RAMESH CHOWDHARY
ON EFFICACY OF IMPLANT THREAD DESIGN FOR BONE STIMULATION

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Malmö University, 2014
Faculty of Odontology
Department of Prosthodontics
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Dedicated to
my father, mother,
my beloved wife, Nagu,
and my two boys, Rohan and Rishab
This thesis represents number 45 in a series of investigations on implants, hard tissue and the locomotor apparatus originating from the Department of Biomaterials, University of Gothenburg and the Department of Prosthodontics/Material sciences, Malmö University, Sweden.


33. **Jonas P Becktor DDS, 2006.** On factors influencing the outcome of various techniques using, endosseous implants for reconstruction of the atrophic edentulous and partially dentate maxilla. Thesis defended 17.11.2006. External examiner: Professor K.F. Moos.


List of Papers


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ABSTRACT

Introduction – The mechanism and efficiency of force transfer by dental implants to surrounding biologic tissues are important determinants in the development of the implant-to-bone/tissue interface and implant longevity. Threads are used to improve the initial stability by maximizing bone contact through an enlarged implant surface area and thereby favor distribution of interfacial stresses. However, knowledge about optimal thread design for an enhanced implant integration in bone tissue is still lacking.

Aim – The aim of this thesis was to evaluate the efficacy of implant micro thread design when combined with macro threads, for bone stimulation. The hypothesis is that the short threads will contribute with compression forces that may stimulate bone healing, while the larger threads will provide with primary stability necessary during the healing process. A further aim was to use an FEA model to describe the optimal thread form for reduced stress concentration immediately after implant insertion as well as after completed bone healing.

Materials and methods – In study 1, Two-dimensional finite element models were made from 8 different thread designs. The crest module and apex of the implants were removed from the implant models, in order to enhance the effect of the thread designs only. Thus, the suprastructures and microstructures of the implants were not considered. All the eight implant models were assumed to be embedded in cortical bone. In addition, a 3D model was used to evaluate stress in the bone generated by 6 different thread designed implants when the
Implant models included the entire implant. In the *In vivo* studies 2 and 3, experimental turned implants with a diameter of 4mm and 8mm in length were prepared with micro threads in between macro threads along the body of the implants. These were used as test implants. Implants without micro-threads were used as controls. In study 4, similar implants were made but with alteration in depth of the macro-thread to improve the possibility for bone stimulation by compression during healing. Insertion and removal torque analysis along with histomophometric analyses were done to evaluate the bone response.

**Results** – In study 1, stresses were calculated using von Mises stress analysis. The stress levels in the bone were in the range of 5-13 MPa in osseointegrated model and 14-107 MPa in immediate 2D models. 3D Analysis results showed the von Mises stress in the range of 4.8-30.9 MPa, when a load of 100N was applied vertically. In Study 4 FEA demonstrated stress levels in the range of 0.28 MPa to 62MPa for the control implant model designs, whereas the test implant models displayed a range of 0.28 MPa to 31Mpa. In study 2, the mean values of the ITQ for the control and test groups in the tibia were 15 and 20 Ncm respectively, and in the femur, the values were 11 and 12 Ncm, respectively. In study 4, the ITQ values were 11Ncm and 14 Ncm respectively in the tibia, and in the femur 13 Ncm and 19 Ncm respectively. The RTQ values for the control and test groups in tibia was 11Ncm and 17Ncm, respectively and in the femur, 13Ncm and 23Ncm, respectively. The histomorphometric analysis of study 3, showed the mean total bone area, BA% (SD) to be higher in the test implants, when compared to the control implants in both the tibia 24 (4), and 21(4), the femoral bone 29 (5), and 25 (7), respectively with no statistical significance. In study 4, the total bone area BA% was higher for the test implants with a mean value of 72% compared to 48% for the control group in tibial bone. In femural bone, the bone area was 63% for the test and 38 % for the control group implants with *p* value of 0.10 for both tibia and femur. Bone to implant contact showed significantly higher value for the test implants in the femur, *p* = 0.04.
Conclusion – The impact of different thread designs, with respect to the magnitude of the transferred stress peak in the bone, was higher for the immediately inserted implants than for the osseointegrated implant model. The stress distribution was more effective in experimental micro-thread implant models, when compared to the non-micro thread models. The addition of pitch shortened threads in the test implant, did significantly improve the primary and secondary stability of the test implants, when mechanically evaluated with ITQ and RTQ analysis in corticular or trabecular bone rabbit bone. Histomorphometrical analysis showed that the addition of the pitch shortened threads in between the macro threads did have a bone stimulatory effect in the femur of the rabbits.
INTRODUCTION

Dental Implants
In the past 40 years, the use of dental implants has become a widely accepted treatment alternative for treating fully or partially edentulous patients, and is regarded as therapeutic alternative to tooth supported fixed restoration or removable dentures. Excellent long term results and high success rates have been achieved using dental implants during the past decades.

Clinical follow-up studies show that survival rates around 95% can be expected on most indications over a 5-year period time. From the literature, it is also evident that some of the risk factors which may led to implant failures are soft bone, limited bone volumes, grafted bone, overload and smoking. In addition, there has been a gradual challenge of biologic limits for osseointegration; implant healing times have been shortened, implants are placed in extraction sockets and immediate loading protocols have been introduced. Whereas these approaches have certainly widened indications for osseointegrated implants and dramatically shortened the treatment time, they have at the same time presented increased risk for failure, at least for minimally rough surfaced implants placed in soft bone qualities. The higher failure rates may be overcome by optimizing the implant surface and design to promote bone integration and stability. One such change could be surface and design modifications to promote bone integration and stability. Numerous experimental studies have found a stronger bone tissue response and increased stability for moderately rough surfaced implants as compared to smoother control implants. As a consequence, most implants
systems of today have a moderately rough surface as opposed to the original Branemark turned and minimally rough surface.\textsuperscript{13} Macroscopical modification of the implant is another theoretical possibility to improve integration, stability and clinical function.\textsuperscript{14} Most dental implant system have some kind of interlocking geometry at the apical aspect of the implant where bone ingrowth may help stabilizing it.\textsuperscript{15} However, it would probably be even more beneficial if macroscopic modification were performed and the consequences on implant stability in different clinical situations understood. A third way of improving the healing of implants would be to use bone growth factors.\textsuperscript{16} This would be especially beneficial in situations of placing implants in defects and tooth extraction sockets in order to facilitate bone growth towards initially uncovered implant surfaces. It is possible that the combination of bone growth factors and implant surface modification may amplify the bone tissue response due to the previously demonstrated stronger bone reactions to moderately rough surfaces.

**The bone tissue response to titanium implants**

Titanium is the first element of group 4A in the periodic table; its atomic number is 22 and atomic weight 47.9. About 0.2 atomic % of the earth crust consist of titanium and titanium is commonly used as a construction material in air, space, marine and chemical industry. Titanium is suitable for machining and has a high strength/weight ratio, corrosion resistance in saline solutions and a resistance to acids. Titanium has a well documented biocompatibility in bone and soft tissues, as it does not provoke any negative tissue reactions. The reasons may be due to its ability to rapidly form a few nanometers thick oxide film in air, which is of highly protection nature. The oxide layer has hydrophilic properties and a dielectric constant close to water, which may facilitate the interaction with bioliquids and biomolecules.

**Integration of turned titanium implants in bone**

A direct contact between bone and screw-shaped titanium implants was first described by Brånemark et al, (1969)\textsuperscript{17} and later by Schroeder et al, (1976)\textsuperscript{18} Since then, numerous reports have demonstrated a direct- bone implant contact for clinically retrieved implants.\textsuperscript{19,20}
Sennerby studied the healing process around screw-shaped machined titanium implants in cortical bone after 3 to 180 days and found that it was characterized by an elderly cellular response, a relative absence of inflammatory cells and a rapid formation of woven bone from the endosteal surface. Seven days after implantation, solitary bone formation was observed in the threads and at the endosteal surface of the cortical bone. With time both types of woven bone fused and increasingly filled the implant threads. The increased bone-titanium surface contact was thus a result of ingrowth of bone from the surroundings and did not start at the implant surface. The final mineralization of the interface bone could have been an acellular process scattered crystals of mineral fused slowly and formed the interface bone. The implant surface not covered by bone was covered by multinuclear giant cells as also reported by Piatelli et al. This phenomena was earlier found by Donath et al, to correlate with the roughness of the implant surface. Johansson & Albrektsson demonstrated an increased resistance to removal torque with healing time. Moreover, more bone-implant contacts were seen with time, which could explain the observed stability increase. Sennerby et al, found differences in implant integration in cortical versus cancellous bone. Differences in removal torque for tibial and femoral inter-articular implants were believed to depend on differences in mechanical properties of the surrounding bone tissue, which may depend on the type of bone and the degree of maturation of the bone tissue. The amount of compact bone in contact with implant correlated with the removal torque. Ultra-structural studies of the bone-titanium interface have usually demonstrated that there is an unmineralized or partly mineralized zone separating the titanium surfaces from the bone. For instance, Sennerby et al, described an amorphous layer separating the mineralized bone and the titanium surface. This zone lacked collagen and mineral. Rupture by unscrewing during removal torque occurred in the innermost amorphous zone. The stability of threaded machined titanium implants was therefore believed to be due to a mechanical interlock with surface irregularities at the microscopical level and/or geometrical deviations at the macroscopical level.
Biomechanics and Implants
Functions of Implants (Force Transfer).

