrewed and the torque was measured.

5.5. Finite element analysis (FEA)

A preliminary FEA was performed to examine the mechanical stress occurring in the implant, in case of different cone angles. For the purposes of FEA, the COMSOL Multiphysics 5.5 software (COMSOL Inc., Burlington, MA, USA) was utilized. For the study, the 24° and 90° cone angle implant and abutment models were modelled, based on the manufactured samples that were presented in this study. In our analysis, the implant and the abutment model were interpreted as one body with perfect tight fit. The number of elements (tetrahedra) were 10589, with a mesh volume of 63.06 mm^3, average element quality of 0.5975 and element volume ratio of 1.29E-4, respectively. The mesh was created and adjusted by the built-in physics-controlled sequence of the COMSOL software, while element size was set to normal in the settings [83]. We tested the mesh-stress convergence with changing the different Comsol meshing options.

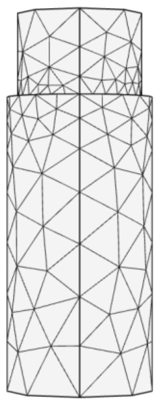
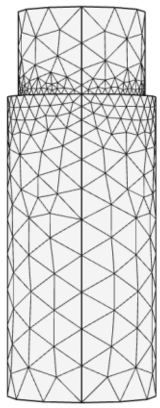
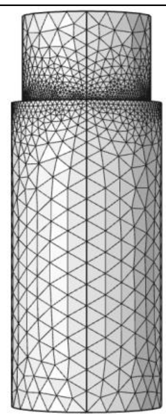
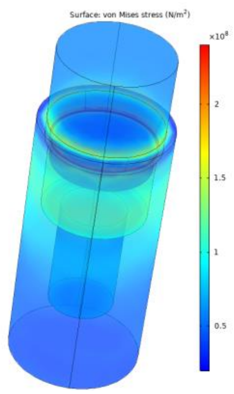
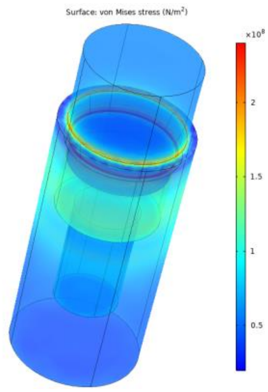
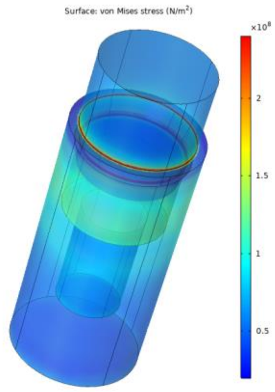
Mesh coarser (2520 number of elements, 1282 triangles)	Mesh normal (10589 number of elements, 3692 triangles)	Mesh finer (91772 number of elements, 17310 triangles)
		
		

Figure 5. The different mesh and stress distribution in the implant

The implants with the smallest (24°) and greatest (90°) cone angle cases available for us were chosen to be included in the FEA. The models were compressed with 400 N at the top, and the Ti material parameters were the following: density: 4500 kg/m^3, Young's modulus: 110 GPa, and Poisson's ratio 0.34. The Yield limit of the CP 4 Ti was considered as 480 MPa [84]. Many publications discuss Ti raw material as tough and resistant to plastic deformation [85,86].

5.6. Inaccuracies due to manufacturing parameters

The mechanical parameters of the test models used in the study, as well as their parameter tolerances, have a significant impact on the performance of the products. All devices used in our study were designed using the Creo Parametric 5.0 software. As part of our investigation, with the help of the design software implant parts were drawn with the acceptable tolerances of the parameters, we compared the tight fit case with the manufactured extreme but still tolerated fit case. With this test, we validated a connection to the extent of vertical deviations in the case of elements manufactured to the worst tolerance values in the opposite direction of the products compared to the ideal, exact size pieces.

5.7. Statistical analysis

Results of the measurements were presented as mean±SEM (standard error of the mean). Statistical analyses were performed using IBM SPSS 23.0 (IBM Corp., Somers, NY, USA) software; one-way analysis of variance (ANOVA) followed by Tukey HSD post hoc tests were performed on the measured values. During analyses, p values < 0.05 were considered statistically significant. A linear regression was performed on the measured data, and the fit equation with the R^2 values were determined.

VI. RESULTS

6.1. Static load results

6.1.1. The first round of static load tests

During the first static mechanical loading tests, the lengths of the abutment implant assemblies were measured after applying 0, 100, 200, 300, 400, 450 and 500 N compression forces; the results of these measurements can be seen in **Figure 6**. Overall, no significant differences were shown between the behaviours of implants from Grade 4 and Grade 5 raw materials in the case of connections with different taper angles (35°: p=0.562; 55°: p=0.666; 75°: p=0.235; 90° p=0.944). The largest strain was obtained for the 35° angle connections, for both Grade 4 and Grade 5 assemblies, respectively, as can be clearly seen in **Figure 6**.

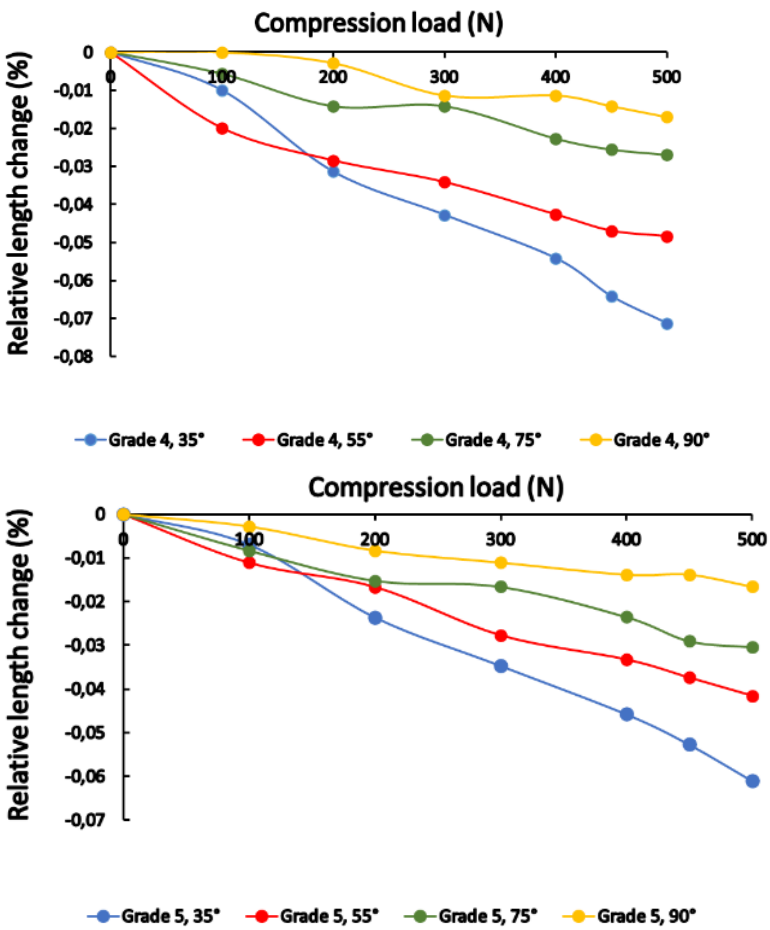


Figure 6. Relative length changes for Grade 4 and Grade 5 Ti implants after different static loads

After the application of 500 N compression load, the reverse torque was measured upon disassembling the implant and abutment parts; the results are presented in **Figure 7.**, with a comparison of Grade 4 and Grade 5 implants. Overall, reverse torque values consistently increased with the conical angle, i.e., the lower values were observed for the 35° conical angle case, while the highest for the 90° case. On the other hand, no significant differences were noted when comparing the reverse torque values of grade 4 and 5 implants with the same conical angle (*p*>0.05).

