A STUDY OF THE DESIGN PARAMETER VARIATION OF ORTHODONTIC MINI-SCREWS USING FINITE ELEMENT ANALYSIS

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ABSTRACT

The purposes of the present study are to determine the maximum torque resistance of mini-screws and to analyze design parameter variation associated with the maximum torque resistance of mini-screws on the biomechanical characteristics of the mini-screw and its surrounding bone using three dimensional finite element analysis. Six mini-screws (1.6 x 6 mm) from each manufacturer (Absoanchor®, O.S.A.S®, Dual Top®, Orlus® and Remark®) were placed in the artificial bone block and measured for the maximum insertion torque with a digital torque gauge. The lowest of the maximum insertion torque value among five manufacturers was selected for use in the finite element program in order to analyze design parameter variation of the mini-screws. The results in the artificial bone block experiment showed that only Absoanchor® mini-screws were fractured and the others survived with a torsional force of 20.07 N-cm and more than 38 N-cm (above digital torque gauge capacity) respectively. In finite element analysis, the stress concentration pattern in mini-screw models showed that the neck portion, from the thread runout to the second thread, was the weak point of mini-screw. In the surrounding bone , the stress concentration pattern was at the upper part of the cortical bone models, whereas the stress concentration of cancellous bone models was not obvious. The design parameter that had a correlation with maximum von Mises' equivalent stress was the thread diameter, r = -0.883.

KEY WORDS: MINI-SCREW/ MAXIMUM INSERTION TORQUE/ FINITE ELEMENT ANALYSIS

60 pages

การศึกษาการแปรฝันของตัวแปรเสริมในการออกแบบมินิสกรูทางทันตกรรมจัดฟันโดยใช้ระเบียบวิธีไฟไนต์เอลิเมนต์

A STUDY OF THE DESIGN PARAMETER VARIATION OF ORTHODONTIC MINI-SCREWS USING FINITE ELEMENT ANALYSIS

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บทคัดย่อ

วัตถุประสงค์ของการศึกษาครั้งนี้เพื่อประเมินการทนทานต่อแรงบิดสูงสุดของมินิสกรูและศึกษาการแปรผันของตัวแปรเสริมในการออกแบบที่สัมพันธ์กับการทนทานต่อแรงบิดสูงสุดของมินิสกรูบนชีวกลศาสตร์ของมินิสกรูและกระดูกที่อยู่ล้อมรอบโดยใช้ระเบียบวิธีไฟไนต์เอลิเมนต์ มินิสกรู (1.6×6 มม.) บริษัทละ 6 ตัว (Absoanchor®, O.S.A.S®, Dual Top®, Orlus® และ Ramark®) ถูกนำไปปักที่กระดูกเทียมและทำการวัดค่าแรงบิดสูงสุดในการขันมินิสกรูด้วยเครื่องวัดแรงบิดที่แสดงค่าเป็นตัวเลข ค่าต่ำสุดของค่าแรงบิดสูงสุดในการขันมินิสกรูทั้ง 5 บริษัทจะถูกเลือกมาใช้ในโปรแกรมไฟไนต์เอลิเมนต์เพื่อวิเคราะห์การแปรผันของตัวแปรเสริมในการออกแบบของมินิสกรู ผลการศึกษาในกระดูกเทียมพบว่ามีเพียงมินิสกรูของบริษัท Absoanchor ที่หักและมีการอยู่รอดของมินิสกรูในบริษัทอื่นๆด้วยค่าแรงบิดสูงสุด 20.07 และมากกว่า 38 นิวตันเซนติเมตร(เกินขีดความสามารถของเครื่องวัดแรงบิดที่แสดงค่าเป็นตัวเลข) ตามลำดับ ผลการศึกษาด้วยระเบียบวิธีไฟไนต์เอลิเมนต์พบว่ารูปแบบความเค้นในมินิสกรูอยู่บริเวณส่วนคอระหว่างรอยต่อของส่วนหัวกับส่วนเกลียวถึงเกลียวที่สองซึ่งเป็นจุดอ่อนแอของมินิสกรู ในกระดูกที่อยู่ล้อมรอบ พบรูปแบบความเค้นอยู่บริเวณส่วนบนของแบบจำลองกระดูกทึบ ในขณะที่แบบจำลองกระดูกโปร่งมีความเค้นไม่ชัดเจน ตัวแปรเสริมในการออกแบบที่สัมพันธ์กับค่าความเค้นสูงสุดคือเส้นผ่านศูนย์กลางส่วนเกลียวของมินิสกรู, r = -0.883

60 หน้า

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CHAPTER I
INTRODUCTION

Anchorage control is one of the most important factors in successful orthodontic treatment. There are many developed methods that obtain proper anchorage. According to Newton's Third Law of motion states "each action has an equal and opposite reaction". Therefore, it is virtually impossible to achieve absolute anchorage in which the reaction force producing no movement, especially with intra-oral anchorage. Traditionally, intra- or extra-oral devices can be an effective reinforcement [1], but demands exceptional patient cooperation [2], also be difficult to predict the treatment outcomes. While dental implant has been used extensively for tooth replacement, the size, bulkiness, cost and invasiveness of prosthetic implants are limited to the orthodontic application [3]. Therefore, orthodontists have developed various designs of implants that have been proposed to facilitate the anchorage control.

Titanium implants for orthodontic anchorage or orthodontic mini-screws in clinical are well documented. Their application has been increasingly in attention among the orthodontists. Since variety of applications such as anteroposterior tooth movement (Class I, II, III), vertical tooth movement (open-bite, deep-bite), transverse tooth movement (uprighting to correct crossbite) and other purposes (uprighting, artificial eruption), etc, are used, orthodontic mini-screws development has been very rapidly increased since 2003. There are many companies designing orthodontic mini-screws that serve orthodontic need, thus the designs, diameters and lengths of each company are different.

According to previous studies in various aspects of the mini-screws which were factors related failure in clinical uses [4-9], initial stability after the mini-screw placement [10], including the optimal insertion torque [11]. These clinical research were not enough for explain the interaction between the mini-screw and the surrounding bone. To understand the mini-screw better, it is imperative to do biomechanics analysis, and Three-Dimensional Finite Element Method (3D FEM) is by far the most reliable method [12]. According to previous studies of orthodontic loading on the mini-screws which resulted that changing geometry of the mini-screws had great influence on the biomechanical properties of both the mini-screw and its surrounding bone [13-17]. Furthermore, stress concentration on the mini-screw during its placement is one of the important considerations because it affected the mini-screw stability and the incidence of the mini-screw fractures were reported [18-20].

The aim of this study is to analyze design parameter variation associated with the maximum torque resistance of the mini-screws on the biomechanical characteristics of the mini-screw and its surrounding bone using 3D FEM.

CHAPTER II
OBJECTIVES

This study is designed to determine design parameter variation associated with the maximum torque resistance of the mini-screws on the biomechanical characteristics of the mini-screw and its surrounding bone using 3D FEM.

The purposes are:

1. To determine the maximum torque resistance of the mini-screws

2. To analyze design parameter variation associated with the maximum torque resistance of the mini-screws on the biomechanical characteristics of the mini-screw and its surrounding bone using 3D FEM

Hypothesis

1. There is no statistical significant of the maximum torque resistance of the mini-screws.

2. There is no correlation of the design parameter variation associated with the maximum torque resistance of the mini-screws on the biomechanical characteristics of the mini-screw and its surrounding bone using 3D FEM

Limitations of the study are:

1. The finite element analysis study may not correspond to the results found under intra-oral conditions. However, the results can predict and give information in the mini-screws selection for clinical use.

2. The result of the study may not be comparable to other studies due to the differences in the study design and material used.

Expected benefits from the study are:

1. This study will investigate various mini-screws available commercially in order to have a better understanding regarding to the maximum torque resistance for proper clinical selection in orthodontic treatment.

2. The result of this study can be applied as basic scientific knowledge for further study in orthodontic specialty.

CHAPTER III
LITERATURE REVIEW

Definition of Orthodontic Anchorage

The term *orthodontic anchorage* is defined as “resistance to unwanted tooth movement” [1]. According to Newton’s third law of motion, every action has an equal and opposite reaction. This is particularly relevant to orthodontics where such action is favorable tooth movement, and the equal as well as opposite reaction is often and unwanted tooth movement. An important aspect of treatment is maximizing desirable tooth movement, while minimizing undesirable side effects. Absolute anchorage is the term used to describe the anchorage unit that remains stationary under orthodontic force. Therefore, it is not possible to achieve absolute anchorage with intra-oral anchorage while extra-oral appliances require cooperation of the patient [21].

Terminology and Definition of Orthodontic Mini-screw

Presently, terminology of orthodontic mini-screws depend on the authors and its size. Mini-implant [22, 23], mini-screw [24-27], micro-implant [28, 29], micro-screw [30, 31] are referred to the group of implants that are 2.5 mm. or less in diameter [32]. In this study, “mini-screw” will be used referring to this group of implant. Mini-screws are small enough to place in any surface of alveolar bone , even in the interradicular regions. The placement and removal is simple and can be performed by orthodontists [22, 23, 28-30]. These mini-screws do not require osseointegration, only rely on mechanical retention between mini-screws and bone. They can be loaded immediately [24, 33] or wait for 2 weeks after insertion allow healing of the gingival [29, 30, 33].