Dental implants are subjected to forces of the muscles of mastication. Parafunctional habits (clenching, bruxing, or engaging) can transmit forces to the supporting bone that could result in destructive lateral stresses and overload.\(^3\) Such loads can vary dramatically in magnitude, frequency, and duration, depending on the degree of the patient's parafunctional habits. Bragger \textit{et al}, indicated that all of the technical complications in his study were associated with bruxism.\(^3\) Therefore, anticipated occlusal and chewing force should be considered.\(^3\) It has been suggested that most nocturnal parafunctional habits can be ameliorated by acrylic resin night-guards.\(^3\) The maximum bite force in parafunction is much greater than the natural forces attained during mastication. In normal dentition without implants, mean maximal vertical (axial) bite force magnitudes in humans can be 469-485\,N at the region of the canines, 583-599\,N at the second premolar region, and 723-738\,N at the second molar.\(^3\) In general, maximal bite force in medial and posterior directions was larger than that in corresponding lateral and anterior directions.\(^3\) Braun \textit{et al}, also showed the mean maximum bite force to be 738\,N with a standard deviation of 209\,N. The mean maximum bite force as related to gender was found to be statistically significant, whereas the correlation coefficients for age, weight, stature, and body type were found to be low. Similar results were also reported by Raadsheer \textit{et al}, who showed average values of the maximal voluntary bite forces as 545.7\,N in men and 383.6\,N in women.\(^3\)

Morneburg and Proshchel measured vertical masticatory forces in vivo using a method that should be insensitive to the location of bite force impact. Their results on implant-supported fixed partial dentures produced a mean total masticatory force of 220\,N with a maximum of 450\,N. The single abutments experienced mean loads of 91\,N (anterior) and 129\,N (posterior) with a maximum of 314\,N.\(^3\) Fontijn-Tekamp \textit{et al}, reported similar results.\(^3\) They showed patients with mandibular implant-retained overdentures had maximal unilateral bite forces in men and women ranging from approximately 50 to 400\,N in the molar regions and 25 to 170\,N in the incisor areas.
Impact of occlusal overload.
Dental implants could be subjected to initial loads during the healing stage. These passive mechanical loads are attributed to mandibular flexure, contact with the first stage cover screw, and/or second-stage permucosal extension. The force of the tongue and oral musculature can generate low loads on implant abutments; however, these could be of greater magnitude with para-functional oral habits or tongue thrust. Nonpassive prostheses applied to implant bodies during the healing phase could result in mechanical loads applied to the abutment, even in the absence of occlusal loads. Once the dental implant is loaded, the bone reacts to micro strain in different ways. Frost proposed the theory of mechanostatics and postulated that bone mass is a direct result of the use of the skeleton. Frost established a mechanical adaptation chart relating trivial loading, physiological loading, overloading and pathologic loading zones to ranges of microstrain. Cell destruction can be seen when the chewing forces are such that they exceed physiological tolerances. This overload zone is reached when microstrains are greater than 3000 units. Excessive occlusal force being generated presents an opportunity for loosening and/or fracture of the implant through bending overload. Bending moments resulting from nonaxial overloading of dental implants can cause stress concentrations exceeding the physiological supporting capacity of cortical bone, leading to an overload situation and various kinds of failures.

It has been stated that overloading of an oral implant can result in loss of the marginal bone or even complete loss of osseointegration. Initial breakdown of the implant tissue interface generally begins at the crestal region in successfully osseointegrated endosteal implants, regardless of surgical approaches (submerged or non-submerged). Early crestal bone loss on average of 1.5 mm is often observed after the first year of function followed by an annual bone loss after the first year of less than 0.2 mm. Six plausible etiologic factors are hypothesized, including surgical trauma, occlusal overload, perimplantitis, microgap, biologic width, and implant crest module. Based on currently available literature, the reformation of biologic width around dental implants, microgap if placed at or below the bone crest, occlusal over-load, and implant
crest module could be the most likely causes of early implant bone loss. Furthermore, it is important to note that other contributing factors such as surgical trauma and peri-implantitis could also play a role in the process of early implant bone loss.\textsuperscript{45} Implant overload can be caused by a multitude of factors, including sub-optimal implant design and size, an insufficient number of implants to support the restoration, improperly splinted abutments, violation of conventional prosthetic limitations for natural dentition, excessively cantilevered pontics; splinting to natural dentition (even with a stress-breaking attachment), improperly positioned implants, the wrong type of restoration for the clinical condition, loss of supporting bone, excessive para-functional forces, and non maintenance of the components.\textsuperscript{42}Crestal bone loss as well as loosened screws is frequently the first detectable signs of implant overload and warrant immediate action. To eliminate or reduce the excessive stress at the crestal bone implant interface is a goal accomplished by balancing the whole arch, reducing occlusal contacts to the area of the implant supported regions, and shortening or eliminating cantilevers whenever possible. This uniquely specific occlusal philosophy of endosteal implant prostheses has been referred to as “implant-protected occlusion.”\textsuperscript{46} Implant orientation and the influence of load direction, the surface area of implants, occlusal table width, and protecting the weakest area are blended together from a biomechanical rationale to provide support for a specific occlusal philosophy.\textsuperscript{46}

Appropriate implant design selection through careful treatment planning is imperative to lower the magnitude of loads imposed on the vulnerable implant-to-bone interface. As the implants function to transfer occlusal loads to the surrounding biologic tissues, functional design objectives should aim to manage biomechanical loads (through dissipation and distribution) to optimize the implant supported prosthesis function.\textsuperscript{47}

The mechanism and efficiency of force transfer by dental implants to contiguous biologic tissues are clearly important determinants in the development of the implant-to-bone/tissue interface and implant longevity. Whether a clinician seeks to gain a better understanding of
implant design rationale and/or to implement biomechanics concepts in patient care, a fundamental, yet clinically relevant, understanding of biomechanics is required.

The distribution of force with natural teeth depends on micromovement induced by the periodontal ligament. Force distribution to the osseointegrated implant interface differs from natural teeth in that implants do not have micromovement associated with force distribution. Because of the lack of micromovement of implants, most of the force distribution is concentrated at the crest of the ridge. Vertical forces at the bone interface are concentrated at the crestal regions, and lateral forces increase the magnitude of the crestal force distribution.

Prudent control of the biomechanical load on dental implants is imperative to achieve long-term clinical success. The design of the final prosthesis and location of force transmission have a definitive influence on the quantification of induced strains and load partitioning among implants. Alterations to the implant crown location and cusp inclination are suggested to limit implant overload. Cehreli et al, who revealed that offset loading increases the magnitude of strains when compared with axial loading. In addition, they believe it is a multifaceted phenomenon when they examined how force has been distributed around implants. This includes, but is not limited to, implant location, angulation, and design of prosthesis.

Biting Forces Generated on Implants

Dental implants are subjected to occlusal loads when placed in function. Because loading could be a determining factor in the treatment outcome, there is a need to further increase our understanding of the biomechanics of oral implant design. The greatest natural forces exerted against teeth, and thus against implants, occur during mastication and are the result of the action of masticatory muscles. The force is related to the amount and duration of function. Forces acting on dental implants possess both magnitude and direction and are referred to as vector quantities. The maximum bite force decreases as muscle atrophy progresses throughout years of edentulousness. However, this force can increase in the years after implant placement. The muscle strength, masticatory dynamics,
and maximum bite force exhibited by adults could all be influenced by sex, muscle mass, exercise, diet, bite location, parafunction, state of the dentition, physical status, and age.\textsuperscript{53-56} It has been found that a vertical force in the long axis of an implant of only 10 mm in length with a 4-mm diameter can accept average maximum biting forces to the supporting bone within the physiological strain limits of the bone.\textsuperscript{57} The most common cause of early loss of rigid fixation during the first year of implant loading may be the result of parafunction.\textsuperscript{58, 59} Para-functional bruxism or clenching often leads to hypersign elements that could affect the force transfer and the quality of osseointegration.\textsuperscript{59}

**Design Elements Affecting Force Transfer and Osseointegration Quality**

**History of Design**

The three-dimensional structure of the implant, with all the elements and characteristics that compose it, is referred to as the implant design. The type of prosthetic interface, the presence or absence of threads, additional macro irregularities, and the shape/outline of the implant are considered some of the most important aspects of implant design. Dental implants can be categorized into threaded and non-threaded, cylindric, or “press-fit” designs. Implant companies have been using a plethora of additional features to accentuate or replace the effect of threads. These features include vents, grooves, flutes, indentations, and perforations of various shapes. Implants can be hollow or solid, with a parallel, tapered/conical, or stepped shape/outline and a flat, round, or pointed apex.\textsuperscript{46}

**Methods for the Assessment (Mechanical Tests, Histology)**

The effects of implant design elements on the quality and strength of osseointegration can be validated through several assessment methods. To prove the value of one design feature over another, outcome variables such as histologic assessments and mechanical testing have been the main methods used.

There are generally 3 types of biomechanical tests: pull-out, push-out, and torque measurement. Push-out and pull-out tests are indicated with cylindric or press-fit implants, whereas threaded implants are more effectively tested with the counter- torque or reverse-torque
test. Thread volume fill and number of cells in contact with the implant surface are two other variables frequently reported in histomorphometric studies. With reference to an implant’s ability to transfer occlusal stresses at the interface, a common method of testing the mechanical performance is by using a pull-out or push through test. The results of such tests are often reported in terms of holding power or shear strength. Implants in a screw shape will resist push-out, but this shape requires a biomechanical interlocking if it is to resist torsion, which tends to loosen the implant along the threads. Removal torque testing measures the resistance to loosening of the implant, not necessarily to occlusal load conditions. Push-out and removal torque force are both proportional to the interfacial shear strength. The interfacial strength ranges from 9 to 19 MPa for cortical bone and 2 to 6 MPa for trabecular bone after 12 weeks of insertion in various animal models. However, these results showed that surface roughness and texture could be a primary factors in the anchorage of dental implants. The interfacial tensile strength of titanium was reported to be 2.5 MPa on average, whereas hydroxyapatite- coated titanium might not be much better. On the other hand, several other studies demonstrated that hydroxyapatite (HA) is superior to sand- blasted, large-grit, acid-etched (SLA), titanium plasma spray (TPS) and/or machined surfaces, even if one explanation to this fact could be the greater press fit achieved with plasma sprayed HA-coated implants that are hundreds of microns wider than non HA coated devices.