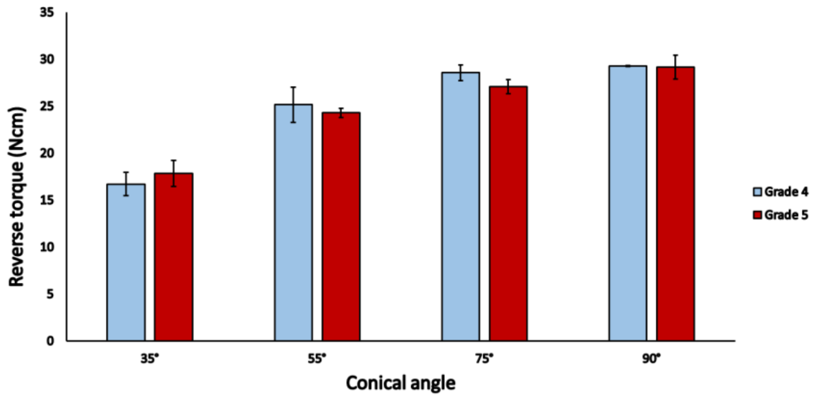
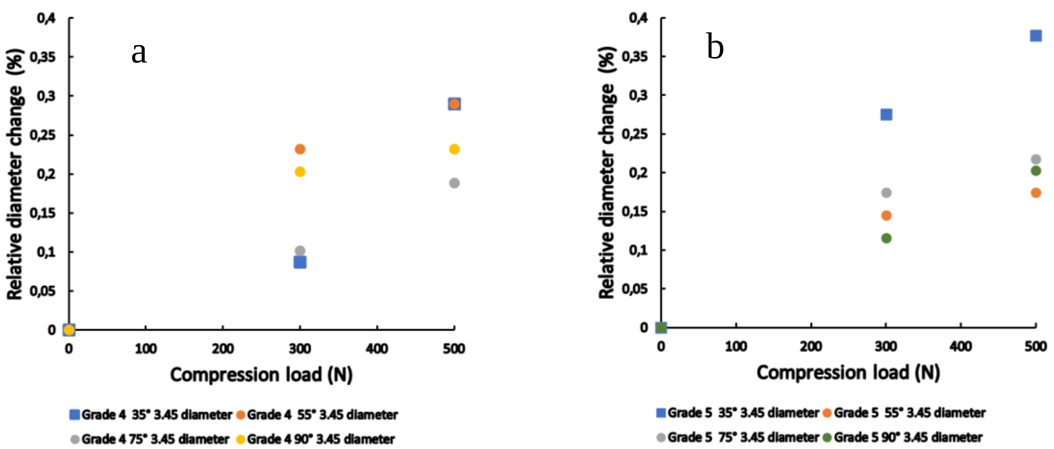


Figure 7. Reverse torque (mean ± SEM) values in case of Grade 4 and Grade 5 Ti implants with different conical angle cases (35°, 55°, 75°, and 90°)

Diameter changes of the implant models corresponding to different loads (0, 300, 500 N) were examined, during which, diameter values were measured at three locations. **Figures 8 a-b and Figures 8 c-d** show the relative diameter change for the Grade 4 and Grade 5 3.45 mm and 3.8 mm diameter implant models, respectively (where each data point represents the average value of 6 samples). No significant differences were found between the case of the different Ti grades.



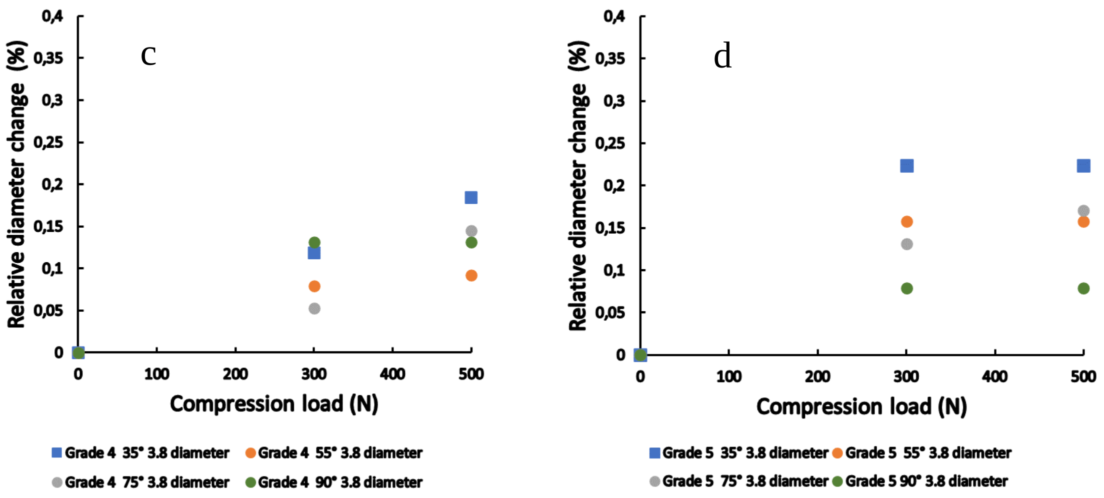


Figure 8. Relative diameter changes of the Grade 4 and Grade 5, Ø 3.45 mm and Ø 3.8 mm Ti implants under different static compression loads, corresponding to the tested internal taper contact angles (35°, 55°, 75°, and 90°)

The result of the first static measurements showed that the smaller the cone angle between the implant and the superstructure, the greater the diameter increase of the implant at the conical closure under load. As a result, the mechanical stress value of the implant body on the bone will be higher, that is, more stress is transferred to the bone, which may lead to increased bone resorption.

6.1.2. The second round of static load tests

During the second static load tests, the device recorded the vertical compression of the abutment in to the implant and force. The representative load compression graphs are presented in **Figure 9**. for each cone angle (24°, 35°, 55°, 75°, and 90°) group. As the force gradually increased, a linear relationship was observed with the compression, i.e. the load was in the elastic region of Ti. Different conical angle implant-abutment connections showed different load curves, i.e. there were differences in how the compression increased due to the load. The smallest compression was obtained with a cone angle of 75°, while the highest was in the case of 35°.

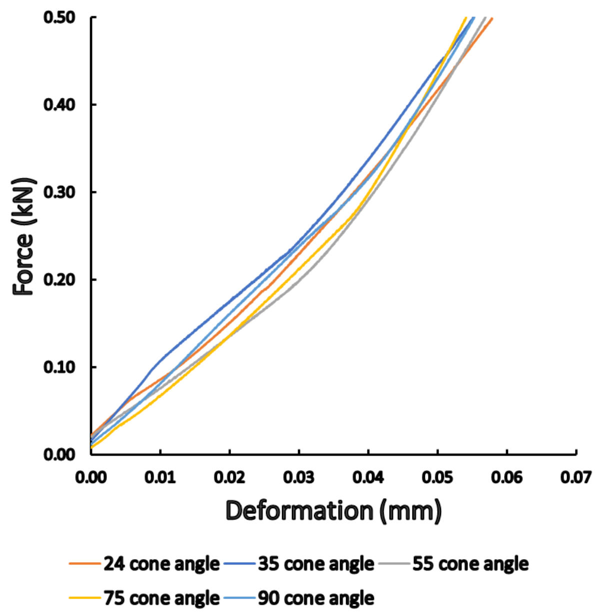


Figure 9. Results of the static load tests. Representative load compression graphs showed differences among different conical angle implants

The compression rate of implants with different cone angles were compared at the highest static force value, as shown in **Figure 10**. Significant differences among the mean compression rates of implants with different cone angles were seen (p = 0.021); however, based on post hoc analyses, only the 35° and 75° cone angle implants were significant different (0.067 ± 0.008 mm vs. 0.044 ± 0.003 mm; p = 0.032).