Development of Orthodontic Mini-screw

Gainsforth and Higley [34] first reported using vitallium screws and stainless steel wires in dog ramus to attain orthodontic anchorage. Unfortunately, subsequent force application resulted in screws loss. In 1964, Linkow [35] reported the successful use of blade implants in a patient to apply Class II elastics for retraction of maxillary incisors.

Brånemark and co-workers [36] reported the successful osseointegration of titanium implants that were stable over five years in bone under light microscopic view. Roberts et al [37] corroborated osseous adaptation of rigid endosseous implants to continuous loading. Titanium implants were placed in rabbit femurs. After loading for four to eight weeks, titanium implants developed osseous contact and continuously loaded implant remained stable. The results indicated that titanium implants provided firm osseous anchorage for orthodontics and dentofacial orthopedics.

Block and Hoffman [38] introduced the onplant to provide orthodontic anchorage in dog and monkey experimental model. The onplant is a thin disc-like structure that is placed on the palate via a surgically created sub-periosteal tunnel. Following a period of osseointegration, futher surgery allows an attachment to made from the onplant to the teeth to provide indirect skeletal anchorage. However, this method has several disadvantage due to the high costs, long waiting period before loading forces, and the special abutment to connect the onplant.

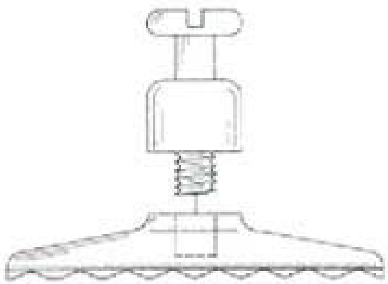


Figure 1 Onplant with internal thread for placement of transgingival abutment

Umemori and Sugawara et al [39, 40] developed the skeletal anchorage system (SAS) to correct the open-bite by intrusion of the lower molars. They applied surgical L-shaped titanium miniplates implanted in the buccal vestibule, and intrusive forces were generated in the molar area by an elastic thread that was tied between archwire and miniplates (Figure 2). Adequate molar intrusion was obtained after approximately six to nine months of treatment.

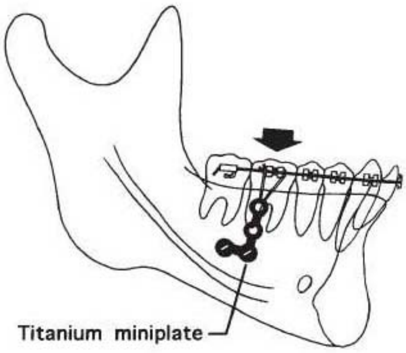


Figure 2 Scheme of the treatment mechanics for open-bite correction with SAS

In the case report by Chung and colleagues [41], they used miniplate with a hook soldered to one end that called the C- tube. It was designed instead of a rectangular slot to minimize torque in the archwire (Figure 3). The disadvantages of this system are the placement required a surgeon, swelling and pain after surgery.

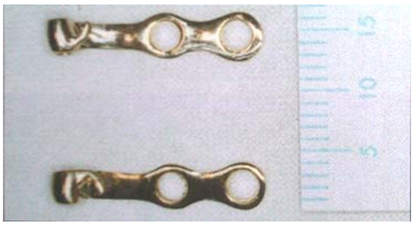


Figure 3 C-tube

To overcome several problems including high costs, large size, long waiting period before loading force and surgical need. Other designs of temporary device that fixed to bone for the purpose of enhancing orthodontic anchorage and subsequently removed after use had been introduced [2, 23, 24]. The most common type of temporary skeletal anchorage is a screw type that has small size, low costs, simplicity of use and wide range of clinical application [2, 23, 24, 29, 42].

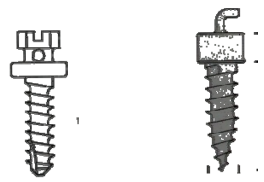
In 1997, Kamoni [23] reported a 1.2 mm mini-implant that modified from surgical mini-bone screw placed in the alveolar bone between the root apices of lower incisors for corrected the deep bite. Costa and colleagues [24] used titanium mini-screws, 2 mm in diameter and the length of 9 mm for orthodontic anchorage. The mini-screws were inserted manually with a screw driver directly though the mucosa without making a flap and were loaded immediately. Two years later, Melsen et al

[33] reported and experimental study in monkeys to evaluate the immediate loading of an implant that was a modification of the screw design in the study of Costa et al. Histological examination found osseointegration even the mini-screws were loaded immediately and the degree of osseointegration increase with time.

Ohmae et al [22] stated the clinical and histological results of titanium mini-implants used as anchors for intrusion in the beagle dog. They mentioned about a low amount of osseointegration, approximately 25%, does not necessarily indicate a negative finding because the mini-implant must be removed after the orthodontic treatment. If the mini-implant integrated bone completely, it would not be able to be removed easily.

Park and co-workers [29] conducted a study of micro-implants use (1.2 mm in diameter and 6 mm in length) as anchorage for treatment of skeletal class I, bialveolar protrusion case. The micro-implants were inserted into the buccal alveolar between the maxillary second premolar and first molar and the mandibular first and second molars. They showed that it could be inserted between the roots of teeth to retract six anterior teeth and intrude mandibular molar.

Lin et al [43] introduced the OMAS bone screw for anchorage reinforcement (Figure 4a). It was also developed in a new OMAS hook screw for coil springs attached in 2004 (Figure 4b).



(a) (b)

Figure 4 Bone screw (a) OMAS, (b) a new OMAS hook screw

Kyung et al [2] introduced absoanchor microimplants for orthodontic anchorage. Absoanchor microimplants was designed in different length, diameter and head (Figure 5).

Mini-screws are developed from 2003 very rapidly so there are various in designs, diameters, lengths and composition of materials that have been manufactured to serve orthodontic needs (Figure 6).

Figure 5 Absoanchor microimplants

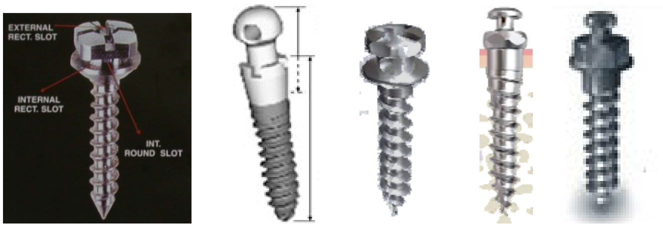


Figure 6 Orthodontic mini-screws

Material of Orthodontic Mini-screw

There are four groups of material reported for the use of implants; bioinert, biotolerant, bioactive and bioresorbable material [44, 45]. The detail of each group will be presented as followed.

1. Bioinert material can be classified into two groups. Commercial pure titanium consists of 99.5 percent titanium and the remaining 0.5 percent is other elements such as carbon, oxygen, nitrogen, and hydrogen. It is excellent biocompatibility and biomechanical characteristics. Another group of bioinert material is titanium alloy which is more common for the mini-screw. This titanium alloy composed of 90 percent titanium and the remaining elements are vanadium and aluminum. Trace elements include carbon, oxygen, nitrogen and hydrogen [46].

2. Biotolerant material consists of stainless steel and cobalt-chromium alloy. A typical 316L stainless steel composition would be 18% chromium, 12% nickel, 2% molybdenum, and 0.03% carbon. Although this alloy is stronger, cheaper, and easier to machine, its corrosion properties are inferior to titanium. For this reason, it has not been approved as a dental implant material [47], [48]. Cobalt-based alloys have been used for decades to make cast partial denture frameworks. Typically, these alloys contain 62% cobalt, 31% chromium, 5% molybdenum, and trace amounts of iron,

magnesium, silicone, and carbon. These alloys cast well with sufficient strength to withstand the occlusal forces applied to partial denture frameworks. Although it is not resistant to corrosion as titanium, cobalt alloys exhibit reasonable biodegradation properties when exposed to human tissues. This corrosion resistance arises from an oxide layer, Cr_2O_3 formed on alloy surface. These alloys are frequently used to fabricate hip prostheses. However, the use of this material group for implant is not very popular.

3. Bioactive material, veteroceramic apatite hydroxide and ceramic oxide material. The bioactive surface of this material enhances the integration between implant and tissue.

4. Bioresorbable material group. The composition is the combination of polylactic and polyglycolic acids. The excellent property of this bioresorbable material is degraded after use [49]. This material is now commercially available in form of mini-screw. However, the cost is still not very economical.

According to Tansalarak's study [50]confirmed that the majority material of the mini-screws were made from Ti-6Al-4V follow by the international standard (ISO) 5832-3:1996 [51], the requirements for the major and minor elemental constituents for Ti-6Al-4V alloy are listed in Table 1.