Research of thread shape has been performed exploring the use of orthopedic bone screws placed into cadaveric bovine or canine long bones. In general, the holding power is a function of the test configuration, cortical thickness and the number of threads engaged, properties of the bone adjacent to the screw thread, and the major diameter of the screw. Reports differ regarding the relative importance of the thread profile. Koranyi et al, found little difference between thread types, whereas others stated that the interface strength does depend on thread shape.

It has been shown that roughened titanium surfaces can improve the clinical prognosis of implants by achieving a higher percentage of
bone implant contact and higher removal torque values in mechanical testing as compared with the traditional, smooth titanium surface. However, high surface roughness alone is not the only criteria to consider for optimal osseointegration, and the scope of this thesis does not focus on surface roughness. The following section reviews available studies addressing the effects of implant shape, diameter, length, and thread design on osseointegration of the bone to implant interface.

Effects of Shape, Diameter, and Length
The macro design or shape of an implant has an important bearing on the bone response; growing bone concentrates preferentially on protruding elements of the implant surface such as ridges, crests, teeth, ribs, or the edge of threads, which apparently act as stress risers when load is transferred. The shape of the implant determines the surface area available for stress transfer and governs the initial stability of the implant. Finite element analysis studies of implants indicate that bone stress distributions and magnitudes vary with implant shape. It is known and accepted that bone respond differently to different types of loading and is weakest under shear loading conditions. Bone is strongest when loaded in compression, 30% weaker when subjected to tensile forces, and 65% weaker when loaded in shear. Transforming shear forces into more resistant force types at the bone interface is the purpose of incorporating threads into the implant design as a surface feature. Smooth cylinder implant bodies result in essentially a shear type of force at the implant to bone interface. Surface features such as threads are incorporated into the design to transform the shear loads to more resistant force types. It is for this reason that, most implant designs are threaded, because the thread shape is particularly important in changing force at the bone interface. The available thread shapes of square, V-shaped, and reverse buttress are intended to reduce the development of shear at the dental implant tissue interface so as to improve long-term success.

The significance in increased implant length for the ability to achieve osseointegration is not found at the crestal bone interface, but rather in initial stability and the overall amount of bone–implant interface.
The increased length can provide resistance to torque or shear forces when abutments are screwed into place. However, the increased length does little to decrease the stress that occurs at the transosteal region around the implant at the crest of the ridge or change its ability to achieve osseointegration. In general, the use of short implants has not been recommended because it is believed that occlusal forces must be dissipated over a large implant area for the bone to be preserved. Less favorable success rates for shorter implants were observed in clinical studies. By the use of engineering statics, the amount of force transferred to the crestal bone from a horizontal occlusal load, relative to the implant length necessary to support that load, was analyzed for hypothetical instances when the implant was embedded in a uniform mass of bone and when it was bi-corticated. The analysis revealed that implants longer than 12 mm will not significantly reduce force transfer proportionately to the increased length. On the other hand, Bruggenkate et al evaluated short ITI implants of 6 mm and reported a cumulative survival rate at 6 years of 94%. This study recommended that short implants should be used in combination with longer implants, especially when used in the less dense bone that is often seen in the maxilla.

The surface area of each implant support system is directly related to the width and height of the implant. Wider root form implants have a greater area of bone contact than narrow implants of similar height and design resulting from their increased circumferential bone contact areas. As a result that occlusal load to the implant causes most stress at the crest of the ridge, the crest of the ridge is where initial bone loss occurs. Consequently, width may be more important than length of the implant once a minimum or optimal height of more than 8 mm has been obtained for initial fixation and resistance to torque.

The posterior regions of the mouth can sustain greater forces, yet often present poorer bone density. The longest implants are typically inserted into the anterior regions of the mouth, where forces of less magnitude and superior bone quality are normally present. A common biomechanical approach, often presented to decrease risk factors in such regions, is to increase implant surface area primarily by focusing on diameter. However, this increases surface area by
only 30% for conventional thread designs despite the fact that forces increase by 300% in the posterior regions. A change in implant diameter and thread design could increase surface area by 300%. Such increases in surface area could decrease stresses to the crestal bone regions and reduce both crestal bone loss and early loading implant failure.\textsuperscript{81,82}

Morrist \textit{et al}, studied the influence of implant diameter and length on implant success rate.\textsuperscript{83} Their results on 3-year survival and stability of various implant lengths (7 mm, 8 mm, 10 mm, 13 mm, and 16 mm) and diameters (3-3.9 mm and 4-4.9 mm) were 90.7% for the 3 to 3.9 mm diameter and 94.6% for the 4 to 4.9 mm group. Survival ranged from 66.7% for the 7-mm implants to 96.4% for 16-mm implants. Overall, the shorter implants had statistically lower survival rates as compared with longer implants, and 3- to 3.9-mm diameter implants had a lower survival rate as compared with 4- to 4.9-mm implants. However, there was no significant difference in crestal bone loss for the 2 different implant diameters between placement and time for abutment operation.\textsuperscript{83}

\textbf{Effects of Thread Design/Geometry.}

Threads are used to maximize initial contact, improve initial stability, enlarge implant surface area, and favor dissipation of interfacial stress.\textsuperscript{46, 84} Thread depth, thread thickness, thread face angle, thread pitch, and thread helix angle are some of the varying geometric patterns that determine the functional thread surface and affect the biomechanical load distribution of the implant.\textsuperscript{85} The influence of the threads can be easily understood as the greater the number of threads that are present as well as the greater the depth of the threads, the more functional surface area that is available.

Thread shapes in dental implant designs include square, V-shaped, and buttress. The face angle of the thread can change the direction of load from the prosthesis to a different force direction at the bone. Under axial loads to a dental implant, a V-thread face is comparable to the buttress compaction created by the tapping procedure. Functionally, there is always applied force acting on bone modified by an implant design, and there is always resisting
force acting on the implant through the viscoelastic properties of trabecular structure. Through biomechanical events in bone, osseous tissue can be stimulated within physiological limitations by implant design to develop along the lines of compressive forces dependent on the implant load-bearing area to sustain equilibrium. A sinusoidal thread design (eg, LaminOss immediate-load implant; Impladent Ltd., Holliswood, NY) has shown, in animal histologic observation, lamellar bone achieved by the function of osteocompression. This implant design allows bone to be molded and compacted circumferentially. Future human clinical research would be necessary to provide a meaningful statistical analysis to validate the importance of this implant design and the function of osteocompression. Hoshaw conducted an experimental animal study to examine the bone modeling and remodeling response of interfacial tissues near screw-shaped v-threaded titanium dental implants (Nobel Biocare, Branemark) under different mechanical loading conditions. The study has shown that threaded implants had higher remodeling rates and less mineralized bone formation when loaded with axial force than the non loaded threaded implants. The general elevation in bone modeling and remodeling activity observed near loaded implants supports Frost’s mechanostat theory and is consistent with the idea that these responses were triggered by tissue micro damage as a direct consequence of the applied loads. They speculated that this cortical bone remodeling around the implants could be an attempt to provide an improved strain orientation of the bone to loads. Square-thread design implants showed, in a 4 beagle dog animal study, that bone grew between the threads, closely adapted to the implant, and that the inferior aspect of the test implant threads showed greater bone to implant contact than the coronal aspect. These results suggest a biologic advantage for the compressive load transfer mechanism for this thread design. This observation was further supported in a case report. As a result of limited sample size, additional histologic reports and clinical data are required to confirm the observations made in this study. However, limited information is currently available in examining the influence of the various thread design and their effect on bone stimulation, and thus enhancing osseointegration.
AIMS

The main aim of this thesis was to know the efficacy of implant thread design for bone stimulation. Hence, the studies included were designed to investigate the following aspects:

**In vitro**

1. To investigate the characteristics of different thread designs, and to determine optimal thread form configuration in an experimental simulation model, which can reduce the stress concentration in the bone immediately after placement and after osseointegration.

**In vivo**

2. To biomechanically evaluate the combined effect of macro and pitch shortened threads on primary and biological stability.

3. To investigate histologically the bone formation properties around the proposed implant design for the static strains generated during implant placement and dynamic strains generated due to the movements of the rabbits.

4. To describe the early bone tissue response to full length micro threads designed between the reduce depth of macro threads, enhancing the initial bone to micro thread contact.
MATERIALS AND METHODS

In vitro

Finite element analysis (FEA) is an accepted theoretical technique mentioned in the literature, to evaluate implant design, its functions, and to optimize the design features. 92-99

In Study1, Initially, eight 3-dimensional (3D) dental implant models of same dimensions were designed using CATIA (Computer Aided Three-dimensional Interactive Application), a computer aided designing (CAD) modeling software, and files of these models were transferred in an initial graphics exchange specification (IGES) format to ANSYS 12.1 Software (Ansys INC, Houston PA, USA). Eight dental implant models were of different thread designs, out of which, the Power thread (model 1), Power acme thread (model 2), Triangular V-thread (model 3), Buttress thread (model 4), Reverse buttress (model 5) were the thread design of the implants, which were existing in the commercial market. In addition to these, three new thread design were selected; Crestal vertical slot power thread (model 6), Spiralock thread (model 7), and the Rounded power thread (model 8) (Fig 1) (All models mentioned were designed with an equal external diameter of 4mm and an equal length of 12 mm. With these models designed, it was planned to conduct study 1 in two stages.

Stage 1: 2D analysis – The aim of this finite element analysis was to evaluate stress generated through different thread designs, in the alveolar bone, when loaded vertically in two different bone to implant interface. The first situation of the bone to implant interface
was designed simulating immediate loading and the other situation was designed for implant loading after osseointegration.