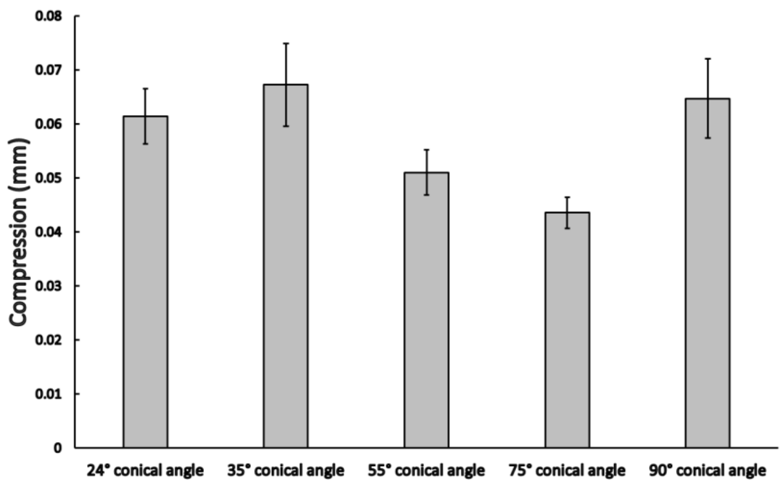


Figure 10. Compression rate (mean ± SEM) among different conical angle implants in the static load tests

After the deload, the irreversible vertical compression of the implant and the abutment was determined. Our results showed no significant differences between the different cone angles (p = 0.08). Cone angles of 24°, 35° and 90° showed a similar mean irreversible compression rate of ~ 0.022 mm (**Figure 11**); on the other hand, the 75° cone angle case showed the lowest irreversible vertical compression.

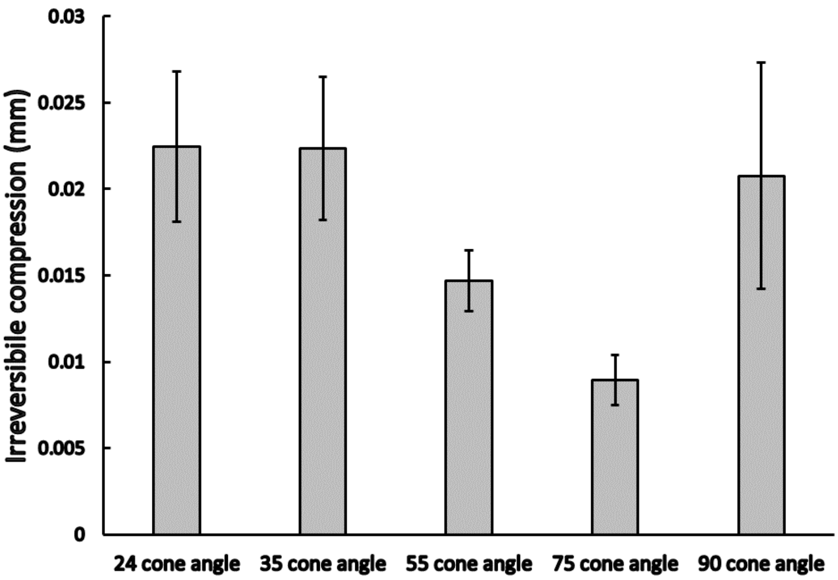


Figure 11. Irreversible vertical compression rate (mean ± SEM) among different conical angle implants and abutment connections

From the area of the load curves, resilience and energy dissipation were determined. The different cone angles cases showed significant differences both in the case of resilience (p = 0.02) and energy dissipation (p = 0.01), as demonstrated on **Figure 12**. Highest resilience values were found in case of 24° (38293 ± 2640 $\frac{J}{m^3}$), and 35° (40221 ± 5194 $\frac{J}{m^3}$) and interestingly, the 75° cone angle case showed the lowest resilience value (25748 ± 1357 $\frac{J}{m^3}$). The dissipated energy showed a similar order, 24° (17165 ± 2325 $\frac{J}{m^3}$) and 35° (16014 ± 3333 $\frac{J}{m^3}$) were the highest, and the 75° (6129 ± 731 $\frac{J}{m^3}$) case was the lowest.

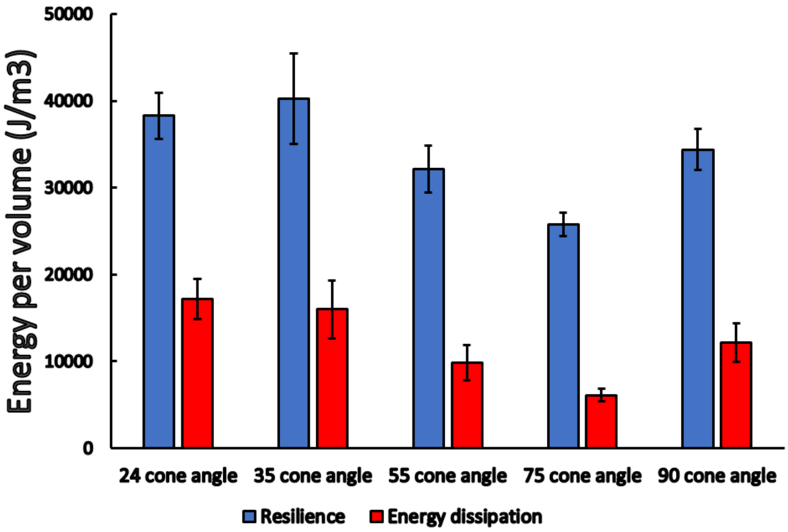


Figure 12. Resilience and energy dissipation (mean ± SEM) among different conical angle implants and abutment connection

Table 2. Results of the post-hoc tests (*p*-values) during pairwise comparisons of resilience (blue) and energy dissipation (red) among different conical angle implants. *p*-values <0.05 are presented in **boldface**

	24°	35°	55°	75°	90°
24°	-	0.721	**0.030**	**0.002**	0.145
35°	0.666	-	0.065	**0.004**	0.260
55°	0.174	0.077	-	0.250	0.493
75°	**0.008**	**0.003**	0.159	-	0.079
90°	0.404	0.215	0.626	0.070	-

6.2. Dynamic load results

6.2.1. The first round of dynamic load tests

During the dynamic load test, the fatigue testing machine recorded the loading head position from which, compression strain of the implants with different conical angles (30°, 45° and 60°) can be determined. It was observed that there is a permanent deformation in the material, as the loading and unloading curves did not coincide. However, due to the elastic properties during the unloading phase, the material can still partially recover its length. It was also studied whether there were any differences in the impression of the implant head into the implant body during the fatigue cycles. **Figure 13** indicates that the implant head and the implant moved indeed closer together. The loading-unloading nature of this test revealed that most of the impression occur in the very first cycles, while it remains constant in the subsequent cycles; this is shown as the curves immediately begin to shift to larger displacement values, and after that there is no essential change between the loading-unloading cycles. Thus, the samples suffered elastic deformation mainly after this early phase.

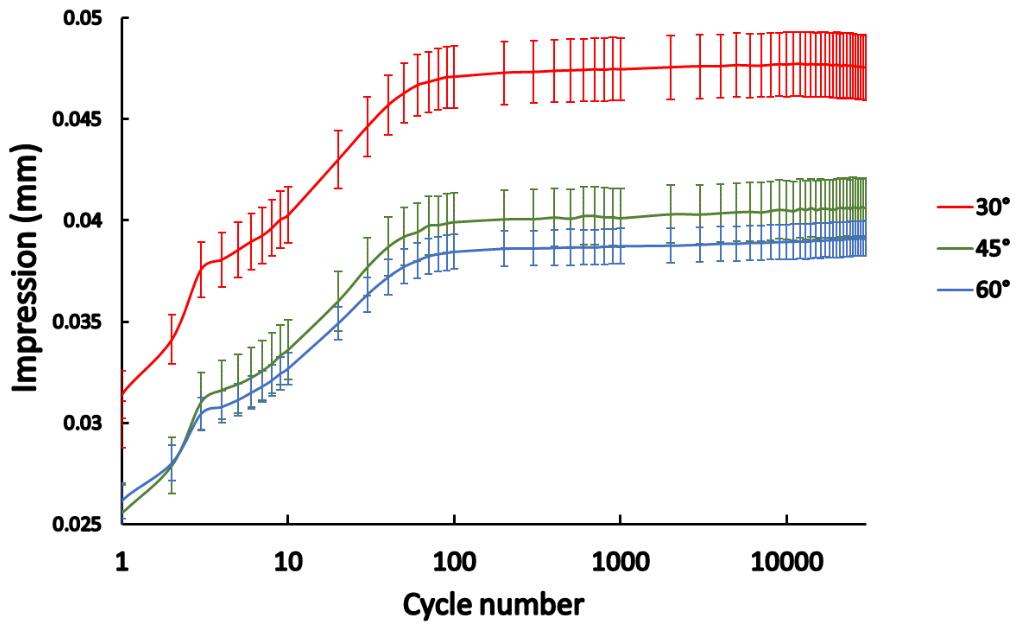


Figure 13. Mean impression levels of the implant head into the implant over the dynamic load test cycles. Sampling was carried out more frequently in the early phase of the test, while in the later stages, sampling was done in every 1000 cycles

Final displacement values – indicating irreversible impressions of the abutments into the implant bodies – were also measured with the dynamic testing machine, following the dynamic test, as shown in **Figure 14**. The highest impression value was measured for the 30° case (0.047±0.002 mm), while the lowest value was found in the 60° case (0.039±0.001 mm), respectively; observed differences between the conical angle groups were statistically significant ($p < 0.001$).