Table 1 Chemical composition

Element	Compositional limits % (m/m)
Aluminum	5.5 to 6.15
Vanadium	3.5 to 4.5
Iron	0.3 max.
Oxygen	0.2 max.
Carbon	0.08 max.
Nitrogen	0.05 max
Hydrogen	0.015 max.*
Titanium	Balance
* Except for billets, for which the maximum hydrogen content shall be 0.010% (m/m)	

Basic Components of Orthodontic Mini-screw

A screw is a simple machine that converts rotational motion into translational motion while providing a mechanical advantage [52, 53]. Generally, a screw has three basic components: a core, a helix (called the thread), and a head (Figure 7). Each component plays an important role in the function of the screw.

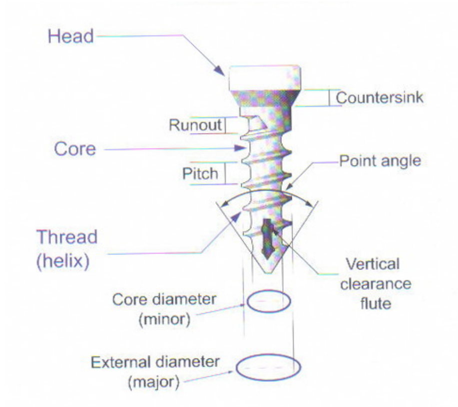


Figure 7 Basic components of a screw [54]

The head of an orthodontic mini-screw basically serves two purposes, first is providing a means for applying twisting torque to the core and thread and the other is acting as an application point for force. Various means of engaging a screwdriver, including a slot, cross-slot, and recessed hex, are available for bone screws [55]. Bone screws are generally used for closed implants, so they require a less prominent head shape. Therefore, a female-type means of engaging a screw-driver is preferable, and the recessed hex has proved to be the most useful for bone screws [52, 55]. On the other hand, orthodontic mini-screws are generally used for open implants, so a male-type means of engaging a screwdriver may be favorable, because it provides the best articulation of a screwdriver and may offer better control during insertion.

The core, which forms the support of the screw, is attached to the head and is wrapped in the helical thread [52, 53]. The cross-sectonal area of the core determines the torsional strength of the screw [52, 53, 56]. Because the torsional strength is proportional to the cube of the core diameter [56], a extremely small enhancement of core diameter can greatly increase the strength of a screw. The greater of core diameter has the lower incidence of screw failure from fracture during insertion of screw. The shank is the part of the screw that extends from the head to beginning of the threads. The spacing between adjacent threads is called the pitch. The led of a screw refers to the distance that the screw will advance with each turn [52, 53]. In a screw with a single thread, the pitch will equal the lead [52, 53]. The cross-sectional shape of the thread is important as well because it is related to insertion methods and stress distribution [52, 53, 57]. The thread shape can be divided in to 4 types V-shape, buttress, reverse buttress and square [58] (Figure 8). Under axial loads to an implant-bone interface, a buttress or square- shape thread would transmit compressive forces to bone. The V-thread design is called a “fixture” in conventional engineering applications and is primarily used for the fixation of the metal parts together because the 30 degree incline of the V-thread design cause the male component of the screw to stretch during preload, which decreases the incidence of screw loosening. The original Branemark implant system had a V-thread pattern in order to place in a threaded osteotomy [59]. The reverse buttress thread shape is flat on the top and is optimized for pullout loads [60].

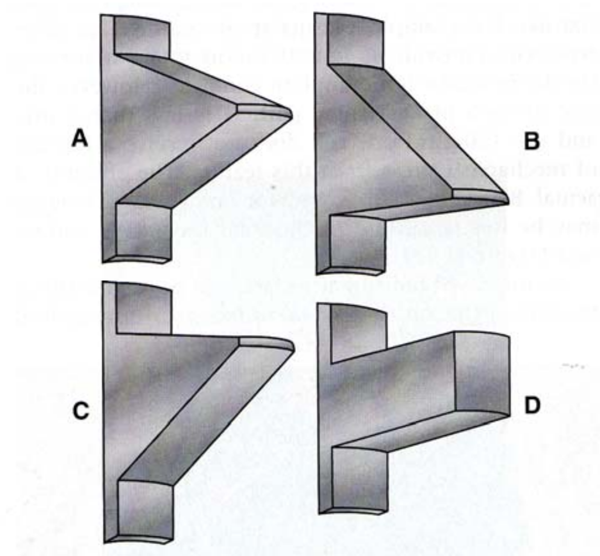


Figure 8 The four basic thread shapes: (A) V-thread, (B) buttress thread, (C) reverse buttress thread, and (D) square thread [58]

The thread diameter is the largest diameter of the thread portion of the screw measured over the thread crests. This is also known as the major or external diameter.

The core diameter is the smallest diameter of the thread portion of the screw measured at the thread root. This is also known as the minor diameter or root diameter.

Screws are classified as pretapped screws, self-tapping screws, or self-drilling screws, according to the method of insertion [52, 57] (Figure 9). The insertion method is also related to the physical properties of materials. Pretapped screws are used in harder, less compressible materials such as in metal or in cortical bone [52, 53, 57]. Because the screw threads cannot readily compress these firm materials, pretapped screws require the use of a tap to precut the thread. Pretapped screws are not suitable for thin bone, such as the maxilla [57].

Self-tapping screws are used in softer, less compressible materials and form threads by compressing and cutting the surrounding materials. They have a fluted leading edge and require only a predrilling procedure, meaning that the tapping procedure is omitted [57]. Self-drilling screws, also referred to as drill-free screws, have a corkscrew-like tip. Therefore, neither predrilling nor tapping procedures are needed.

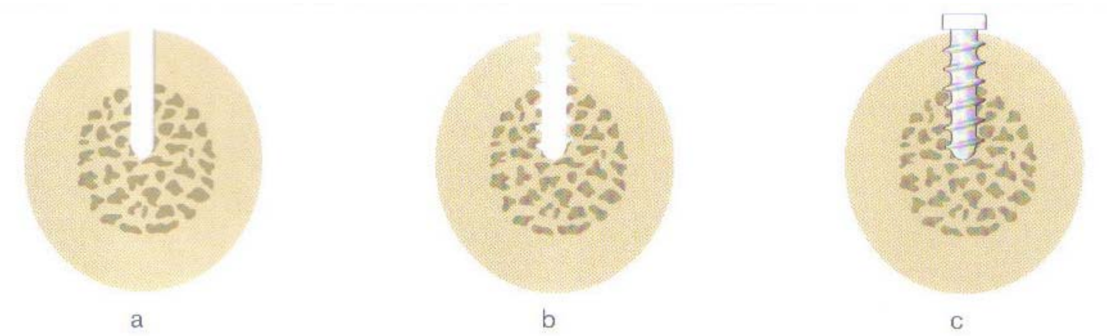


Figure 9 Pretapped screws are inserted after (a) drilling and (b) tapping procedures. The drilling procedure is omitted when self-tapping screws are used, while all drilling and tapping procedures are omitted with used of self-drilling screws, which involve only (c) an insertion procedure

Failure of Orthodontic Mini-screw

Failure of the orthodontic mini-screw can be categorized into hard tissue-mini-screw interface failure, soft tissue-mini-screw interface failure, psychological failure, and mini-screw failure from fracture [54].

Hard tissue-mini-screw interface failure

Failure at the hard tissue-mini-screw interface results in loosening of the mini-screw [8, 9, 11, 61]. According to studies of the success rate of mini-screws, most failures result from the loosening of the mini-screw shortly after implantation [8, 20]. Early failure at the hard tissue-mini-screw interface is related to primary stability [62], which is obtained from mechanical support from the surrounding bone tissue. In other words, primary stability is related to the thickness of the cortical bone at the implantation site [63], the amount of damage caused by surgical trauma, and the closeness of the contact between the bone and the mini-screw. Later failure at the hard tissue-mini-screw interface is related to the type of interface formed through the healing process following implantation [8]. Long-term failure is also associated with type of stress loaded on the mini-screw. Formation of fibrous tissue at the bone-mini-screw interface is regarded as the most important risk factor in the loosening of screws [64], and shear stress is more detrimental to the bone-mini-screw interface than compressive or tensile stress [65]. Primary stability, biocompatibility of mini-screw, and the trauma resulting from implantation all contribute to type of interface that is formed [65]. Primary stability, which is the mechanical stability present immediately following implantation, has significant effects on both short-term and long-term stability [65, 66].

Soft tissue-mini-screw interface failure

Plaque accumulation around the mini-screw or persistent mechanical irritation can cause soft tissue interface problems, such as acute or chronic inflammation or infection. Epithelial hyperplasia or epithelial covering may also occur. In severe case, infection can progress to abscesses. The potential for this kind of problem to develop is significantly increased when the mini-screw is placed on movable tissue [67].

Some investigators have suggested that chronic inflammation around mini-screws is a risk factor for loosening [9, 61], but these reports have been disrupted. Inflammation around the mini-screw could also be a consequence of loosening.

The mini-screw should be removed immediately from patients with infection plus any general symptoms such as fever or abscess, sustained discomfort, and affected adjacent periodontal attachments.

Phychological failure

Psychologically, mini-screw placement is not always accepted by patients or the parents of patients. A cost benefit analysis of mini-screw placement should be thoroughly explained at the consultation [68]. For example, mini-screw placement is one of several treatment options to relieve crowding. However, to achieve nonsurgical correction of a long face, placement of an implant is the only option.