**Geometry**

Two-dimensional finite element models were made from the 3D CATIA files of 8 different thread designs. The crest module and apex of the implants were removed from the implant models, in order to enhance the effect of macro thread designs only. 6mm (length) threaded section was isolated to enhance the effect of only the macro thread design. Thus, the supra-structures and microstructures of the implants were not considered. All the eight implant models were assumed to be embedded in cortical bone.

**FE mesh**

Finite element meshing of the first 8 models was specific of both the conditions of analysis. For the first simulation of immediate loading of the implant after placement (before integration), the bone to implant interface in the 2D models was made of frictionless contact elements. This would allow minor displacement between the implant and the bone. Under the conditions, the contact zone transfers pressure and tangential forces (i.e. friction), but not tension. The friction coefficient was set to 0.13. Similarly, in the other group of other eight more 2 dimensional models were made with bonded (i.e., fully osseointegrated implant) bone to implant interface, wherein there will be no movement of the elements between the implant and bone interface (Fig 2).
**Fig. 1.** Three-dimensional computer-aided designs of 8 different thread-designed implants (A, V-thread; B, power acme; C, power; D, buttress; E, reverse buttress; F, crestal vertical slot power; G, rounded; H, spiral lock).

**Fig 2.** Illustration of implant to bone contact (before and after integration).
Boundary conditions:
All the 2D Implant bone models were assumed to be fixed at the lateral and apical borders of the bone, as the width of the bone segment of the models (the distance between the fixed surface and the implant surface) was substantially larger than the maximum dimensions of the implant.

This would make these boundary conditions not affected by resulting bone stresses in the implant bone interface. All the models in both conditions, before integration (i.e. frictional interface) and after integration (i.e., fully bonded interface) were loaded vertically with 100N from the top. To simulate the clinical situation, material properties were given to the all the models (Table 1). Stress and strain distributions along the bone-implant interface were calculated in newtons.

Stage 2: 3D analysis: The aim of this finite analysis was to evaluate stress in the bone generated through various thread designed implants, only in bonded (i.e. after osseointegration) situation of the bone to implant contact.

Geometry – when the implant models designed included, the thread shape, crest module, body of the implant and the apex. Only 6 thread designs were selected from the original 3-dimensional IGES data of 8 models, which was used for stage 1 study. All the implant models in this study where constructed in an isotropic bone.

FE mesh –The finite element mesh was constructed using tetrahedral elements for the 3D modeling. Four-node linear cells were used instead of 10-node quadratic elements, since the later significantly complicate the computation of contact pressures. Bone-implant interfaces were designed as frictionless. All the models were statically loaded in vertical and oblique direction at 45° angulations. The amount of load applied in both the conditions was 100N. Stress and strain distributions along the bone-implant interface were calculated. In all the simulations the same material properties were assumed (Table 1).
Table 1. Mechanical Characteristics of studied materials.

<table>
<thead>
<tr>
<th></th>
<th>Youngs module MPa</th>
<th>Yield strength Mpa</th>
<th>Poisson’s Ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Titanium</td>
<td>96000</td>
<td>930</td>
<td>0.34</td>
</tr>
<tr>
<td>Cortical bone</td>
<td>12000</td>
<td>100</td>
<td>0.3</td>
</tr>
</tbody>
</table>

In study 4, a 2-Dimensional finite element analysis was planned with the aim to understand the pattern of stress distribution in bone, generated through the prototype implant design. A Pair of two-dimensional model was designed using CATIA, modeling software and transferred in an initial graphics exchange specification (IGES) format to ANSYS 12.1 (Ansys, NC, Honstan PA USA). The first model consists of a prototype dental implant, having macro threads at equal intervals and micro threads designed in between these macro threads. The other model consists of only macro threads at equal intervals with no micro threads in between them. Both the implant models were of 4mm in diameter and 12mm in length. The implants were assumed to be embedded in homogenous bone, with frictionless connection between the implant and bone. The macro threads and micro threads had V-shaped thread design. To enhance the efficiency of the threads, the crest module and apex was eliminated. The bone was assumed to be fixed at the lateral and apical borders, thus not affecting the resulting bone stresses in the implant bone interface. Material properties were given to the models and the models were statistically loaded vertically with 100N from the top of the models.

**In vivo**

**Prototype implant preparation**

In study 2 and 3, two sets of turned commercially pure titanium implants (4 mm in diameter, 8 mm in length, Grade 4) in total 40 each for both the studies, which included 20 each of test and control respectively, were prepared. The thread design of test group implants possessed V-shaped micro threads of 0.26 mm in thread pitch with 0.1±0.02mm of thread depth designed in between the macro threads,
which were also V-shaped threads. The control group of implants possessed only V-shaped macro threads at regular pitch intervals of 2.5mm, with 60° thread angulation and thread depth of 0.5mm. The control group implants did not possess micro threads (fig 3).

![Image of implants with V-shaped thread angles.](image)

**Fig 3.** The study implant with microthreads and control implant without microthread used in the study.

In study 4, based on the CAD model designs and the finite element analysis of these models, 48 implants which included two sets of oxidized surfaced prototype titanium implants (4 mm in diameter and 8 mm in length, grade 4 titanium) of 24 each were prepared, the basic implant design for control and test group implants was similar to that of implants used in study 2 and 3, but with alteration in the macro thread depth in relation to micro thread. The control implants possessed V-shaped macro threads at regular pitch intervals of 2.5 mm, with 60° thread angulation and thread depth of 0.4 mm, as compared with 0.5 mm in study 2 and 3. V-shaped micro threads of 0.26 mm in pitch-height with 0.1 to 0.2 mm of thread depth, similar to our previous study implants were added in between the macro threads for the test group implants (Fig 4).
Implant surface analysis
In study 2, 3 and 4, topographical analysis was performed for both test and control group of implants, using optical interferometry (MicroXAMt, PhaseShift, Tucson, AZ, USA). From each group, three implants were randomly selected and measured at nine areas each (three top areas, three thread valleys, three flank areas). The parametric calculation was performed after errors of form and waviness had been removed with a 50 μm² sized Gaussian filter and the following 3D parameters were selected: $S_a$ (μm) = the arithmetic average height deviation from a mean plane; $S_{ds}$ (μm²) = the density of summits; $S_{dr}$ (%) = the developed surface ratio; $Sci$ = core fluid retention index (Table 2,3) (Fig 4).

Table 2. Interferometer values of control and test implant surface of study 2 and 3 implants.

<table>
<thead>
<tr>
<th></th>
<th>Mean±SD $S_a$ (μm)</th>
<th>$S_{ds}$ (μm²)</th>
<th>$S_{dr}$ (%)</th>
<th>Sci</th>
</tr>
</thead>
<tbody>
<tr>
<td>Control</td>
<td>0.15±0.05</td>
<td>172360.69±49219.06</td>
<td>5.18±4.64</td>
<td>1.29±0.33</td>
</tr>
<tr>
<td>Test</td>
<td>0.23±0.13</td>
<td>168324.13±31618.06</td>
<td>6.26±5.18</td>
<td>1.15±0.29</td>
</tr>
</tbody>
</table>

Table 3. Interferometer values for control and test implant surface of study 4 implants.

<table>
<thead>
<tr>
<th></th>
<th>Mean±SD $S_a$ (μm)</th>
<th>$S_{ds}$ (μm²)</th>
<th>$S_{dr}$ (%)</th>
<th>Sci</th>
</tr>
</thead>
<tbody>
<tr>
<td>Test</td>
<td>0.25±0.05</td>
<td>189,422.31±52,397.04</td>
<td>7.59±6.28</td>
<td>1.42±0.32</td>
</tr>
<tr>
<td>Control</td>
<td>0.18±0.7</td>
<td>167,721.23±44,382.06</td>
<td>6.34±5.21</td>
<td>1.31±0.21</td>
</tr>
</tbody>
</table>
Surgical procedures  
*(Study 2, 3 and 4) and insertion torque measurements (Study 2)*

In Study 2 and 3, a total of 20 Swedish lop-eared healthy rabbits were used with an average weight of 3.7-5.2 kg. Before surgery, hind legs, the area of implant placement were shaved and disinfected with 70% ethanol and 70% chlorhexidine. The animals were anesthetized with intramuscular injections of a mixture of 0.15-mL/kg medetomidine (1 mg/mL Dormitor; Orion Pharma, Sollentuna, Sweden) and 0.35 mL/kg ketamine hydrochloride (50 mg/mL Ketalar; Pfizer AB, Sollentuna, Sweden). Lidocaine hydrochloride (Xylocaine; AstraZeneca AB, Södertälje, Sweden) was administered as the local anesthetic at each insertion site at a dose of 1 mL.

For the rabbits of study 2, 40 prototype implant of 20 each, test and control were alternately inserted in the left and right tibial and femoral metaphysis. After careful opening of the periostal flaps, implant sites were prepared with standardized drilling procedure. While placing the implants, the insertion torque (Ncm) was measured using a specially designed device (Elcomed, W&H, Milano, Italy). Fascia and skin flaps were closed in separate layers using single bioresorbable suture (Vicryl 4.0; Johnson & Johnson/Ethicon, Somerville, NJ). Postoperatively, buprenorphine hydrochloride (0.5 mL Temgesic; Reckitt Benckiser, Slough, UK) was administered as an analgesic for three days. After 4 weeks, these rabbits were sacrificed with an overdose (60 mg/mL) of pentobarbitalnatrium (Apoteksbolaget AB, Stockholm, Sweden).