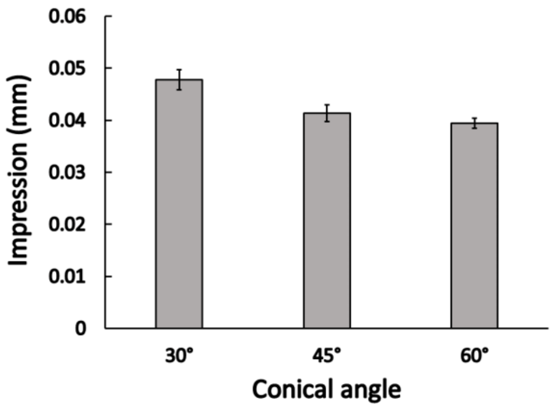


Figure 14. Displacement values (mean ± SEM) measured for each conical angle group (30°, 45° and 60°) after the fatigue test

After the dynamic loading test, reverse torque values were also measured upon disassembling the implant head and implant body; the resulting reverse torque values are shown in **Figure 15.** Similar observations were obtained to our previous measurement, the lowest torque values were shown in for the 30° conical angle case, while the highest were seen for 60°; significant differences were noted among the mean reverse torque values; significant differences were observed between the average torque (p = 0.003) in case of 30° (18.7 ± 1.01 Ncm), 45° (21.25 ± 0.67 Ncm) and 60° (24.03 ± 0.59 Ncm). Additionally, based post hoc test, a significant difference between the 30° and 60° conical connections (p = 0.043) was verified, while this was not the case for the 30° vs. 45° and 45° vs. 60° comparisons.

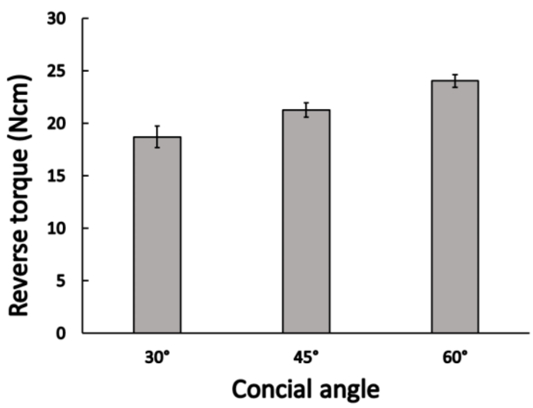


Figure 15. Reverse torque (mean ± SEM) measured for each conical angle group (30°, 45° and 60°) upon disassembling the implant head and implant after the fatigue tests

6.2.2. The second round of dynamic load tests

During the dynamic load test, the fatigue machine recorded the load head position, which was in a direct relationship with the vertical compression of the implants (i.e. how much the abutment slipped into the implant structure). The device also recorded the given force values over time, therefore load-compression graphs could be analyzed. The load-compression results for the whole 15000 cycles are presented in **Figure 16**; it may be observed that there are different degrees of average vertical compression based on the different conical angle. The highest compression was measured in the case of 35° and 55° while the lowest was at the 75° and 90° conical angle implants. Due to the elastic properties during the deload, the material may still deform back. The vertical compression occurred in the very first cycle, while it remains constant thereafter, which may be identified on **Figure 16**; in the beginning of the compression cycles, there was a sudden rise in compression and after that there was no change in compression rate.

The samples only deformed elastically mainly after this early phase. At the end of the fatigue test, the compression was gathered before the deload phase. The results of the vertical compressions at the end of the dynamic load may be observed from **Figure 17.**, where significant differences were observed among the different cases (p = 0.029); comparative analyses showed that there is a significant difference between the 35° and 75° (0.049 ± 0.004 mm vs. 0.037 ± 0.002 mm; p = 0.011 see Table 3), and in case of 55° and 75° conical angle implants (0.046 ± 0.003 mm vs. 0.037 ± 0.002 mm; p = 0.009 see **Table 3**). The permanent deformations were also measured after the dynamic load test: the loading head deloaded the samples and the final position was recorded; these results may be seen in **Figure 18.**, where significant differences between the mean permanent deformations were noted (p = 0.032).

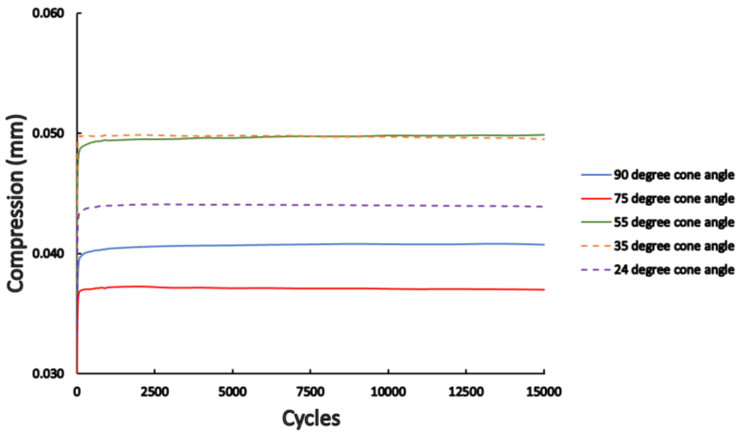


Figure 16. The mean measurements for dynamic compression force among different conical angle implants for all the 15,000 cycles

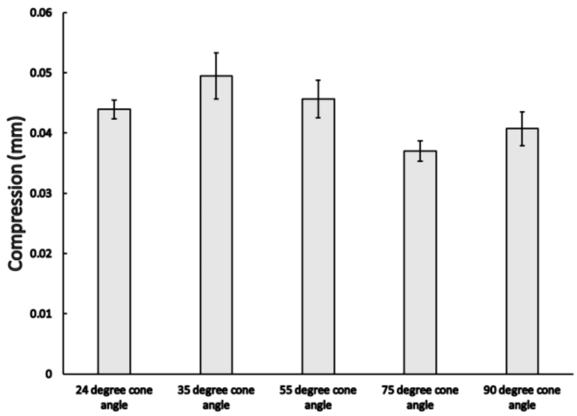


Figure 17. Total vertical compression (mean ± SEM) among different conical angle implants at the 15000th cycle

Table 3. Results of the post-hoc tests (*p*-values) during pairwise comparisons of mean vertical compressions among different conical angle implants. *p*-values < 0.05 are presented in **boldface**

	24°	35°	55°	75°	90°
24°	-				
35°	0.234	-			
55°	0.204	0.932	-		
75°	0.141	**0.011**	**0.009**	-	
90°	0.490	0.065	0.055	0.422	-