Mini-screw failure from fracture

Mini-screw fracture may occur during surgical placement or removal [65, 66, 69] but will occur during orthodontic force application. An intrinsic limiting factor regarding mini-screw fracture, the torsional strength of mini-screw depends on the physical properties of the material and is proportional to the cube of diameter [56]. Increased torsional stress during placement can lead to mini-screw bending or fracture, or produce small cracks in the peri-implant bone, that affect mini-screw stability [70-72].

The best way to prevent fracture is to increase the diameter and to use stronger materials such as chrome-cobalt alloy [73]. However, both of these changes are impractical since a large implant cannot be placed interproximally and stronger material has inferior biocompatibility. The fracture site depends on the cause of fracture. The fracture of implants can be prevented by elimination of the possible causes of fracture. The design of the apical tip was altered to increase the mechanical strength of the tip, and a lateral cutting groove was added to prevent stress concentration. Because the torsional strength is proportional to the cute of the core diameter, a very small enhancement of core diameter can greatly increase the strength of a screw. The mini type diameter should not be used where cortical bone is comparatively thick.

To prevent the fractures, self-drilling mini-screws should be inserted slowly, with minimal pressure, to assure maximum mini-screw-bone contact. A purchase point or a predrilling is recommended in regions of dense cortical bone, even for self-drilling mini-screws [18]. A short mini-screw is recommended for these areas for prevention of fracture. Modifications of the design, proper manipulation, and use of the pre-drilling procedure can minimize mini-screw fracture.

Mini-screw fracture during removal, the mini-screw head could fracture from the neck of the shaft during removal. The recommended minimum diameter is 1.6 mm for self-drilling mini-screws that are 8 mm or longer placed in dense cortical bone [18]. The proper placement technique can minimize the risk of mini-screw fracture during its removal. If the mini-screw fractures flush with the bone, the shaft might need to be removed with a trephine.

Factor Affecting the Maximum Insertion Torque of Orthodontic Mini-screw

Host factors

The condition of the hard tissue depends on the age and sex of patient and on the location of the mini-screw placement site; the quantity and quality of host bone bed at the implantation site also greatly influence stress distribution on mini-screw.

Bone Quality is depended on bone density and bone quantity. Lekholm and Zarb [74] classified bone density radiographically into four types based on the amount of cortical versus trabecular bone in given area of the alveolar bone (Figure 10):

D1: Homogeneous cortical bone

D2: Thick cortical bone with dense trabecular core

D3: Thinner cortical bone with dense trabecular core

D4: Thin cortical bone with low density trabecular core

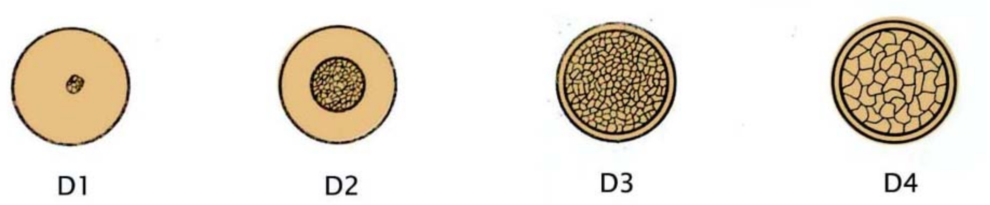


Figure 10 Lekholm and Zarb bone density classification [74]

Misch proposed an extension of this idea, by providing comparative material of differing resistance to drilling to aid classification as subjective classification. Drilling and placing implants into D1 bone is similar to drilling into oak or maple-like. D2 bone is similar to the tactile sensation of drilling into spruce or white pine wood. D3 bone is similar to drilling into balsa wood. D4 bone is similar to drilling into styrofoam [75].

A review of literature by Misch [75] found the location of different bone densities may be superimposed with the different regions of the mouth (Table 2) D1 bone is almost never observed in maxilla. In the mandible, D1 bone is observed twice as often in the anterior region compared with the posterior region. The bone density D2 is the most common bone density observed in the mandible. The maxilla presents D2 bone less often than the mandible. Bone density D3 is very common in the maxilla. Almost half of the posterior mandible also presents with D3 bone. The softest bone, D4, is most often found in the posterior maxillae (approximately 40%), especially in the molar regions. The mandible presents with D4 bone less than 3% of the patients. In conclusion, the anterior maxilla is usually D3 bone, the posterior maxilla is D4 bone, the anterior mandible is D2 bone, and the posterior mandible is D3 bone.

Table 2 Usual anatomic location of bone density type (% occurrence)

Bone	Anterior Maxilla	Posterior Maxilla	Anterior Mandible	Posterior Mandible
D1	0	0	6	3
D2	25	10	66	50
D3	65	50	25	46
D4	10	40	3	1

Friberg et al [76] applied the cutting resistance measurements to access bone quality during implantation confirmed this bone density locations distribution. The measurement of cutting resistance values during low-speed tapping in autopsy specimens were higher in mandible compared with maxilla, and there was tendency towards greater value in the incisors regions than in the premolar regions in both jaws because posterior regions tends to have a thinner, more porous cortex and fine trabeculae [77].

Norton and Gamble [78] demonstrated that an objective scale of bone density based on the Houndsfiled unit is strongly correlated with subjective quality score and also correlated with the region of the mouth (Table 3). Houndsfiled scale is the X-ray attenuation unit that is mostly used in computed tomographic scanning and characterizes the relative density of a substance. Each pixel is assigned a value between -1 and 1k. The value of zero equals water, and soft tissue such as muscle tissue equals +40, air (-1000) and bone (+50 to +2500) [79].

Table 3 Correlation between subjective quality classification, objective scale of bone density, and region of the mouth

Bone Quality (Lekholm and Zarb)	**Bone density range (HU) (Norton and Gamble)**	**Region of interest**
Q1	> +850	Anterior mandible
Q2/3	+500 to +850	Posterior mandible Anterior maxilla
Q4	0 to +500	Posterior maxilla

Bone quantity refers to bone thickness. Cortical bone in the mandible is thicker surrounding the implant than in the maxilla. Deguchi et al [80] quantitatively evaluated cortical bone thickness in various locations in the maxilla and mandible with computed tomographic scanning for orthodontic implants. They found that less cortical bone thickness was observed at buccal region distal to the second molar compared with other areas in the maxilla and more cortical bone was observed on the lingual side of the second molar compared with the buccal side. In the mandible, mesial and distal to the second molar, significantly more cortical bone was observed compared with the maxilla. Cortical bone thickness resulted in approximately 1.5 times as much at 30^{o} compared with 90^{o}.

Ono et al [81] investigated cortical bone thickness in the buccal posterior region mesial and distal to the first molar, where mini-implants are often placed by using computed tomography. They found that cortical bone thickness was measured from 1 to 15 mm below the alveolar crest at 1 mm intervals. The average cortical bone thicknesses ranged from 1.09 to 2.12 mm in the maxilla and 1.59 to 3.03 mm in the mandible. The greater the height, the thicker the cortical bone tended to be, and the mandibular cortical bone was significantly thicker than that of the maxilla. The cortical bone was thinner in females than in males in the region of attached gingiva in the maxilla mesial to the first molar.

Bone density measured by computerized tomography (CT) was related to maximum insertion torque value [82, 83]. Mini-screw inserted in the high bone density area has significant higher maximum insertion torque than mini-screw placed in the low density bone. Cortical bone thickness also influences the maximum insertion torque. As the thickness of cortical bone increases the maximum insertion torque increases [26, 84].

Operator factors

The torsional stress to mini-screw and surrounding bone is also dependent on direction, redirection and over force during mini-screw placement, as well as including the dexterity of the operator.

Mini-screw factors

Materials consist commercial pure titanium, titanium alloy and stainless steel have been reported to use in mini-screw construction. Widely used material is titanium alloy (Ti-6Al-4V) because of its good biocompatibility and biomechanical characteristics, with a much higher strength than commercial titanium and a better corrosion resistance than stainless steel. Malaith et al [85] suggested that implant materials have a modulus of elasticity at least 110,000 N/mm^2.

According to previous studies mini-screw geometry [26, 84, 86] was the one of the important factor that affect the insertion torque value. You et al [86] found that external diameter, unthreaded shank height, head slot and self-tapping cutting flute design had the greatest impact on screw strength the intermediate pitch value yielded the highest maximum insertion torque values, which is about 0.8-0.9 mm for 2 mm diameter screw. Thread pitch length more or less than this value decreased the maximum insertion torque. For 2 mm screw, the maximum insertion torque value decreases as the unthreaded shank increased. However, the screws with 1.5 mm or smaller diameter, the variations in the unthreaded shank and pitch lengths did not seem to greatly affect the insertion torque. Thread depth and core diameter did not affect the insertion torque significantly. Song et al [26] determined torque depending on the mini-screw design in relation to artificial cortical bone. Tapered shape mini-screw had significantly more insertion torque value than the cylindrical mini-screw. The part of the mini-screw that makes contact with cortical bone affects insertion torque value the most. The contact area with the cortical bone of the cylindrical shaped mini-screw is not as wide as the tapered form screw, therefore the insertion torque value in the tapered screw is greater than the cylindrical group. Among the tapered form screw, mini-screws with more degree of taper has higher insertion torque value. Lim et al [84] found that the external diameter of the mini-screw is the most influencing in determining the insertion torque, it could predict 80% of maximum insertion torque. Mini-screws with larger external diameter have higher insertion torque. Longer mini-screws required more torque to tighten them, especially the cylindrical screw. During insertion of the screw, the insertion path in cortical bone is formed so that when the diameter in the cortical bone reaches the maximum diameter of the screw, the increase in torque is mainly affected by the insertion depth in the medullary bone [26].