In Study 4, 12 lop-eared rabbits were used. A total of 48 modified prototype implants of 24 each, test and control were placed contra laterally in left and right legs of the rabbit, one in the proximal tibia and one in the distal condyle of the femur in each hind leg, similar to our previous studies. The osteotomy sites after preparing for the adequate dimensions were pre-tapped at 35rpm, before inserting the implants. At the time of insertion, ITQ (Ncm) was measured in 9 rabbits out of 12, using the same special device (Elcomed, W&H, Milano, Italy). After 4 weeks, which was used in our previous studies all the rabbits were sacrificed with an overdose (60mg/mL) of Pentobarbitalnatrium (Apoteksbolaget AB, Stockholm, Sweden).
The 3 rabbits, which were not used to measure ITQ, were also sacrificed and the specimens were prepared for histomorphometric analysis.

The Malmö/Lund regional animal ethics committee approved all the three In-vivo studies.

**Removal Torque Analysis**

In study 2, all the implants of the 10 rabbits, and all the implants in 9 rabbits in the study 4, after 4 weeks of implant placement were subjected to RTQ test using an electronic device, with a special connector. A device used for many years as a standard technique and is described in our previous publications. The instrument was connected to the implants, and a fixed rotation rate of reverse torque was applied until failure of the bone-implant interface occurred. The peak values of resistance to reverse torque rotation were recorded Ncm. To calculate for the initial offset generated by the RTQ device, 3 Ncm was subtracted from each obtained value.

**Surgical procedures for histomorphometric analysis (Study 3)**

In study 3, Ten Swedish lop-eared rabbits and 3 rabbits in study 4 were used for histomorphometric analysis of the prototype and modified prototype implants respectively. The detailed surgical procedures from disinfection to anesthesia and placement of implant was similar carried out as explained earlier.

The control and test implants were placed contra laterally in left and right legs of the rabbit, both in the proximal tibia and the distal condyle of the femur. Postoperatively, each animal received oxytetracycline hydrochloride antibiotic (Terramicin, Pfizer pharmaceuticals, Finland; 1ml/3kg of bodyweight). The animals were euthanized after 4 weeks by an overdose of pentobarbital (Mebumal, ACO AB, Solna, Sweden).

**Histologic processing and histomorphometry**

In Study 3, All implants and surrounding bone tissue were retrieved *en bloc* from the sacrificed rabbits after 4 weeks, and the specimens were fixed by immersion in 4% buffered formaldehyde. The specimens
were dehydrated in a graded series of ethanol and embedded in light-curing resin (Technovit 7200VCL, Kulzer, Friedrichsdorf, Germany). Cut and grind sections were prepared from the embedded blocks, with a final thickness of 20 μm, and were stained with toluidine blue and 1% pyronin-G. Histomorphometric analysis of the sectioned specimens were performed with a Nikon Eclipse 80i microscope (Teknooptik AB, Huddinge, Sweden) equipped with an Easy image 2000 system (Teknooptik AB) using x1.8 to x100 objectives. The bone-to-implant contact (BIC) and the bone area (BA) around the macro thread and micro threads of the test and control implants were quantified.

Similarly, in Study 4, all the implants from the 3 rabbits planned for histomorphometric evaluation were retrieved en bloc, surrounded by bone and fixed by immersion in 4% buffered formaldehyde. Later the specimens were dehydrated and the sections were prepared to a thickness of 20mm with the similar technique as described above for study 3.

The qualitative analysis aimed at describing the early bone formation events at the macro and in particular the micro threads. In this study as well, the histometric evaluation comprised of measurements of the degree of bone-implant contacts and bone volume in a specified area, along the implant threads.

Statistics
For the study 2, the ITQ and RTQ values were analyzed by Wilcoxon signed-rank test in tibia and femur. The level of significance was set at 5% for all the analysis. Comparison of the histomorphometric values in study 3 and study 4 was also carried using Wilcoxon signed-ranked test.
RESULTS

**In vitro**

In study 1 and 4, stress and strain levels of the analysis were calculated using von Mises stress analysis.

2D Analysis-In study 1, the finite analysis of all the 8 models in bonded conditions (after osseointegration), at the implant bone interface, were found to have minor impact of the thread design on the stress levels in the bone. The stress levels in the bone were in the range of 5-13 MPa, with least stress in the reverse buttress thread design with 5 MPa and 13 MPa in v-thread design. Whereas, in frictional implant bone interface (immediately after implant placement) situation, the thread designs had a clear effect on the stress levels in the bone, with much variation in the stress levels amongst the different thread design. V-thread provided the lowest stress level at 14 MPa, Power acme, Buttress threads and Spiralock gave stresses in the range of 22-26 MPa. The crestal vertical slot gave maximum stress level at 80 MPa. Maximum stress was seen in Reverse buttress and rounded thread design with 107 MPa (Table 4) (fig 5, 6).
<table>
<thead>
<tr>
<th>Name</th>
<th>Model</th>
<th>Frictionless (MPa)</th>
<th>Bonded (MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Power thread</td>
<td>Model 1</td>
<td>33</td>
<td>10</td>
</tr>
<tr>
<td>Power acme thread</td>
<td>Model 2</td>
<td>22</td>
<td>8</td>
</tr>
<tr>
<td>Buttress threads</td>
<td>Model 3</td>
<td>24</td>
<td>12</td>
</tr>
<tr>
<td>Reverse buttress threads</td>
<td>Model 4</td>
<td>107</td>
<td>5</td>
</tr>
<tr>
<td>V-triangular threads</td>
<td>Model 5</td>
<td>14</td>
<td>13</td>
</tr>
<tr>
<td>Crestal vertical slot thread</td>
<td>Model 6</td>
<td>80</td>
<td>9</td>
</tr>
<tr>
<td>Rounded power slot threads</td>
<td>Model 7</td>
<td>107</td>
<td>7</td>
</tr>
<tr>
<td>Spiralock threads</td>
<td>Model 8</td>
<td>26</td>
<td>11</td>
</tr>
</tbody>
</table>

Fig 5. Two-dimensional bonded (osseointegrated) model analysis results in 8 different thread designs (A, V-thread; B, power acme; C, power; D, buttress; E, reverse buttress; F, crestal vertical slot power; G, rounded; H, spiral lock). (Study 1)
Fig 6. Two-dimensional frictionless model analysis results in 8 different thread designs (A, V-thread; B, power acme; C, power; D, buttress; E, reverse buttress; F, crestal vertical slot power; G, rounded; H, spiral lock). (Study 1)

3D Analysis – The study 1, 3D Analysis results showed the von Mises stress in the range of 4.8-30.9 MPa, when load of 100N was applied vertically. The Buttress threads and spiralock geometry provided the lowest peak stress of 8 MPa and 5 MPa respectively. The highest peak stress of 30MPa was found in Power Acme thread design. In the 45-degree oblique load application, spiral lock provided the lowest stress level of 13 MPa, and Power acme provided the highest stress level of 57 MPa. Power acme thread design provided the highest increase in peak stresses of 26 MPa between vertical and 45 degree. The buttress threads provided the smallest difference of 4
MPa between vertical and 45-degree load. (Table 5).

**Table 5.** 3D analysis results for vertical load and oblique load. (Study 1)

<table>
<thead>
<tr>
<th>Thread design</th>
<th>Vertical load (Mpa)</th>
<th>Oblique load, 45-degree (Mpa)</th>
<th>Difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Power thread</td>
<td>16.3</td>
<td>17.7</td>
<td>1.4</td>
</tr>
<tr>
<td>Power acme thread</td>
<td>30.9</td>
<td>57.13</td>
<td>26.23</td>
</tr>
<tr>
<td>V-triangular thread</td>
<td>12.6</td>
<td>30.13</td>
<td>17.9</td>
</tr>
<tr>
<td>Buttress thread</td>
<td>8.4</td>
<td>12.8</td>
<td>4.4</td>
</tr>
<tr>
<td>Crestal vertical slot</td>
<td>11.3</td>
<td>31.9</td>
<td>20.6</td>
</tr>
<tr>
<td>Spiral loc thread</td>
<td>4.8</td>
<td>12.8</td>
<td>8</td>
</tr>
</tbody>
</table>

In study 4, the finite element analysis of the prototype test and control implant design showed, that the stress levels were in the range was 0.28 Mpa to 62Mpa for the control implant model design, where as it was in the range of 0.28 Mpa to 31Mpa for the test implant model (Fig 7).

**Fig 7.** Von Mises stresses generated in test and control implants. (Study 4)
In vivo
Surface topographical analysis
In study 2 and 3, the test and control prototype implants were confirmed to have a smooth surface, i.e. <0.5 lm in average height deviation, which is comparable to abutment surfaces. As there was no statistically significant difference of height deviation, the density, the ability to retain fluid (the Sci parameter) or the enlarged surface area between the two prototype implant design (Table 2). The roughness values for the valleys in between the V-shaped threads of the test implants presented significantly higher values than other parts, however, they were still smooth. Thus, the influence of micrometer structures was negligible.

In study 4, similarly, the test and control implant samples were both found to have a smooth surface i.e < 0.5um in average height deviation, even though they were having the oxidized surface. There was no statistically significant difference between the two-implant design regarding their height deviation, the density, the ability to retain fluid (the Sci parameter) or the enlarged surface area (Table 2).