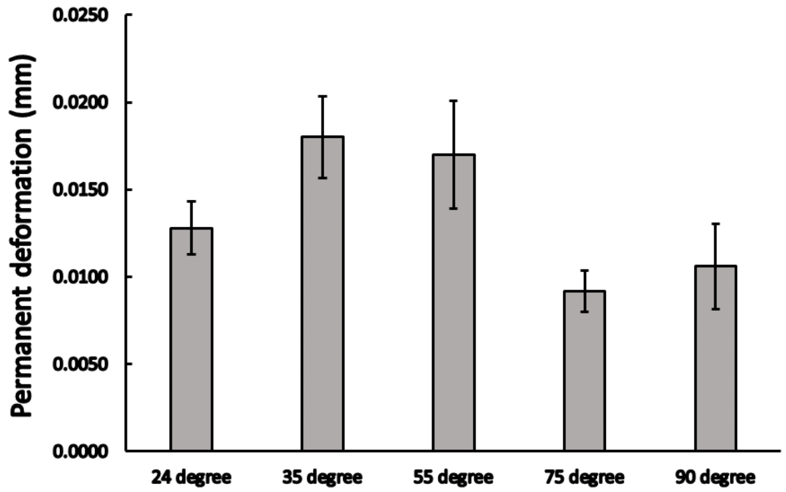


Figure 18. The irreversible (permanent) deformation (mean ± SEM) among different conical angle implants in the implant-abutment system after the dynamic test

After the dynamic load test, the lowest torque needed to roll apart the abutment and implant were also measured (**Figure 19**). Lowest torque values were noted for the 24° case (13.1 ± 1.26 Ncm), while the highest was for the 90° case (29.4 ± 1.1 Ncm). Significant differences were observed between the mean torque both in the case of static and dynamic load tests, ($p < 0.001$ in both cases). With the exception of the 24°- and 35°-degree conical angle connections ($p = 0.384$ and $p = 0.994$), there were significant differences among every other case both after the static and dynamic load test ($p < 0.05$).

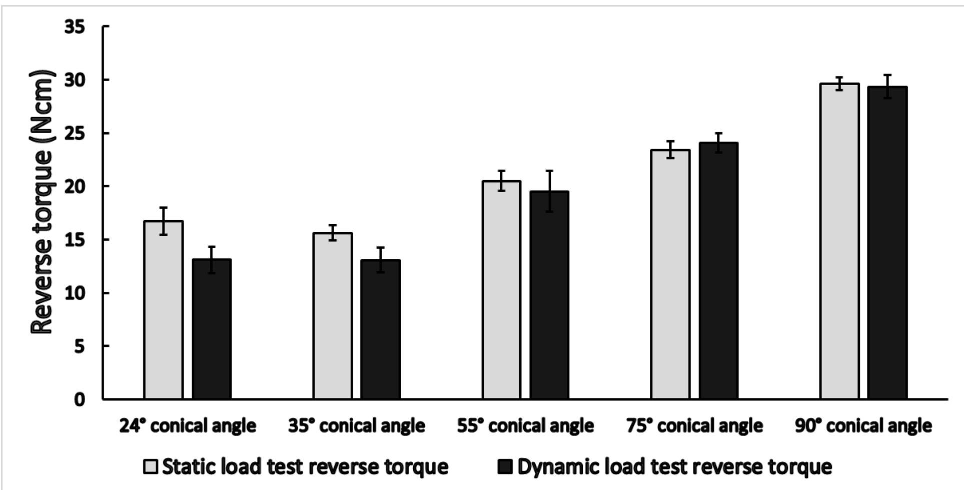
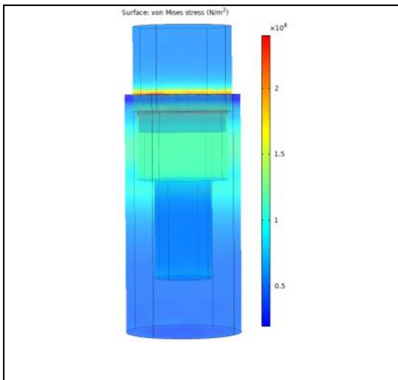


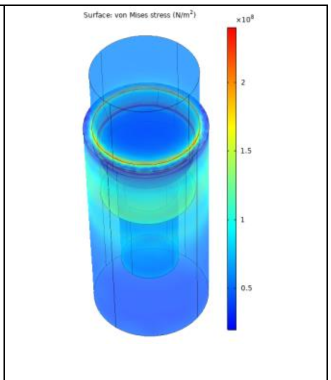
Figure 19. Reverse torque (mean ± SEM) needed to roll apart the implant head and implant after the fatigue test among different conical angle implants

6.3. Finite element analysis

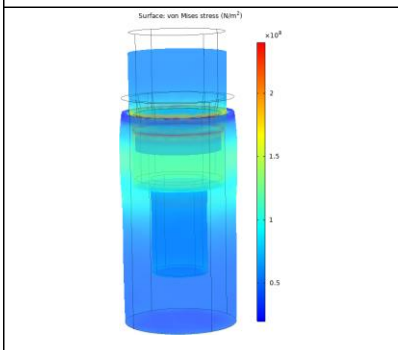
The FEA showed a pronounced difference between the two selected cone angle implant models in von Mises stresses. In the case of 24°, the calculated mechanical stress presenting in the implant was roughly 3-times greater than in the case of 90°. Also, around 130 MPa of mechanical stress was concentrated in the upper third of the implant, with the highest stress values seen at the conical surface. **Figure 20 a** and **b** shows the mechanical stress and deformation in case of 24° implant, while **Figure 20 c** and **d** shows the same mechanical stress but with the deformation being scaled by 100. The horizontal deformation, in barrel-shape is demonstrated on the scaled **Figure 20 c.**, on the other hand, this barrel-shape deformation cannot be observed on the scaled figures for the 90° case (**Figure 20 e-h**). In the case of 90° cone angle implant, around 60 MPa mechanical stress was distributed equally on the implant wall and the mechanical stress peaked at the conical connection. The highest mechanical stress value noted was around 300 MPa in the case of 24°, while it was only around 160 MPa in case of the 90° conical angle case.



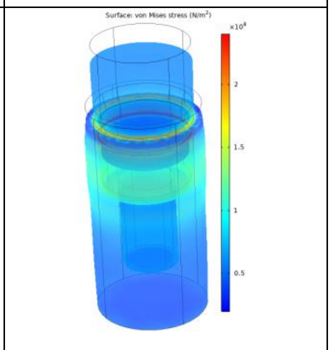
(a)



(b)



(c)



(d)

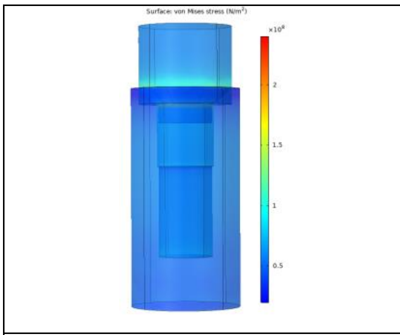
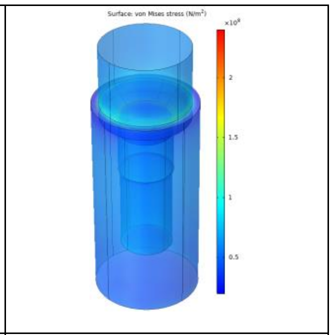
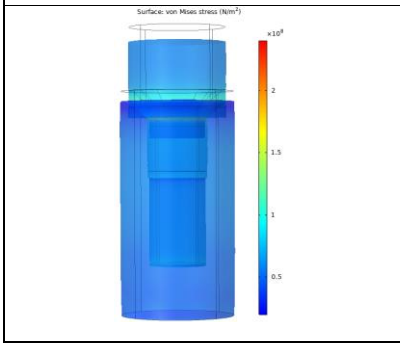
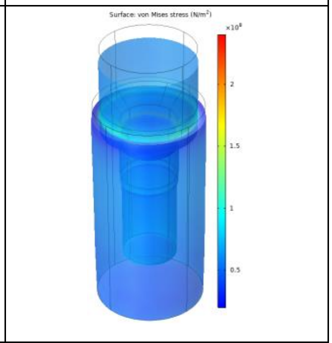
	
(e)	(f)
	
(g)	(h)

Figure 20 a-h. Finite element analyses of the mechanical stresses in case of compression at 24° **(a-d)** and 90° **(e-h)** conical angle implant and abutment model geometries

The final stress distributions after the calculation in the different meshing size options did not change drastically the results. The following figure shows the different mesh and stress distribution figures. To evaluate mechanical stress, a line was defined on the conical surface between the abutment and the implant body (**Figure 21 and 22**).