In clinical study, Motoyoshi et al [11] studied the insertion torque used for tightening 124 mini-screws in 41 orthodontic patients. The diameters of mini-screws were 1.6 mm and 8 mm long. The mini-screws were all placed in the self-tapping method, into the buccal alveolar bone of the posterior region. The peak insertion torque value was recorded at the terminal turning when the taper-shaped screw was tightened into the bone by a torque screw driver. The 6-month success rate was 85.5%. The mean insertion torque ranged from 7.2 to 13.5 N-cm, depended on the location of the mini-screws. The success rate for mini-screws with an insertion torque value ranged from 5 to 10 N-cm was significantly higher than that for implants with insertion torque 5 N-cm or less, and more than 10 N-cm.

In Srinok’s study [87], the insertion torque was distributed in a range of 0.5-16 N-cm. The measurements were investigated in 25 pateints with 42 mini-screws (1.4 mm in diameter and 7 mm in length). The mean of the optimal insertion torque was 4.62-7.08 N-cm.

Anka [42] reported the ideal torque value at the end of implantation is between 15 N-cm and 20 N-cm. The torque value should not less than 10 N-cm otherwise the mini-screw will not be able to resist immediate loading. Sites with dense bone, where the final torque value can be above 20 N-cm, should be pre-drilled to avoid fracturing the mini-screw.

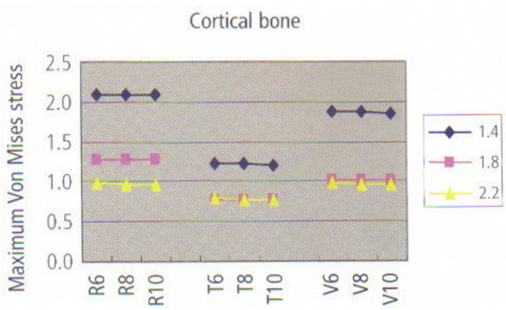
According to the procedure factor, there are two methods for mini-screw insertion, self drilling and self tapping methods. The self tapping method, mini-screws were inserted into bone after drilling a pilot hole, has disadvantages such as thermal necrosis of the bone [88], more instruments and time required. The pilot hole size also affects the holding power of mini-screw, which should not exceeds 85% of external diameter of the screw otherwise the holding power will decrease rapidly [89]. Self drilling screw, which enables the mini-screw to insert without drilling, was developed to avoid these problems [90]. The self-drilling method has higher insertion torque, more bone-screw contact [71, 90].

Table 4 Summary of the past studies related with the insertion torque of mini-screws

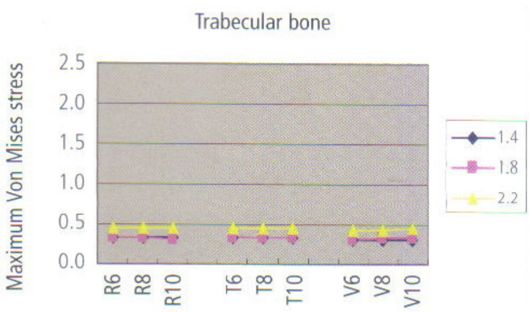
Researcher	**Type of study**	**Type of screw**	**Diameter (mm) × Length (mm)**	**Insertion torque (N-cm)**
Song et al (2007) [26]	Laboratory	Cylinder Taper Taper	1.5×6 1.6×6 1.6×7	20.33 37.89 36.20
Lim et al (2008) [84]	Laboratory	Cylinder Cylinder Cylinder Cylinder Cylinder Cylinder Cylinder Taper Taper Taper	1.2×8 1.5×7 1.5×8 1.5×9 1.8×8 2.0×8 2.5×8 1.5×6 1.5×7 1.5×8	16.5 19.5 20.9 23.0 31.3 51.1 80.6 32.6 35.6 37.3
Motoyoshi et al (2006) [11]	Clinical	Taper	1.6×8	7.2-13.5
Srinok (2008) [87]	Clinical	Cylinder	1.4×7	4.62-7.08
Anka (2006) [42]	Review article		-	15-20

Previous Finite Element Studies Related With the Mini-screw Geometry

A nonlinear finite element model analysis using two-dimensional models, which reflected the condition of the bone-implant interface immediately after implantation, was used to investigate which screw parameters affect early stability [17, 54]. Lee [17] found the length of the mini-screw that extend 4 mm and 6 mm which extend in bone were shown to have little effect on the distribution of stress, but the thread design and the diameter had a significant effect on the distribution (Figure 11). The diameter mediates a significant effect on the stress distribution of the bone [17] (Figure 11 to 12). In cortical bone, the thicker the diameter, the more favorable the stress distribution [17, 91-93]. The thread design, or cross-sectional shape of the thread, is related to both the stress distribution under loading and the implantation method.(Figure 13). The reverse buttress thread provides the easiest insertion but is least advantageous in terms of stress distribution [60]. A trapezoidal or rectangular shape results in more difficult insertion but provides the most advantageous distribution of stress.



(a)



(b)

Figure 11 Maximum von Mises stress in (a) cortical bone and (b) trabecular bone acoording to implant length (6, 8, 10 mm), diameter (1.4, 1.8, or 2.2 mm) and thread design (reverse buttress (R), trapezoidal (T), or V-shaped (V)

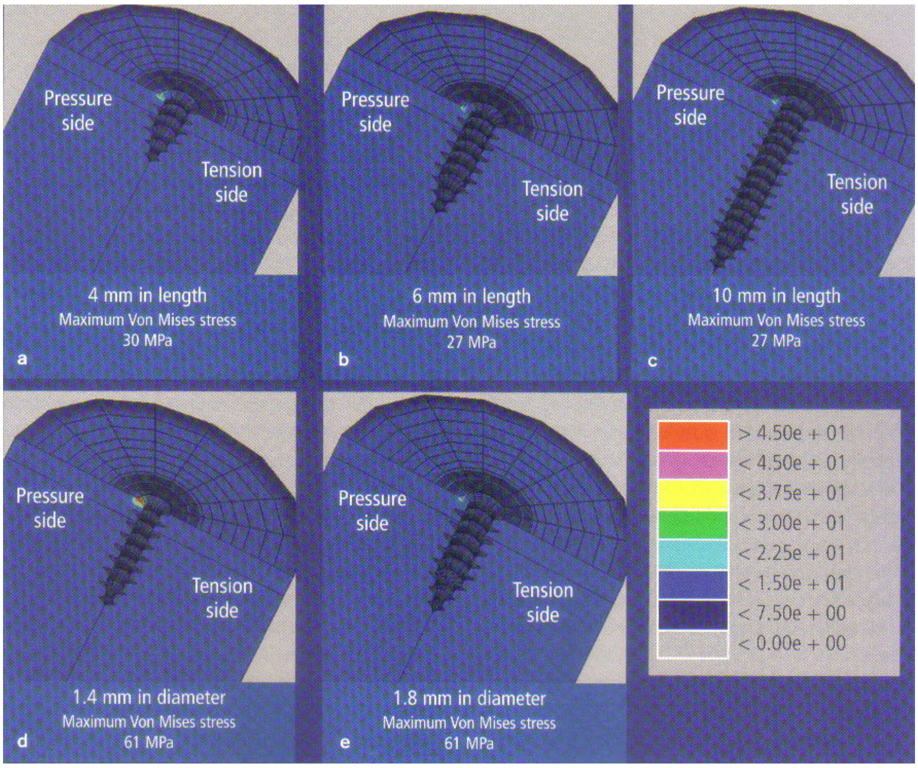


Figure 12 (a to c) von Mises stress distribution according to implant length (1.2-mm cortical bone thickness, 1.8-mm implant diameter, trapezoidal thread, and 200 g orthodontic load). Cortical bone can tolerate 45 to 60 MPa of stress. (d and e) von Mises stress distribution according to diameter (1.2 mm cortical bone thickness, 6 mm implant length, trapezoidal thread, and 220 g orthodontic load)

Motoyoshi et al [16] studied the biomechanical influences of three models that various thread pitches from 0.5 to 1.5 mm using finite element analysis. They found that the thread pitch 0.5 mm model was the least of the maximum stress as compared with the other models but the thread pitch variance did not different in stress distribution pattern.

Fongsamootr et al [15] analyzed an influence of diameter and thread length of mini-screw on the stress distribution in mini-screw and surrounding bone using finite element method. The results showed that the maximum von Misses stress in mini-screw occurs on neck of screw are increased. For maximum principle stress in cortical bone, occur near outer surface of bone surface, its value decrease when diameter and thread length of screw are increases.