ITQ and RTQ results
In study 2, the mean values of the ITQ measured by the ITQ device for the control and test groups in the tibia were 15 and 20 Ncm respectively, and in the femur, the values were 11 and 12 Ncm, respectively. The mean RTQ values measured by the RTQ device for the control and test groups in the tibia were 7 and 9 Ncm, respectively, and in the femur, 7 and 7 Ncm, respectively. As most of the mean values were higher for the test group, the statistical analysis presented statistically significant differences between control and test implants for ITQ and RTQ in tibia and RTQ of femur, whereas there was no significant difference between control and test group for Femur ITQ (Table 6, 7, 8).
Table 6. Insertion torque values of the study implants in rabbits. (Study 2)

<table>
<thead>
<tr>
<th>Rabbit no.</th>
<th>Tibia (control)</th>
<th>Femur (control)</th>
<th>Tibia (Test)</th>
<th>Femur (Test)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>35.9</td>
<td>6.0</td>
<td>41.5</td>
<td>14.1</td>
</tr>
<tr>
<td>2</td>
<td>13.3</td>
<td>13.0</td>
<td>16.1</td>
<td>16.3</td>
</tr>
<tr>
<td>3</td>
<td>28.0</td>
<td>14.5</td>
<td>24.7</td>
<td>9.5</td>
</tr>
<tr>
<td>4</td>
<td>1.7</td>
<td>3.1</td>
<td>9.7</td>
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<tr>
<td>5</td>
<td>32.3</td>
<td>16.0</td>
<td>19.7</td>
<td>12.8</td>
</tr>
<tr>
<td>6</td>
<td>13.2</td>
<td>13.5</td>
<td>13.1</td>
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<td>7</td>
<td>1.9</td>
<td>23.5</td>
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<td>11.7</td>
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<tr>
<td>8</td>
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<td>9</td>
<td>1.5</td>
<td>12.8</td>
<td>27.0</td>
<td>18.5</td>
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<td>10</td>
<td>19.2</td>
<td>12.1</td>
<td>16.8</td>
<td>14.4</td>
</tr>
<tr>
<td>Mean</td>
<td>15.7</td>
<td>11.8</td>
<td>20.6</td>
<td>12.8</td>
</tr>
<tr>
<td>Median</td>
<td>13.25</td>
<td>12.9</td>
<td>18.3</td>
<td>13.45</td>
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<tr>
<td>Standard Deviation</td>
<td>12.88</td>
<td>6.2</td>
<td>8.9</td>
<td>5.1</td>
</tr>
</tbody>
</table>

Table 7. Removal Torque values of the study implants at 4 weeks of time. (Study 2)

<table>
<thead>
<tr>
<th>Rabbit</th>
<th>Tibia (Control) Ncm</th>
<th>Femur (control) Ncm</th>
<th>Tibia (test) Ncm</th>
<th>Femur (test) Ncm</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>4</td>
<td>4</td>
<td>7</td>
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<td>2</td>
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<td>5</td>
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<td>10</td>
<td>6</td>
<td>6</td>
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<td>10</td>
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<tr>
<td>Mean</td>
<td>7.9</td>
<td>7.9</td>
<td>9.1</td>
<td>7.7</td>
</tr>
<tr>
<td>STDV</td>
<td>4.3</td>
<td>4.3</td>
<td>4.4</td>
<td>3.1</td>
</tr>
</tbody>
</table>
Table 8. Wilcoxon signed rank test analysis results. [Study 2]

<table>
<thead>
<tr>
<th></th>
<th>ITQ</th>
<th>RTQ</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibia</td>
<td>3.57</td>
<td>2.47</td>
</tr>
<tr>
<td>Femur</td>
<td>1.79</td>
<td>2.47</td>
</tr>
</tbody>
</table>

In Study 4, the ITQ value measured by the ITQ device for the control and test groups in the tibia and femur demonstrated no statistical difference. The values were 11Ncm and 17Ncm respectively in the tibia, and in the femur 13 Ncm and 23 respectively. The RTQ values for the control and test groups in tibia was 11Ncm and 17Ncm, respectively and in the femur, 13Ncm and 23Ncm, respectively.

Histomorphometric analysis
In study 3, the histomorphometric analysis showed the mean total bone area, BA% (SD) were higher in the test implants, when compared to the control implants in both the tibial bone 24 (4), and 21(4), and in the femoral bone 29 (5), and 25 (7), respectively (Table 9). No statistical significant differences were confirmed for the BA% (p=0.13 for both compared groups, Table 9).

<table>
<thead>
<tr>
<th>Total BA%</th>
<th>Test implants Mean(median)±SD</th>
<th>Control implants Mean(median)± SD</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibia</td>
<td>24.61(23.27)±4.63</td>
<td>21.26(21.12)±4.39</td>
<td>0.131</td>
</tr>
<tr>
<td>Femur</td>
<td>29.55(29.55)±5.60</td>
<td>25.20(24.56)±7.52</td>
<td>0.131</td>
</tr>
</tbody>
</table>

Wilcoxon’s signed –rank test, with significance level of 0.05.

The amount of new BA% in the tibial bone of the rabbits was 29 (7) for the test implants, and 29(7) for the control implants, thus showing no statistical significant difference (p=0.86). On the other hand, in the femur, the new BA% in the femur was 32 (6) for the test implants, and 27(5) for the control implants, and the test implants had significantly higher new BA% (p=0.04) (Table 10). The mean BIC% (SD) was 11(4) and 13(6) for the test and control implants in the tibia and, 13(6) and 8(3) in the femur, respectively (Table 11). The statistical analysis showed no significant differences.
Table 10. New bone area (BA)±SD in the tibia and femur of the rabbit for the test and control group. (Study 3)

<table>
<thead>
<tr>
<th>New BA%</th>
<th>Test implants Mean(median)±SD</th>
<th>Control implants Mean(median)±SD</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibia</td>
<td>29.78(28.83)±7.12</td>
<td>29.67(26.44)±7.16</td>
<td>0.859</td>
</tr>
<tr>
<td>Femur</td>
<td>32.84(32.5)±6.04</td>
<td>27.31(28.19)±5.66</td>
<td>0.041*</td>
</tr>
</tbody>
</table>

* In comparison to significance level of 0.05.

Table 11. Bone-to-implant contact (BIC)% in the tibia and femur of the study rabbits for test and control group of implants after 4 weeks. (Study 3)

<table>
<thead>
<tr>
<th>BIC%</th>
<th>Test implants Mean(median)±SD</th>
<th>Control implants Mean(median)± SD</th>
<th>P-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibia</td>
<td>11.4(11.88)±4.21</td>
<td>13.91(12.06)±6.07</td>
<td>0.42</td>
</tr>
<tr>
<td>Femur</td>
<td>13.66(11.49)±6.86</td>
<td>8.48(7.92)±3.31</td>
<td>0.09</td>
</tr>
</tbody>
</table>

Table 12. Total bone volume around the test and control implant. (Study 4)

<table>
<thead>
<tr>
<th></th>
<th>Test</th>
<th>Control</th>
<th>Test</th>
<th>Control</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>59.57</td>
<td>56.13</td>
<td>59.57</td>
<td>39.81</td>
</tr>
<tr>
<td></td>
<td>98.05</td>
<td>30.6</td>
<td>70.6</td>
<td>42.45</td>
</tr>
<tr>
<td></td>
<td>61.19</td>
<td>57.63</td>
<td>59.88</td>
<td>34.41</td>
</tr>
<tr>
<td>Mean</td>
<td>72.93</td>
<td>48.13</td>
<td>63.35</td>
<td>38.89</td>
</tr>
<tr>
<td>SD</td>
<td>21.7</td>
<td>15.2</td>
<td>6.2</td>
<td>4.9</td>
</tr>
<tr>
<td>P-value</td>
<td>0.10</td>
<td>0.10</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 13. New bone volume around test and control implants after four weeks in study 4.

<table>
<thead>
<tr>
<th></th>
<th>Test</th>
<th>Control</th>
<th>Test</th>
<th>Control</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibia</td>
<td>19.32</td>
<td>22.74</td>
<td>19.32</td>
<td>1.36</td>
</tr>
<tr>
<td></td>
<td>39.7</td>
<td>13.38</td>
<td>33.68</td>
<td>10.78</td>
</tr>
<tr>
<td></td>
<td>36.01</td>
<td>49.02</td>
<td>18.15</td>
<td>14.11</td>
</tr>
<tr>
<td>Mean</td>
<td>47.51</td>
<td>33.63</td>
<td>43.53</td>
<td>23.82</td>
</tr>
<tr>
<td>SD</td>
<td>19.07</td>
<td>13.2</td>
<td>22.7</td>
<td>17.2</td>
</tr>
<tr>
<td>P-value</td>
<td>0.11</td>
<td>0.028*</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
The histologic observation presented deeply stained new bone formation around the implant for both test and control implants (fig 8,9). It was observed in the test threads that the newly formed bone seemed to follow the thread morphology, even in the marrow regions of the femur (Fig 10,11).

**Fig 8.** Descriptive light micrographs with the magnified section of the test implant in the tibia after 4 weeks of healing. The implant is protruding a dense cortical bone (CB) and into a bone marrow cavity consisting of a loose connective tissue (LCT) Formation of the new bone (NB). Original magnification 4X. (Study 3)

**Fig 9.** Descriptive light micrographs with the magnified section of the control implant in the tibia after 4 weeks of healing. The implant is protruding a dense cortical bone (CB) and into a bone marrow cavity consisting of a loose connective tissue (LCT) Formation of the new bone ("NB"). Original magnification 4X. (Study 3)
Fig 10. Descriptive light micrographs with the magnified section of the test implant in the femur after 4 weeks of healing. Dense cortical bone (CB) is surrounded by the new bone (NB). New bone is seen along the contour of the microthread implants. Original magnification 4X.

Fig 11. Descriptive light micrographs with the magnified section of the control implant in the femur after 4 weeks of healing. Loosened cortical bone (CT) is surrounded by new bone (NB). Original magnification 4X. (Study 3)
In study 4, the total bone area BA% was higher test implants with mean value of 72 when compare to 48 in the control group in the tibial bone. Similarly, in femur bone it was 63 for test and 38 for control group implants. Morphometrical analysis revealed increasing bone contact and bone area values for test implants in the tibia and femur (Table 12, 13). New bone area of test implants in femoral bone showed statistical significance (p<0.028), when compared with that of control group implants. Femur showed statistical significance in bone volume and bone to implant contact for test implants. This was statistically verified when pooling the data for the new bone.