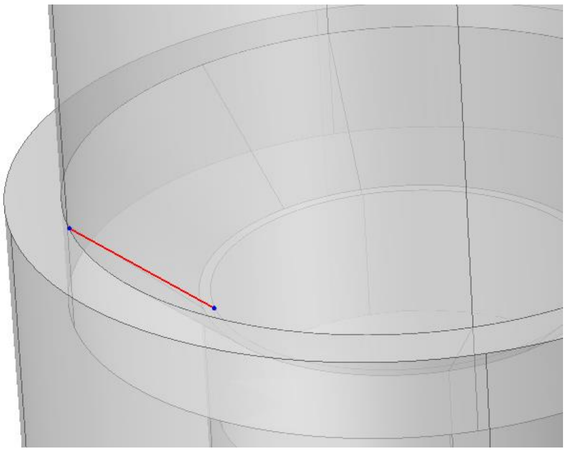


Figure 21. The selected line on the conical interface in the case of 90 degrees connection

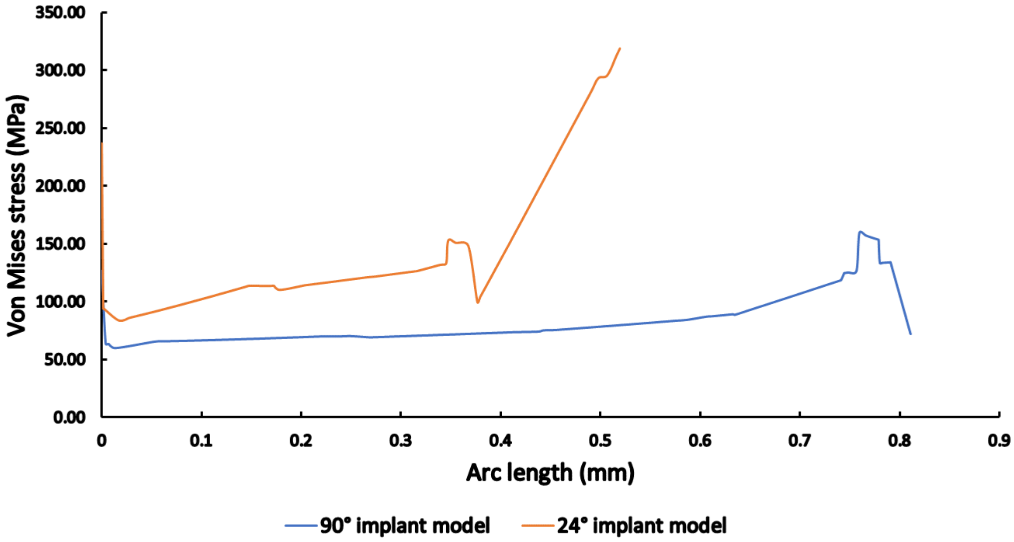


Figure 22. The mechanical stress distribution along the selected line on the abutment and implant connection in case of 24° and 90°

Furthermore, we also added a diagram of the mechanical stress distribution along the implant height on the side which shows the stress in the wall of the implant **(Figure 23).** Here it can be seen that clearly the stress is much higher on the implant wall in the case of the 24° implant model, corresponding to values near 120 MPa, while the stress values were only around 60 MPa in the case of the 90° model **(Figure 24)**.

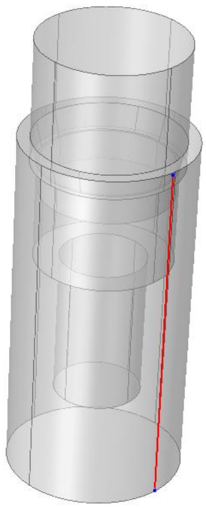


Figure 23. The selected line on the implant side for the calculation of the mechanical stress

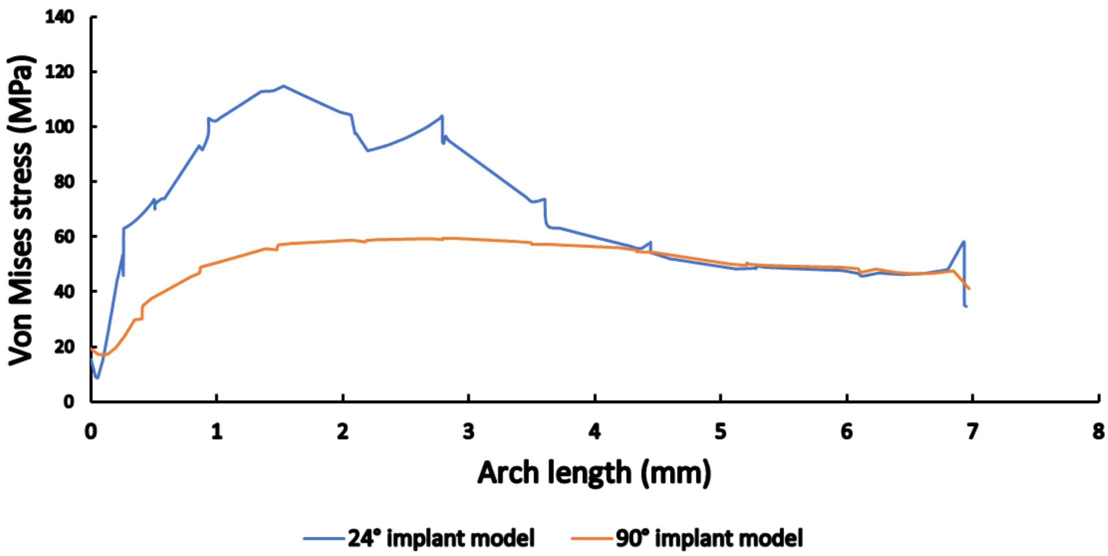


Figure 24. The mechanical stress distribution along the implant height on the side

6.4. Inaccuracies due to manufacturing parameters

Several implants on the market from different manufacturers were examined, but with the same internal design, and measured a ±1° cone angle difference. For the implants in our samples, a taper angle tolerance of +0/-0.5° was accepted, but during the production of the samples, a deviation close to zero degrees is preferable. For the abutments, the same tolerance of 0/+0.5° was acceptable (**Figure 25**).

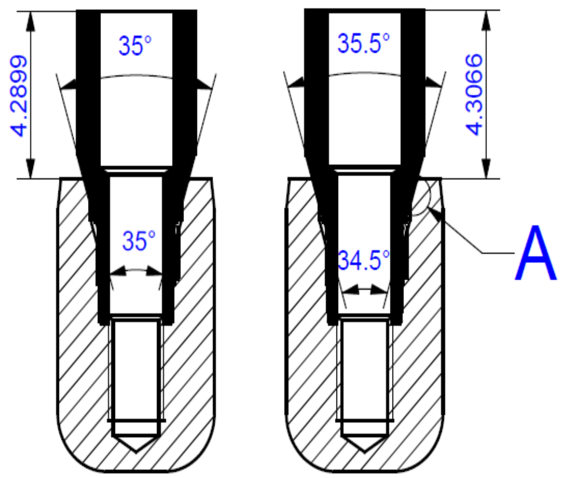


Figure 25. Schematic drawing of the conical implant and abutment head, marked with the angles and lengths

We checked that when the tolerances are matched in the worst possible pairing, the height differences of the assembled pieces and the cone intersections show the deviations, summarized in **Table 4**:

Table 4. Implant abutment height difference for contacting angles with opposite tolerances

Ideal connection angle	Negative tolerance of the implant cone angle	Positive tolerance of the abutment cone angle	Resulting height difference [mm]
24°	23.5°	24.5°	0.0271
35°	34.5°	35.5°	0.0167
55°	54.5°	55.5°	0.0071
75°	74.5°	75.5°	0.0041
90°	89.5°	90.5°	0.003

VII. DISCUSSION

The aim of our study was to understand the effect of different taper-angle implant abutment relationships on the long-term survival and clinical success of dental implants. During our investigation, two issues were identified: on the one hand – according to the results of the mechanical tests – the screw loosening, as well as the horizontal deformation, based on the measurements and the results of the FEA.