CHAPTER IV
MATERIALS AND METHODS

Materials

1. The various available self-drilling Ti-6Al-4V mini-screws were chosen based on each having a characteristic structure with a similar trade diameter and mini-screw length, 1.6 mm in diameter and 6 mm in length.

1.1 Absoanchor® SH1516-06

1.2 O.S.A.S ® 2.5 mm x 1.6 mm x 6 mm

1.3 Dual Top® 1.6 mm x 6 mm

1.4 Orlus® 1.6 mm x 2 mm x (1 mm + 6 mm)

1.5 Remark® 1.6 mm x 6 mm

2. An experimental artificial bone block (Sawbones; Pacific Research Laboratories Inc, Vashon Island, WA, USA) consisted with the 1.5 mm thickness of E-Glass-filled epoxy sheet and the 15 mm thickness of solid rigid polyurethane foam.

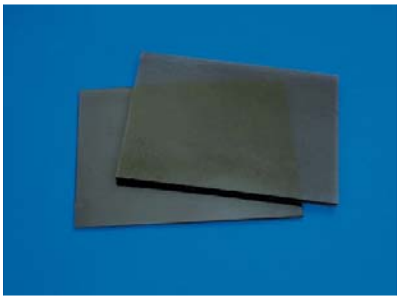
3. Digital torque gauge (MGT 50Z model, Mark-10 Corp.,USA) This torque gauge can display the peak value of torque and the capacity range from 0 to 38 N-cm with the resolution of 0.05 N-cm.

4. X-Y table (Panmanee House Co, Ltd)

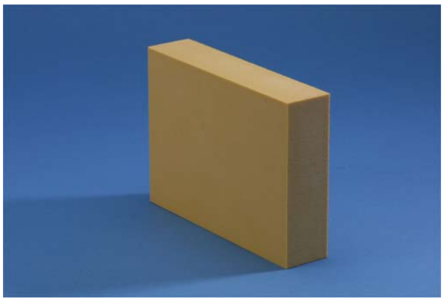
5. Profile projector (Mitutoyo, ModelPH361A)

(a) (b) (c) (d) (e)

Figure 13 The orthodontic mini-screws (a) Absoanchor®, (b) O.S.A.S®, (c) Dual Top®, (d) Orlus® and (e) Remark®



(a)



(b)

Figure 14 The experimental artificial bone block (Sawbones; Pacific Research Laboratories Inc, Vashon Island, WA, USA) consisted with (a) E-Glass-filled epoxy sheet and (b) solid rigid polyurethane foam

Figure 15 Digital torque gauge (MGT 50Z model, Mark-10 Corp., USA) attached with movable arm of X-Y table (Panmanee House Co, Ltd)

Figure 16 Profile projector (Mitutoyo, ModelPH361A)

Methods

The Ti-6Al-4V mini-screws were used to insert in an experimental artificial bone block (Sawbones; Pacific Research Laboratories Inc, Wash) that fabricated and used for measure the insertion torque, as reported by Song et al and Lim et al [26, 84]. The 1.5 mm thickness of E-Glass-filled epoxy sheet were constructed, and attached to solid rigid polyurethane foam using an acrylate bond. The dimensions of block (L×W×H) were 90×60×16.5 mm.

The profile projector (Mitutoyo, ModelPH361A) was used for measure all mini-screws geometry. These data were used for analyze geometrical factor and construct the three dimensional mini-screw model in the finite element study. The geometrical factor in this study were the thread diameter, core diameter, pitch and taper length (Figure 17).

Thread diameter is the largest diameter of the thread portion of the screw that measured over the thread crests.

Core diameter is the smallest diameter of the thread portion of the screw that measured at the thread root.

Pitch is the length between the thread crests.

Taper length is the length of the thread portion that is not parallel.

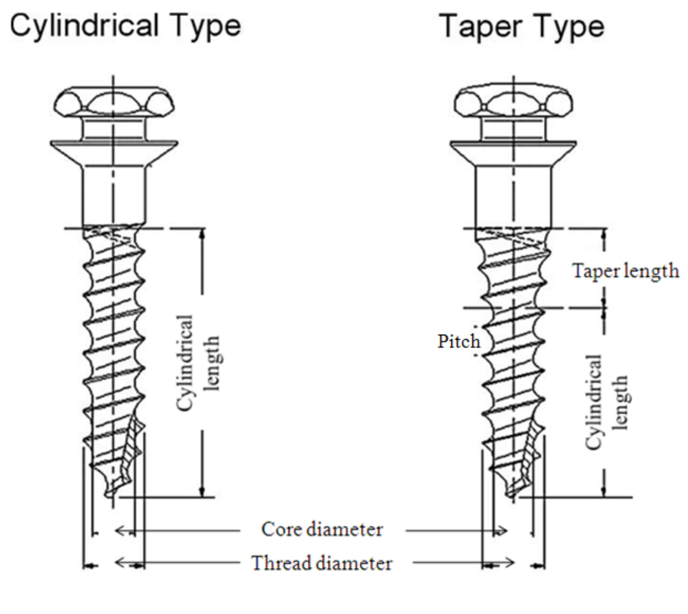


Figure 17 Geometrical measurement in the mini-screws

Maximum insertion torque test

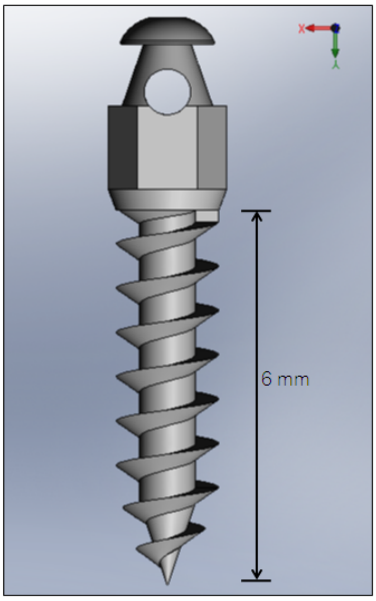
The digital torque gauge (MGT 50Z model, Mark-10 Corp., USA) with the movable arm of X-Y table (Panmanee House Co, Ltd) were used to determine the maximum insertion torque values. After setting the mini-screw tip to contact the artificial bone sample perpendicularly, turned the digital torque gauge with a rotational speed of three rotations per minute until its fracture or unable turning anymore. Six of mini-screws from each manufacturer were implanted into the artificial bone at 10 mm intervals according to ASTM F543-02, maximum torque should be tested with a minimum of five screws ad spacing should have as minimum distance of 5X the diameter of the screw [94].

Figure 18 The setting of mini-screw contacted the artificial bone

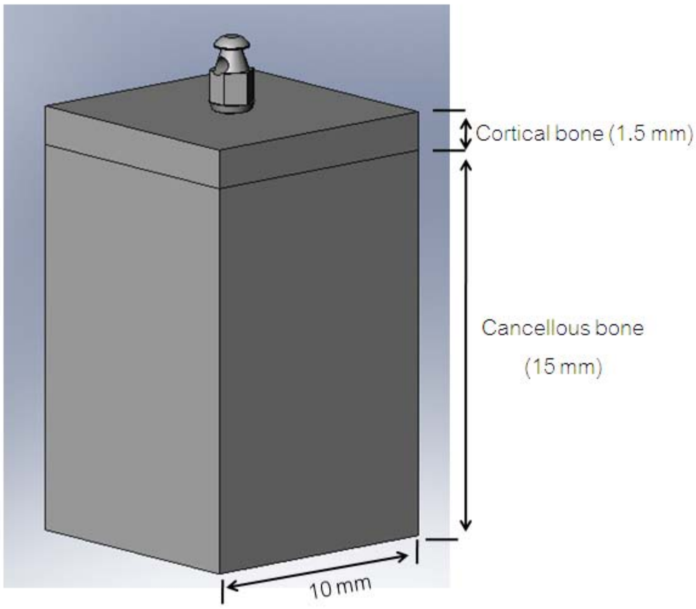
Finite element study

Preparation for finite element model

Three dimensional finite element models were constructed with the SolidWorks 2007 CAD program. The human bone models were created with homogenous 1.5 mm cortical and 15 mm cancellous bone thickness. The mini-screw models were assumed to be made of titanium alloy (Ti-6Al-4V). Shape and size were simulated as the mini-screw in the maximum insertion torque test that were measured with profile projector and used average values to construct the mini-screw models. Both bone and mini-screw models were assumed to be homogenous, isotropic, and linearly elastic. Material properties of the finite element models were shown in Table 5. The mini-screw models were inserted into the bone models from cortical bone surface until thread portion of the mini-screws were completely embedded. Each model was meshed in ABAQUS Version 6.7 (Inc., 1080 Main Street, Pawtucket, RI) program. The element type was a 10 node modifies quadrati tetrahedral (C3D10M). The total number of nodes and elements for the model were approximately 100,000 and 520,000, respectively on average.