Light microscopy observation revealed that bone formation from the periosteal and endosteal surfaces as well as at and around the dislocated bone particles. More primary bone contacts with the implant and bone formation was seen on the surface of the bone trabeculae for the samples in the cancellous bone. Solitary formation of woven bone was also seen in the bone marrow close to blood vessels, but at a distance from existing bone and the implant surface. Osteoblasts were observed towards the underlying layer at the implant surface. It was obvious that implants placed in cancellous bone had more bone at its interface due to bridging between bone trabeculae and the implant surface (Fig 12-13).

Fig 12. Descriptive light micrographs with the magnified section of the test and control implant in the tibia after 4 weeks of healing. Dense cortical old bone (OB) is surrounded by the new bone (NB). NB is seen along the contour of the micro thread implants. Original magnification 4°—. LCT = loose connective tissue. (Study 4)
Fig 13. Descriptive light micrographs with the magnified section of the test implant in the femur after 4 weeks of healing. Dense cortical old bone (OB) is surrounded by the new bone (NB). NB is seen along the contour of the micro thread implants. Original magnification 4X. (Study 4)
DISCUSSION

In vitro
Implants with extended and modified screws are highly recommended in implant dentistry today. Today, there are different types of externally threaded implants, which vary in thread pitch, shape and depth. The thread pitch is the number of threads per unit length. The thread shape defines the shape of the thread’s cross section such as square, V-shape, or buttress. The difference between the minor and major diameters of the thread is defined as thread depth. Since the morphology of screw threads plays an important role in the load transfer from dental implant to the surrounding bone, usage of different thread configurations for different bone qualities have been suggested. Screw threads positively influence the load transfer from the implant to the surrounding bone since an ordinary screw is able to transmit stresses in any direction. The threads have inclined faces and can carry vertical force in normal stress perpendicular to the interface. Use of cylindrical implant is not suggested anymore due to their concerns of achieving primary stability, thus leading to questionable osseointegration and higher failure rate. They also develop micro tension in peri-implant bones which leads to bone resorption. An important biological principal of bone is that it responds favorably to compressive loading (without the presence of a ligament) but not to shear forces. Therefore, screw thread implant designs have been adapted focus on reducing the surrounding shear forces by reducing the height of the thread profile (reducing the contribution of any one thread) with an increase in the number of threads per unit area of the implant surface. To better understand the stress dissipation in
various thread designs, study 1 of this thesis was planned. Study 1 was focused on the interfacial reactions of various thread designs’ response to the bone, 2 dimensional models in two situations. The First analysis was immediate loading of the implant model after placement (frictionless) and the second analysis was loading after osseointegration (bonded). Amongst the three new thread designs selected in this study, the crestal vertical slot power thread design was selected with the hypothesis, that it would dissipate the stress deeper in the bone rather than concentrating the stresses at the crest. Spiral lock thread design has earlier demonstrated that the load percentages on the first engaged thread using a spiral lock are significantly low when used in medical anchor.101

The size and thread profile cannot only be chosen on the basis of how it performs in an idealistic model but should also be validated when applied on the intended implant design and with the bone anatomy in which it is to perform. 106,107 In the 3D models, the complete implant with crest module and apex was constructed and embedded in the bone. Arramon et al, 108 mentioned in his study that, bone remodeling is controlled predominantly by the strains developed inside the bone. Hence, a dental implant should be designed with a thread design such that, when the surrounding bone is loaded, the thread design should reduce or avoid high stress peaks. The greater the applied load upon an implant, the higher is the bone stresses.107 Previous studies on stress dissipation have revealed that bone loss/resorption is initiated at the alveolar crest around the implant neck.109-111 In study 1, most of the thread designs, in the bonded (osseointegrated) situation of the 2D models, revealed maximum stress at the implant neck region of the bone to implant interface, when models were vertically loaded. The results of 2 dimensional models having frictionless contact (immediately after implant placement) showed that, different thread designs created different stress levels in the bone. The stress was not accumulated at the neck region of the bone to implant interface from any thread design. To maintain bone mass, the bone must be mechanically stimulated, such that remodeling of the bone occurs111 Stresses above physiological levels, will stimulate osteoclastic activities and it results in micro-damages and bone resorption.110 It is been mentioned that, if the implant is loaded with 1000 cycles per day in a bone with a Young
modulus of 12 Gpa, 12 Mpa is needed to maintain bone mass.\textsuperscript{111} But a stress level of 48Mpa may be over the physiological level.\textsuperscript{111} Thread designs in the immediately loaded (frictionless) 2D model showed disproportionate strain distribution in the bone, when compared with the thread designs in the osseointegrated (bonded) 2D model. In an osseointegrated bone implant interface, the load would be dissipated evenly in compressive and tension sites. However, for the immediately loaded implant, the compressive and frictional forces can be transferred into the bone areas only where the implant surfaces have contact. Therefore, force mainly passes through the contact sites, resulting in excessive strains, whereas strains are abnormally low at noncontact sites. These disproportionately low and high strains might result in a high risk for surrounding bone loss as a result of disuse atrophy or overloading resorption.\textsuperscript{112} In the Immediately loaded situation, crestal vertical slot thread design implant model did not distribute any load in the coronal part of the bone, thus indicating that no stimulation would be created in this region, which may further lead to decrease in bone mass, whereas, other thread designs provided some mechanical stimulation in the coronal part. The peak compressive stress arises outside the point where the lower flank of the thread passes into the curved top as seen in the reverse buttress design of the 2D frictionless model (Fig.6). It is reported in the literature that stresses (compressive) are more evenly distributed in the cases when the implant thread shape is in square shape rather than V-thread shape.\textsuperscript{46, 113} However, the current study findings of the 2D frictionless analysis showed that V-thread shaped thread design generated less stress, which is in supportive with conclusions of the study by Hansson and Werke, that the smaller the flank angle the longer should be the straight part at the bottom of the thread, both expressed in terms of a factor multiplied by the length of the curved part as well as in terms of a factor multiplied by the depth of the thread, and for most thread profiles, a straight part at the bottom of the thread seems to be unfavorable.\textsuperscript{107} Consequently, the bigger the bottom radius, the lesser is the flank support. Results reported by some literature based on complete osseointegration on the bone-implant interface are very similar to our data, that the stresses were concentrated at and around the implant neck.\textsuperscript{38, 114-119}
Similarly in study 4, the finite element models of prototype implants, when vertically loaded showed that, addition of micro threads along the length of the implant increased the capacity of the implants to take up axial load, by transforming shear stress into compressive stress at the interface.\textsuperscript{13, 49, 53,120}

\textit{In vivo}
The \textit{In vivo} studies were conducted to evaluate the primary, secondary stability and the volume of bone growth and bone to implant contact in the prototype implants. Study 2 was done to evaluate the primary and secondary stability of the test implants. The unique feature of the test implants was that the v-shaped threads were distributed not only in the upper portion of the implant, but spread over the entire implant in between the macro threads. It was anticipated that the V-shaped pitch shortened threads would be effective in force distribution during compression. Hence, the hypothesis was that the addition of these pitch shortened threads in between the macro threads would improve both primary and biological stability, whereas the implants with only macro threads would mainly have an impact on the primary stability. Furthermore, it was also hypothesized that the addition of pitch shortened threads would increase the available surface for bone incorporation. The observation time of 4 weeks was selected to investigate the effect of the implant design at early stages of new bone apposition. The ITQ and RTQ were measured to evaluate the effect of the two different implant designs. The results presented that the mean values were higher for the test implants for both ITQ and RTQ. However, there was no statistically significant difference between the two groups. Due to our initial aim to exclude the biological effect of other parameters, i.e. surface roughness or chemical modification, minimally rough turned surfaced implants were used in study 2 and 3. The RTQ results presented low values. There could be concerns that, the implant itself may not have been stabilized or osseointegrated. However, as the RTQ values presented in study 2 were calibrated to exclude the possible addition of the initial offset created by the device, the calibrated values represent the biomechanical interactions between the implant and the surrounding bone.
In study 2, it can be speculated that the pitch shortened v-shaped threads designed in test implants successfully relieved excessive stress concentrations, achieving adequate stress peaks that may have triggered bone formation. However, the upgrade in biological responses seemed to be moderate in the study 2, mainly due to the static load in the study model. It is possible that under dynamic load conditions, the idealized thread geometry might release excessive stress and persistently stimulate bone formation. This is of interest, as depending on the degree of strain, bone metabolism will metronomically swing to either osteogenesis or osteoclastogenesis. Molecular biological analyses indicate that such adequate mechanical stimulation accelerates osteogenic cell differentiation, accompanied by the upregulation of osteogenic markers such as collagen type-I or Cbfa-1. It has also been reported that micro-machined grooves influences the characteristics of bone deposition around implants. Grooved topography is thought to stimulate cell migration and induce guided contact osteogenesis, the extent of which is dependent upon the dimensions of the groove. Hall et al, proved that geometric grooves, which are clearly larger than osteoblasts, stimulate directed bone growth and deposition within the grooves, consequently optimizing implant stability. Hudieb et al, demonstrated in their finite element study that, the effectiveness of the threads with shortened pitches is not only in withstanding off-axis loading, but also in supporting the hypothesis of the effect of these threads in transforming shear stress into a compressive one, hence, its effects can be much estimated after loading the implants.