The results of our mechanical tests conclude that as the cone angle of the implant superstructure connection increases, the extraction torque of our fixing screw decreased proportionally, i.e. the larger the connection angle, the loosening of the screw under load was smaller. The management of screw loosening is a common issue in clinical practice; the change in the torque is affected by the continuous, periodical repeating loads. Our static and dynamic results both showed the highest change in case of lower conical angles. This resulted in the decrease of the reverse torque needed to take apart the implant and the abutment. However increasing the conical angle of the abutment improved the results, i.e. it has led to lower rates of compression and less decrease in the case of the small angle cases.

Both the first static and first dynamic measurements reveal the same results. It can be stated that the conical closure was also clearly visible in the case of the implant-superstructure connection with different tapers. There is also a direct proportionality in the case of the reverse torque, i.e., the larger the cone angle of the connection, the greater the reverse torque value we measured.

Screw loosening of taper-connected implants, i.e., a loose state of the clip between the implant and the retaining screw, can be a significant problem, as the screw loosening increases the risk of removal of dentures and the possibility of screw failure. Screw loosening of implants can be traced back to various reasons, such as incorrect tension of the screw, excessive load on the screw, a defect in the material or size of the fixation screw, wear between the implant and the fixation screw, or continuous slippage of the screw due to continuous loading. Therefore, it is important that the implant and retaining screw are properly secured and that dentists inspect the suprastructure and implants to prevent screw loosening.

The reversible and irreversible shape deformations of titanium implants can have a significant impact on the success of the implant surgery and the long-term stability of the implant. In reversible deformation, deformation occurs during surgery or subsequent loading (mastication) due to the elasticity of the titanium implant, but the implant returns to its original shape when the load is removed. This type of shape deformation is usually associated with micro- and macro-deformations due to loading, which can lead to fatigue fracture in the long term. During irreversible shape deformation, the implant does not return to its original shape, even after the load is removed. This is usually due to exceeding the yield point or excessive deformation during surgery. Irreversible deformation can have a serious effect on the stability of the implant, as the shape of the implant changes and, as a result, the implant does not fit properly with the implant bed. In order to prevent irreversible deformity, dentists must be careful to choose the right size and shape of implant and carefully plan and perform surgical procedures. Planning, with pre-planning and simulation, plays an important role in selecting the right sized and shaped implants for the patient and ensuring the correct fit of the implant to the implant bed. Performing professional procedures and using appropriately chosen implants can reduce the risk of irreversible shape deformations and improve the long-term stability of the implants.

Irreversible vertical deformation may also cause compression of the implant and the abutment, if the taper angle of the connection is small, a phenomenon that may be exacerbated by manufacturing inaccuracy, i.e. the height of the implant and superstructure may change as a result of the load. In this case, the occlusal height also decreases, thereby changing the occlusion, which leads to further biological (malocclusion, traumatic occlusal forces, peri-implant bone loss, temporo-mandibular dysfunction) and mechanical (screw loosening, fracture, superstructure deformation, fracture) issues [87]. The inaccuracy of the taper angles of the implant and the abutment has a considerable influence on the compression under load. The greater the dimensional error from manufacturing, the greater the conical surface shrinkage [88]. In our results the inaccuracy of the taper angles of the implant and the abutment has a significant influence on the compression under load as well. The greater the dimensional error from manufacturing, the greater the impression of the abutment into the implant body along the conical surfaces.

Ti is traditionally described as a material which is tough and resistant to plastic deformation; the resistance of Ti to fatigue may be measured by the rotating cantilever beam test, during which the strains are predominately elastic, both upon initial loading and throughout the test, in accordance with the ASTM E466 standard. Grade 4 Ti material has a fatigue limit ~

310 MPa [89]. During our experiments, we did not experience the plastic properties of Ti during fatigue, in addition, there was no force that would exceed the yield strength of Ti, according to our FEA results. However, it has been described by Zhao et al. that low-velocity impact damages could decrease the compressive failure strength of Ti honeycomb sandwich structures by up to 15% [90]. While this was not assessed in our tests, but fatigue due to cyclic loading may cause changes in the material yield stresses and stress concentrations, in which case the plastic property of the Ti alloy theoretically could have had an impact on the results.

Of note, the widely accepted view that inadequate occlusion may lead to biological complications is poorly reported in the literature, and it is difficult to demonstrate that the consequences are pathognomonic for the presence of overload [91]. Even with the most modern digital technologies, the applied tests cannot give a reproducible, quantifiable absolute or relative value [92]. In terms of clinical relevances, single crown or bridge on the implants may have different consequences. This can also cause rotation in the case of crown or restoration with cantilever. After the static load test, we also performed a high load measurement in each group, where the implant abutment contact was loaded with a 2 kN force. thereafter these measurements was no detectable extension torque in the fixing screw, because the degree of compression was greater than the thread height of the fixing screw. Because of this, the connection of the implant to the abutment was not damaged, and the implant did not break.

The mechanical stability of the implant-abutment interface is one of the most important factors for long-term successful implant restorations. It has been reported in the literature that the tapered connection is the most reliable for dental implants. However, there is little evidence on how the mechanical properties of the implant-abutment contact are influenced by the small or large taper angle of the contact and the quality of the Ti material used for the abutment. Therefore, the aim of our mechanical investigations was to simulate the effect of chewing forces on implant-superstructure models with different taper angles and to investigate their mechanical stability in order to assess which tapered connection represents a long-term, successful and safe solution for small diameter implants. Implant-abutment screw loosening leads to denture failure in the short term and implant failure in the longer term, and we investigated how the torque values of the screw torque change under load for different tapered-angle implant-abutment models. For which taper-angle connection the highest torque values are retained, i.e. for which one long-term success is likely.

Based on the FEA methodology and models, many studies showed that implant geometry, bone quality, and site of implant placement affect load transmission mechanisms, thereby subsequently also affecting peri-implant bone resorption [93]. Maximum stress areas may be located at the implant neck, and possible overloading could occur in the form compression in the compact bone (due to lateral components of the occlusal load) and in the form tension at the interface between cortical and trabecular bone (due to vertical intrusive loading components) [93-95.]. Load transmission may also occur in the abutment-implant interface zone, which may also lead to the above mentioned phenomenon [96-98]. Our FEA analyses also confirmed that the implant-abutment connection greatly influences the distribution of forces in different ways at different heights between the implant and the bone. In the FEA, the mechanical stress was better distributed over the entire surface of the implant in the case of the 90° implant-abutment connection, compared to the 24° connection. Regarding the 24° connection, the mechanical stress was greatest in the area where the cones meet, which represents the part of the implant with the smallest wall thickness; in addition, in the case of the 24° model, not only vertical but also horizontal deformation occurs. This horizontal deformation may lead to peri-implant bone resorption in the cortical bone. When the taper angle is increased, more of the load is transferred to the implant wall than to the fixing screw. For this reason, the higher load on the smaller taper angle resulted in greater screw loosening, as more force is transmitted to the screw.

The formation of biological width after implant placement is an important factor in the prevention of peri-implant bone loss [99]. Adaptation and remodeling of these soft tissues may have considerable roles in facilitating secondary stability and the long-term survival of implants, as they absorb forces that act on the implant, thus reducing the transmission of forces acting on the jawbone. The viscoelastic absorbent properties of soft tissues around the implant have been described, both at the cellular [100] and tissue levels [101], respectively. The forces acting on the dental prosthesis are distributed and continue to affect the superstructure, the implant, the implant connection and the bone. Through the implant-abutment connection, which mentioned may cause loosening of the fixing screw, irreversible vertical compression and overloading of the bone in different ways at different heights depending on the design. Manufacturing inaccuracies in the implant-abutment connection may cause both mechanical and biological problems.