(a)



(b)

Figure 19 (a) diagram of the mini-screw model design (1.6 mm diameter and 6 mm length), (b) detail of the mini-screw and the bone model dimension (cortical bone thickness 1.5 mm, cancellous bone thickness 15 mm. and width of bone block 10 mm)

Table 5 Material properties for the constituent material [95, 96]

Material	Young's Modulus (MPa)	Poisson's Ratio
Ti-6Al-4V	115,000	0.34
Cortical bone (test)	12,400	0.26
Cancellous bone (test)	1,190	0.30
Cortical bone (actual)	13,700	0.26
Cancellous bone (actual)	1,370	0.30

Boundary condition and solution phase

In this study used the couple concentrated force that converted from the lowest mean among each manufacturer of the maximum insertion torque value applied at the head of the mini-screw parallel to the Z axis. The interface between the mini-screw and the surrounding bone element was fixed, as the most tightening condition was assume. Nodes surrounding the bone elements were restricted to three degrees of freedom. The contact surface between cortical and cancellous bone model was ties together. The outer surface of the bone model was constrained.

Postprocessing phase

The assessments of the stress distribution pattern on the mini-screw and the surrounding bone model were performed using the von Mises' equivalent stress and the first principal stress at each nodal point that were calculated using the ABAQUS program .

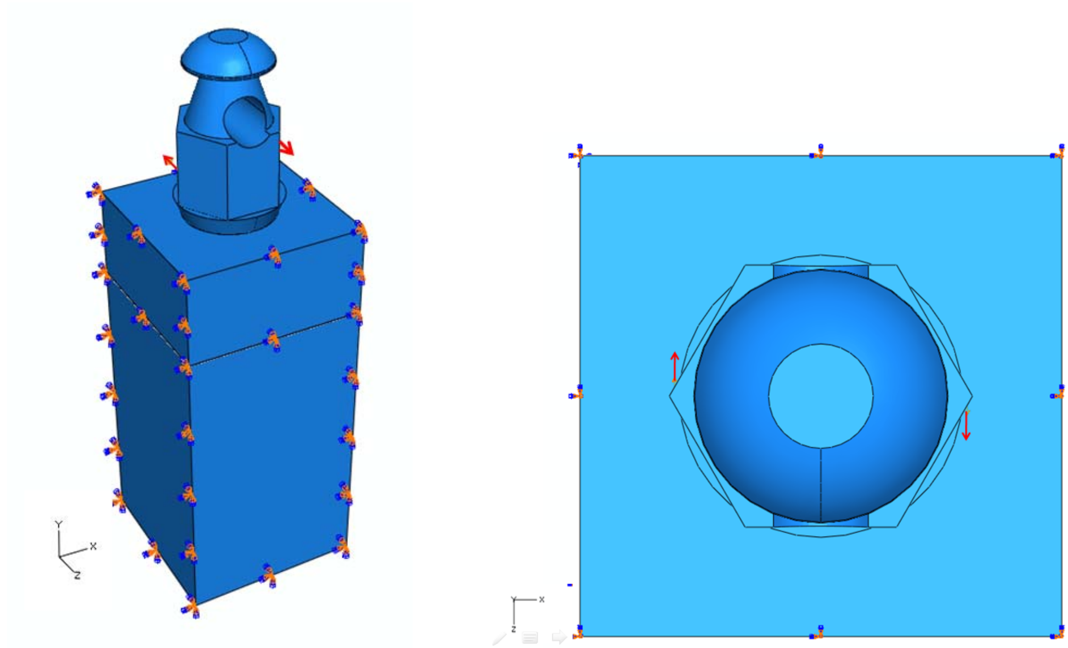


Figure 20 Direction and position of force application on the mini-screw model

Statistical Analysis

Means and standard deviation were used to describe the maximum insertion torque values and the geometry of the mini-screws.

Pearson correlation was used to determine the correlation between the geometry of the mini-screws and the maximum von Mises' equilavent stress.

CHAPTER V
RESULTS

The results of this study that determined the design parameter variation of the mini-screws using finite element analysis were divided into three parts as follows;

1. The geometrical dimension of the mini-screws
2. The maximum insertion torque in the artificial bone block
3. The stress distribution in the mini-screws and their surrounding bone in the finite element study

The Geometrical Dimension of the Mini-screws

The geometrical dimension of the mini-screws used in present investigation were measured with profile projector and listed in Table 6. The mean value for geometry dimension of mini-screws are carried out in millimeter (mm) and demonstrated in the Table 6.

Table 6 The geometrical dimension of mini-screws

Manufacturer	Measurement (mm)			
	Thread diameter	Core diameter	Pitch	Taper length
Absoanchor®	1.50	0.82	0.67	-
O.S.A.S ®	1.55	1.11	0.61	-
Dual Top®	1.62	1.00	0.73	-
Orlus®	1.58	0.91	0.74	1.77
Remark®	1.64	0.90	0.70	1.49

The data of the measurement samples in this study could divided the mimi-screws into two type, cylindrical and taper type. The cylindrical type was composed of a parallel thread along the whole length of the thread part, showing different the core and thread diameter and pitch among Absoanchor®, O.S.A.S® and Dual Top®. The taper type had different range of increasing the core and thread diameter, pitch and taper length between Orlus® and Remark®. Although this study was selected all mini-screws based on having the characteristic structure with the similar trade 1.6 mm in diameter and 6 mm in length.

The Maximum Insertion Torque in the Artificial Bone Block

In the maximum insertion torque experiment was found that all mini-screws of absoanchor® fractured when the thread portions were completely implanted in the artificial bone block and then turned them until unable turning anymore. The portion of all mini-screw fractures was the neck portion which between the first and the second thread of the mini-screws. The maximum insertion torque value of these mini-screws is listed in Table 7.

Figure 21 The fracture portions of Absoanchor® mini-screws

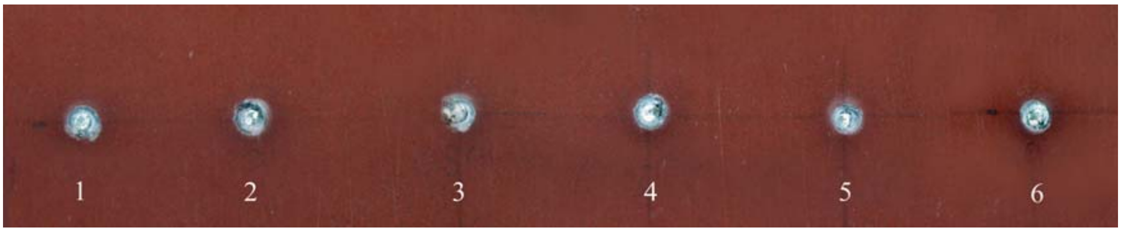


Figure 22 The implanted portion in the artificial bone block of Absoanchor® mini-screws

Table 7 The maximum insertion torque value of absoanchor ® mini-screws

No.	Maximum insertion torque (N-cm)
1	20.10
2	19.60
3	21.50
4	18.75
5	20.60
6	19.85
Mean$\pm$SD	20.07$\pm$0.93

However, these fracture characteristic were not found in the other manufacturers. All of them could completely implant in the artificial bone block and the maximum insertion torque value were higher than torque gauge capacity that showed maximum 38 N-cm.

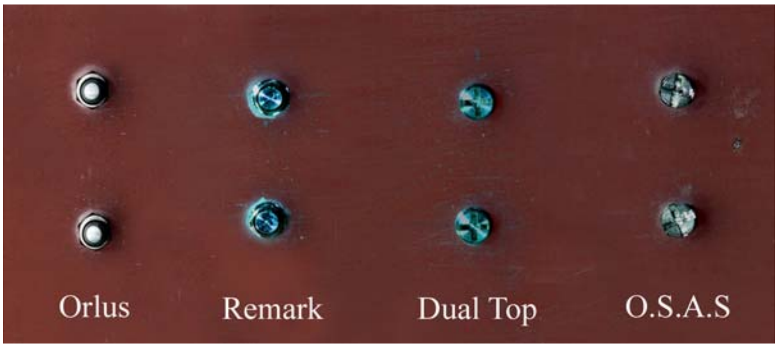


Figure 23 The completely implanted of Orlus®, Remark®, Dual Top®, and O.S.A.S ® mini-screws

Table 8 The comparison of the maximum insertion torque value among five mini-screw manufacturer

Manufacturer	**Maximum insertion torque (N-cm)**
Absoanchor®	20.07 $\pm$ 0.93
O.S.A.S®	> 38*
Dual Top®	> 38*
Orlus®	> 38*
Remark®	> 38*

* The torque value was above the digital torque gauge capacity

The Stress Distribution in the Mini-screws and their Surrounding Bone in FEA

In the FEA, the lowest of the maximum insertion torque value among five manufacturers was selected for use in the finite element program. This value was the maximum insertion torque in Absoanchor®, 20.07 N-cm, and converted to the concentrated force that were applied as the couple force at the head portion of all mini-screws. The force depended on head diameter of mini-screws so there were different forces application at the head of mini-screws that are shown in Table 9. The stress concentration pattern of all mini-screw models were mainly on the neck part of mini-screw, in the area between the thread runout and the second thread. The stress concentration pattern of cortical bone models was at the upper part of model that around the thread runout, the first and the second thread. However the stress concentration of cancellous bone models was not obvious. The maximum Von Mises' stress of mini-screw model, the maximum first principal stress in cortical and cancellous models in each manufacturer are also shown in Table 9.