Study 3 of this thesis was focused on the bone response to these pitched shortened v-thread test implants in the femur and in the tibia of the rabbit. It was hypothesized that if the implants were granted primary stability by the pitch shortened/macro threads, the micro threads in between would stimulate bone formation. Furthermore, the expanded surface area generated by these micro threads provided enlarged surface area, which may have contributed in the increased bone incorporation as compared to the non-micro threaded implants. As the surface roughness was smoother than the original Branemark turned surface, which was claimed to be around Sa 0.7–0.8 mm, the biomechanical evaluation conducted in the study 2 resulted in low removal torque values, with no significant differences between the same test and control implants utilized in the study 3. It has been
suggested that removal torque testing may at times be too dynamic to capture the sensitive biologic interactions especially when the bone is undergoing a mineralization stage.\textsuperscript{128} Furthermore, the very smooth turned surface topography may not provide sufficient mechanical interlocking to withstand any torsional shearing stress. This was one of the reasons for conducting the study 3, to observe histologically at a relatively early stage of 4 weeks, the status of new bone formation around the test and control implants. The histologic analysis revealed new bone formation around the implant for both designs. However, of note was that new bone formed following the structure of the micro threads and seemed to be encapsulated by new bone (Figs 12,13). Although this did not show a statistically significant difference in the BIC, probably due to the smooth surface topography, new bone area in the femur presented statistically higher values for the test implants compared with the control implants. Furthermore, although not significant, there was a tendency that the mean BIC values in the femur for the test implants were higher than the control implants ($P = 0.09$). This is an indication that this implant with specific macrogeometry may be effective especially in bone qualities such as in the femur. It has been suggested that the rabbit femur simulates the human maxilla, where the bone quality may be poor.\textsuperscript{129} Thus, it can be suggested that the specific design may be beneficial when placed in compromised situations or in poor bone quality to enhance osseointegration.

Although no additional load was applied to the implant apart from the shear force applied during insertion, the dynamic force to the femur caused by the animal movement in the longitudinal direction may have triggered the alignment of hydroxyapatite (HA) crystals during bone remodeling. This load-dependent HA alignment has been reported by Nagisa \textit{et al}, that the biological apatite (B\textit{A}p) forms in correlation with variations in stress distribution.\textsuperscript{130} In their study, the B\textit{A}p in the bone formed differently depending on the force they bare. This may have been one of the reasons why bone formed along the micro threads, due to the optimal stress distribution around them, which stimulated the osteogenic cellular response. Thus, the effect of micro thread application is suggested to enhance bone formation from a geometric and shear force distribution perspective.
In study 4, a new prototype test implant was made, where in the basic design was similar to the prototype test implant used in study 2 and 3. But, in the new prototype test implants, the effect of micro thread stimulation for bone formation was enhanced with decreasing the macro thread depth, thus increasing the initial contact of micro threads with bone. This initial contact of micro thread will stimulate early bone formation at bone to implant surface. The stress levels were reduced in a prototype test implant finite model, when compared with the control implant model, when loaded vertically in the finite element analysis. Addition of micro threads increased the capacity of the implants to take up axial load, by transforming shear stress into compressive stress at the interface. Although the values of ITQ and RTQ between test and control were not statistically significant, but the mean values in the femur were higher, which may be due to the differences in bone support around the implant in two different types of bone. This has been clearly identified in the histomorphometrical measurements, seen in femoral bone with these study 4 test implants. Although in the samples, where the morphometrical measurements of the test implants placed in the femur showed approximately two times more of the supporting bone than in the tibia. One reason for this finding may be that the compression and stress distribution was more pronounced in the cancellous bone than in the cortical bone. Higher strains mean higher interface stresses and a higher frictional force, which increases the insertion torque for the test implants. Similarly, higher RTQ values were obtained in the test implants, when compared with control implant group, this probably reflecting the biomechanical properties of the trabecular bone at the femoral site (Table 7). Histomorphometric analysis values pooled showed, significantly higher amount of bone volume around the test implants in the femoral bone of the rabbit, when compared around the control implants. In study 4, to evaluate the statistics, Wilcoxon test was used as it has large power advantages over t-test, when the sample size is small. With the new prototype implants in study 4, it was found that bone formation occurred comparatively more near the micro threads than on a plane surface, thus indicating that micro threads promote bone stimulation. This was supported by the statistically significant values of bone to implant contact of the
study, where micro threads of the test implants were more often in contact in the bone after placement, thus allowing contact guidance, which may allow the cells to migrate in the direction, with the micro threads, as substrates. Thus bone formation would start on the surface and proceeded in the direction toward the surrounding tissues and along the implant surface, as mentioned by Davies and Hosseini and colleagues.\textsuperscript{135, 136} Thus, bone conduction leads to bone formation along the implant surface in parallel with and increased thickness of the bone by appositional growth. Thus, bone to implant contact was higher for the test implants in the femur bone of the rabbits, suggesting that micro thread configurations offered improved conditions for osseointegration.\textsuperscript{21, 137}
CONCLUSIONS

In vitro
1. The different designs of threads showed variation in the magnitude of the stress peak in the bone more effectively when immediately loaded.
2. Maximum stress accumulation was observed at the first thread in most of the osseointegrated implants, irrespective of the thread design.
3. The stress dissipation of more effective in prototype micro-thread implant models, when compared to the non micro thread design model.

In vivo
1. Due to the addition of pitch shortened threads in the test implant, it did significantly improve the primary or secondary stability of the test implants, when mechanically evaluated for ITQ and RTQ analysis in corticular or trabecular bone of the rabbits. Thus, with the proposed implant geometry, the effects of these threads could be confirmed immediately after placement of the implant.
2. Histomorphometric analysis also showed that the addition of the pitch shortened threads in between the macro threads did show more bone formation in the femur of the rabbits. Hence, it can be suggested that, this novel design may contribute to enhance osseointegration in poor bone qualities such as in the maxilla.
3. In the improvised test implants concluded that, the bone formation occurred more often near the micro threads than on plane surface, thus indicating that micro threads promote bone stimulation. Statistically significant bone growth in femur indicates that cancellous bone is more sensitive to micro thread stimulation.
FUTURE PERSPECTIVE

In future studies, both from a biomechanical and cell biological point of view, it is of great interest to observe the effects of the pitch shortened implants in clinical conditions, which may present different biological outcomes. It could be also of interest to understand the effect of immediate loading and immediate increase in load on these pitch-shortened thread design in dental implants.
POPULÄRVETENSKAPLIG SAMMANFATTNING


Syfte – Denna avhandling har som syfte att utvärdera olika gängprofiler i en matematisk modell för att finna en optimal gängdesign för kraftöverföring, omedelbart efter implantat insättningen såväl som när implantatet är färdiginläkt i benvävnaden. Ett ytterligare syfte är att utvärdera en experimentell implantatmodell med kortare gängor mellan de något större gängorna, hypotesen är att de små gängorna kan stimulera omgivande ben till nybildning via kompression medan de större ger nödvändig primär stabilitet för inläkning.

Material och metoder – I arbete 1 användes en 2D finit element analys (FEA) för att undersöka 8 olika gängdesigner. I denna modell var marginal och apikal del av implantaten avlägsnade så att enbart den gångade delen undersöcktes. Modellen var beräknad
för kortikalt ben. En 3D modell av FEA användes som komplement i syfte att utvärdera överförda spänning i benvävnad, i denna 3D modell undersöktes 6 olika gängdesigner och i denna beräkning var hela implantatet, dvs även marginal och apikal delen med. I In vivo arbetena 2 och 3 användes som test ett experimentellt svarvat implantat med en diameter av 4mm och 8mm längd. Implantatet var framställt med korta gängor mellan större gängor längs med hela implantatet, som kontroll användes samma typ av implantat men utan de korta gängorna. I studie 4 användes liknande implantat men de större gängorna var i denna studie mer avrundade och något avkortade. Implantatens primära stabilitet och benintegration utvärderades med den kraft som krävdes för att installera implantaten och urvrudningskraft efter inläkning. Histomorfometri användes också för att på ljusmikroskopisk nivå beskriva implantatets kontakt med benvävnad.

**Resultat** – I arbete 1 var 2D och 3D beräkningar gjorda med hjälp av von Mises spänningsanalys. Spänningsnivåerna i benet var i intervallet 5-13 MPa för den inläkta implantatmodellen och 14-107 MPa direkt efter implantat insättning. Kompletterande 3D analys visade på spännings i intervallet 4.8-30.9 MPa då implantateten belastades med 100N i vertikal riktning. I arbete 2 var medelvärdet för den kraft som krävdes för att skruva in kontrollimplantaten (ITQ) i tibia 15 Ncm och för test implantaten 20 Ncm. Motsvarande värden var för femur var 11 respektive 12 Ncm. I arbete 4 krävdes i tibia ett ITQ värde om 11Ncm för kontroll implantaten och 14 Ncm för test implantaten. I femur var ITQ för kontroller 13 Ncm och 19 Ncm för testimplantaten. Efter inläkning mättes den urvrudningskraft (RTQ) som krävdes för att avlägsna implantaten. I tiba var detta värde 11 Ncm för kontrollimplantaten och 17 Ncm för testimplantaten. I femur krävdes 13 Ncm för kontrollerna och 23 Ncm för test gruppen. Den histomorfometriska analysen i arbete 3 visade att medel benarean (BA % (SD)) i gängorna var högre för testimplantaten än för kontrollerna i både tibia och i femur, skillnaden var dock inte signifikant. I arbete 4 var medelbenarean i
tibia för testimplantaten 72% jämfört med 48% för kontrollerna. I femur var benarean i gängorna 63% för testimplantaten och 38 % för kontroll implantaten, p-värde 0.1. Mängd ben i kontakt med implantatytan visade signifikant högre värden för test implantaten i femur, p-0.04.

Slutsats – Betydelsen av gängdesign med avseende på överförd spänningsnivå till omgivande benvävnad är större vid implantatinstallation än när implantaten väl är inläkta i ben. Spänningsfördelning påverkades gynnsamt av kombination korta och större implantat gängor. Signifikant högre ITQ och RTQ observerades både i kortikalt och trabekulärt ben för implantat med denna kombination jämfört med implantat tillverkade endast med större gängor. Histomorfometrisk analys visade på en benstimulerade effekt av kombination korta och större implantatgängor i trabekulärt ben.
ACKNOWLEDGEMENTS

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ON EFFICACY OF IMPLANT THREAD DESIGN FOR BONE STIMULATION