In order to achieve long-term implantation success – from a mechanical point of view – it may be crucial that the prosthetic phase takes place at the implant or abutment levels. Important examples from clinical practice may include situations with a large axis deviation, where the contact part of the abutment is reduced in the dental technique phase in order to facilitate the placement of the restoration. However, this may have a detrimental effect on the fit between the implant and the abutment, thereby changing the distribution of masticatory forces between the implant body and the surrounding bone; in many cases, this may lead to the breakage of the retaining screw of restoration or the fracture of the bridge itself. The implant-abutment relationship affects the indications in which the implants may be used. It is extremely important that in the case of special indications (immediate implantation, immediate loading, or cracking technique) that the implant-abutment connection should serve the best possible force distribution on the surface of the implant, i.e. transfers the masticatory force to the bone on the largest possible surface.

In the study of Paepoemsin et al., the removal torque of three different types of abutment screws were evaluated after mechanical cyclic loading. In their paper, flat head and tapered screws were used, and after the first 10 min of dynamic load and after 1 million cycles, they compared the reverse torques of each group. Similarly to our results, they have shown a statistically-significant decrease in the measured reverse torque. Our results also indicated that during the dynamic load test, there was a change, which decreased the tight of the grip of the abutment head into the implant [102]. Benjaboonyazit et al. also studied the emergence of loose connections due to fatigue. In their experiments, they used 3.75 mm Octatorx-cone implants and tightened the screws with 30 Ncm force, after which, a very long, 2 million cycle dynamic load test was performed. Their results were also consistent with our findings, i.e. without any load, they obtained reverse torque values over 27 Ncm, which decreased to less than 16 Ncm after their fatigue test protocol. However, our study showed that this decrease of the reverse torque may be moderated with an increased conical angle. Comparing the static 55° results, which resulted in 24-25 Ncm reverse torque, after the dynamic load test, the 60° case decreased to very similar values, i.e. around 24 Ncm. These result also highlighted that increasing the conical angle indeed helps maintaining a stronger grip for longer periods even in the case of periodical loads [103]. Joo-Hee and Hyun-Suk performed cyclic load tests as well on Grade 4 Ti implants with an external hex connection. To follow the International Organization for

Standardization (ISO) protocol, they loaded the implants in a 30° angle, and after a 1 million cycle of 300 N (which is equivalent to 30 kg) force, they measured the reverse torque at 15.2 Ncm; they also determined the torque values before the dynamic test, which was 25.2 Ncm. Their results coincide with the findings of the present study, as in their case, the decrease in the reverse torque was roughly 40% compared the pre- to post-dynamic test [104]. The above referenced articles support the results of our tests, i.e. under loading, the greater the angle of the implant-superstructure-connection was, the smaller the amount of screw loosening could be measured.

In summary, according to our mechanical tests, the amount of fixing screw loosening changes significantly with the change in the taper angle of the connection. The larger the angle of the implant-superstructure connection, the smaller is the screw loosening due to loading.

As a result of our finite element analyses, it can be concluded that in the case of connections with a smaller taper angle, the masticatory forces acting on the implants was concentrated on the upper third of the implant-body; this may result in horizontal, irreversible deformation in the implant neck, which could lead to increased risk of cortical bone resorption. While in the case of a larger taper angle, the masticatory force is evenly distributed over the body of the implant.

Long-term and safe rehabilitation of edentulous patients still has numerous challenges, due to the numerous variables that affect implant survival and patient satisfaction; thus, all advances aiming for the optimalization of prosthodontic treatment may have considerable real-world implications for clinical practice. Our studies aimed to highlight the importance of the implant-abutment relationship – that is, the effect of conical angles and the Ti grade of the implants – in the context of screw loosening and irreversible deformation, both being detrimental for durable restorations. Overall, it was demonstrated that increasing the taper angle of the connection has an inverse relationship with screw loosening; these findings were further supported by finite element simulations, suggesting that the accumulation of masticatory forces, which may result in horizontal deformation in the implant neck, is more likely for smaller angles. To ensure high secondary stability and clinical satisfaction, judicious treatment planning is critical, which includes implant design. Therefore, further studies to confirm and complement the existing body of evidence is definietly warranted.

VIII. NEW FINDINGS

a. **Conical angle, but not implant material affected implant behaviour and reverse torque values under static loading:** under a static compressive load of 500 N, implant deformation (numerical, but no significant difference), resilience (significant differences) and dissipated energy (significant differences) decreased consistently, while values of the reverse torque, increased consistently (significant differences) with increasing conical angles. Implant composition (titanium Grade 4 vs. 5) had no significant effect on vertical or horizontal deformation or reverse torque values.

b. **Conical angle affected implant behaviour and reverse torque values under dynamic fatigue:** during dynamic fatigue tests, vertical compression (significant differences) and irreversible deformation (significant differences) was highest for the 35° conical angle, but decreased consistently, while values of the reverse torque increased consistently (significant differences) with increasing conical angles.

c. **Higher conical angles resulted in lower mechanical stress values and more advantageous stress distribution in our finite element model:** during FEA, the calculated von Mises stress for the 24° implant was considerably higher than in the 90° case (highest stress values: 300 MPa vs. 160 MPa). A horizontal barrel-shaped deformation was observed in the model with the 24° conical angle connection, which was not shown for the 90° case.

IX. SUMMARY

During our static load tests no statistically significant differences were found between the Ti Grade 4 and Grade 5 implants having the same conical angle in the reverse torques upon disassembling implant abutment and implant bodies.

The smaller the angle of the connection between the implant and the abutment, the more the implant and the superstructure move together under the same force, which was directly proportional to the decrease of the reverse torque values for both materials.

Based on our findings, for small diameter implants (Ø 3.3 - 3.4 mm), it is recommended to use higher conical angle connection to avoid larger deformations in lengths and diameters of the implant at the connection and essential torque reduction of the fixing screw.

According to our mechanical tests, the amount of fixing screw loosening changes significantly with the change in the taper angle of the connection. The larger the angle of the implant-superstructure connection, the smaller is the screw loosening due to loading.

As a result of our finite element analyses, it can be concluded that in the case of connections with a smaller taper angle, the masticatory forces acting on the implants was concentrated on the upper third of the implant-body; this may result in horizontal, irreversible deformation in the implant neck, which could lead to increased risk of cortical bone resorption. While in the case of a larger taper angle, the masticatory force is evenly distributed over the body of the implant, this is also determined by a number of other factors, including the thread design, shape, length and diameter of the implant.

Our results may contribute to the understanding of the long-term success of dental implants.

X. ÖSSZEFOGLALÓ

Statikus terheléses vizsgálataink során nem találtunk statisztikailag szignifikáns különbséget az azonos kúpszögű titánium Grade 4. és 5. alapanyagokból készült implantátumok között a felépítmény és az implantátumtestek szétszerelésekor fellépő kihajtási nyomatékértékek között.

Minél kisebb volt az implantátum és a felépítmény közötti kapcsolat szöge, annál nagyobb mértékben mozdult együtt az implantátum és a felépítmény azonos erő hatására, ami egyenes arányosságot mutatott a kihajtási nyomatékértékek csökkenésével mindkét anyag esetében.

Eredményeink szerint a kis átmérőjű implantátumok esetén (Ø 3,3 – 3,4 mm) a nagyobb kúpszögű implantátum-felépítmény kapcsolat alkalmazása ajánlott, hogy elkerüljük az implantátum hosszának és átmérőjének nagyobb maradandó deformációját, valamint a rögzítőcsavar lényeges kihajtási nyomatékcsökkenését.

Mechanikai vizsgálataink eredményeként megállapítható, hogy a rögzítő csavar kilazulásának mértéke jelentősen változik a csatlakozás kúpszögének változásával. Minél nagyobb az implantátum-felépítmény csatlakozás szöge, annál kisebb a terhelés miatti csavarlazulás mértéke.

Végeselemes analízisünk eredményeként megállapítható, hogy a kisebb kúpszögű kapcsolatok esetében az implantátumokra ható rágóerők az implantátum-test felső harmadára koncentrálódtak; ez az implantátum nyakában horizontális, irreverzibilis deformációt eredményezhet, ami a kortikális csont reszorpciójának fokozott kockázatához vezethet. Míg a nagyobb kúpszögű kapcsolatokkal rendelkező implantátumok esetén a rágóerő egyenletesen oszlik el az implantátum testén.

Eredményeink hozzájárulnak a fogászati implantátumok hosszú távú sikerességének megértéséhez.

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