Table 9 Stress value in different mini-screw models with torsional force

Manufacturer	Head diameter at force application (mm)	Force application at the head (N)	Maximum of von Mises' stress (MPa)	Maximum first principal stress (MPa)	
				Cortical bone	Cancellous bone
Absoanchor®	1.73	115.88	2305.9	536.75	2.45
O.S.A.S ®	3.08	65.09	1,088	300	1.99
Dual Top®	3.17	63.27	1,002	371	1.20
Orlus®	2.83	70.94	785	188	0.72
Remark®	2.73	73.98	444	112	0.60

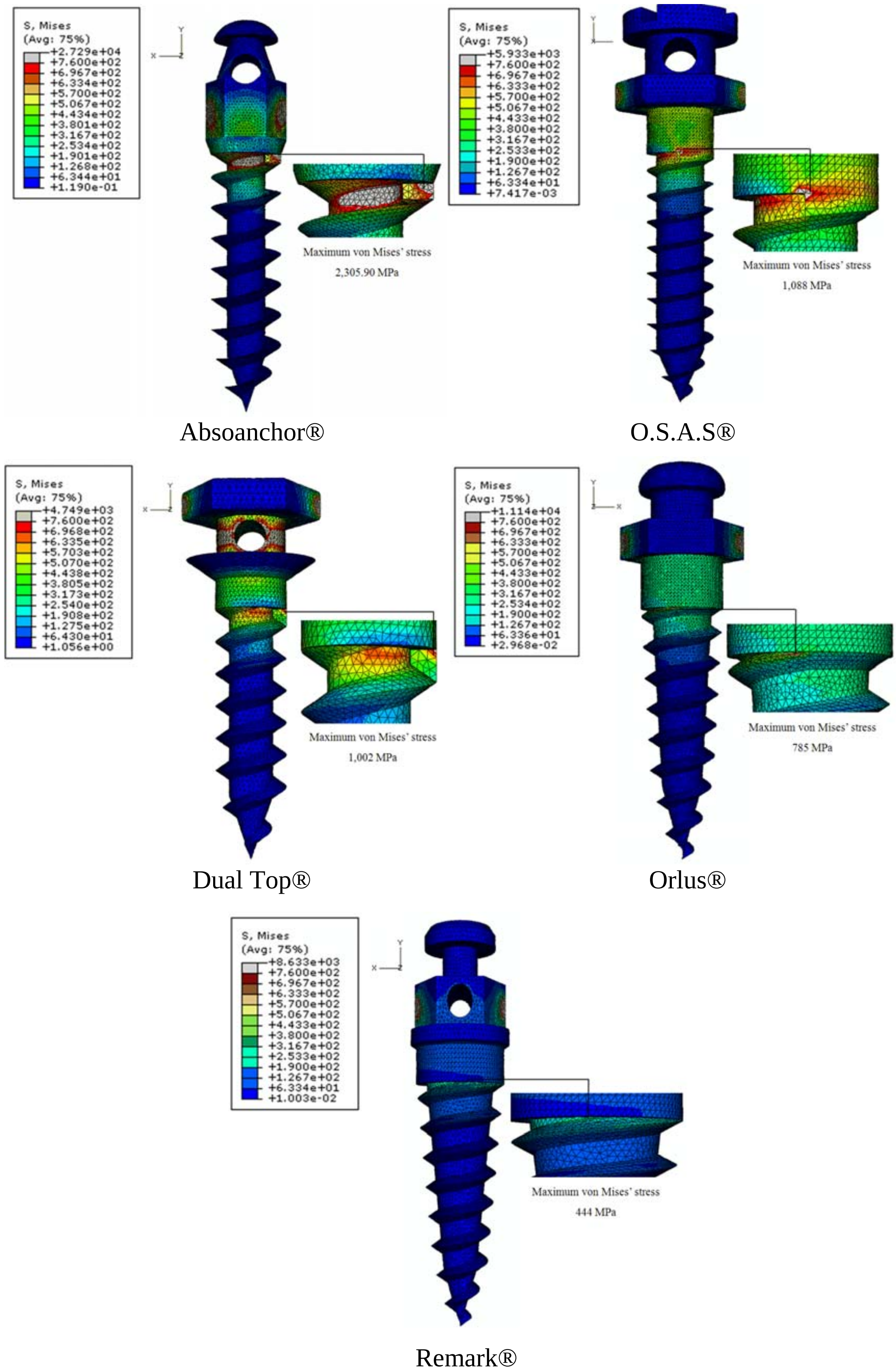


Figure 24 Pattern of the stress distribution in all mini-screw models

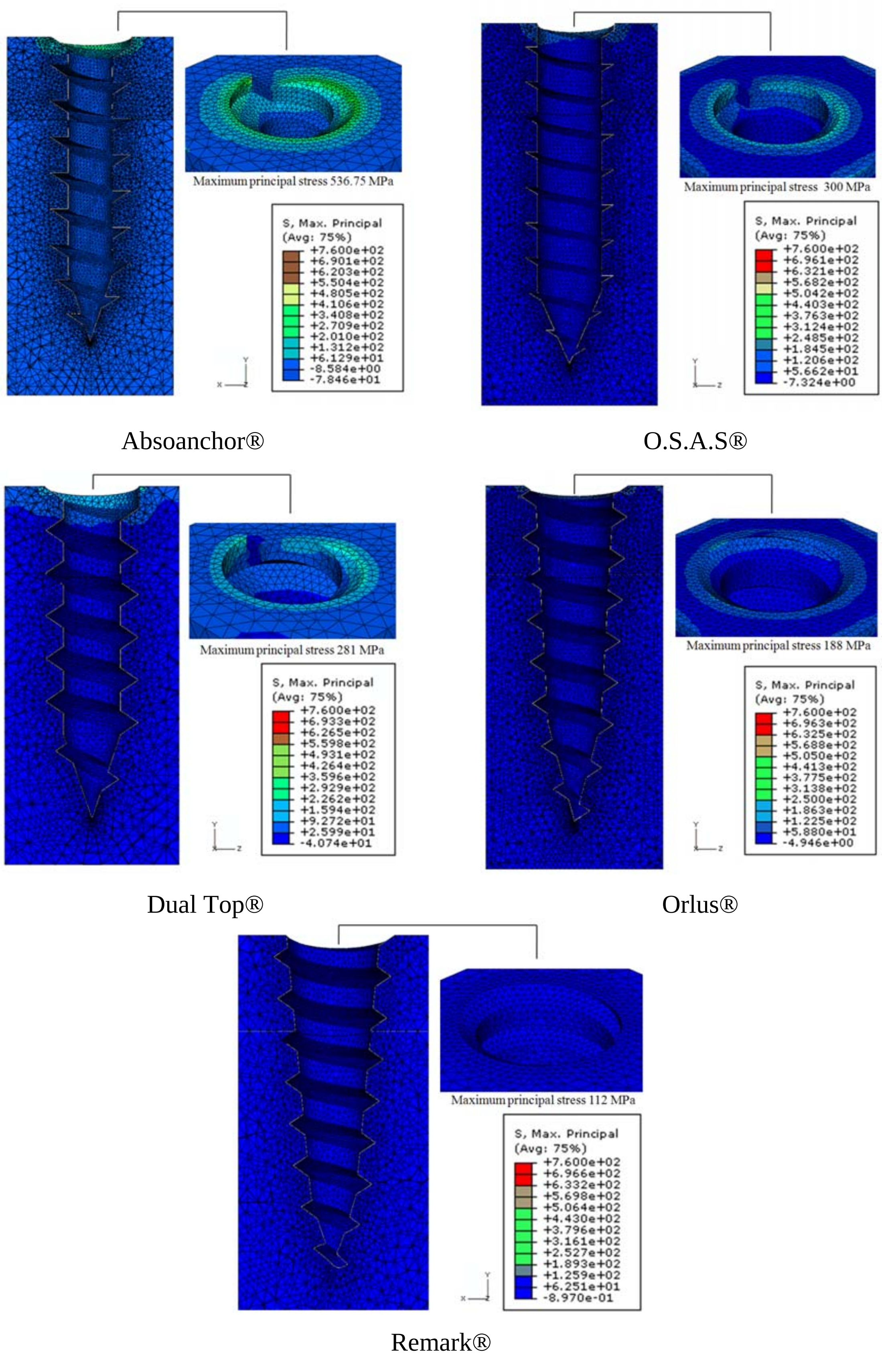
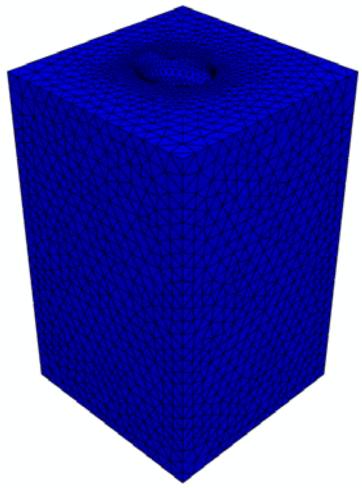
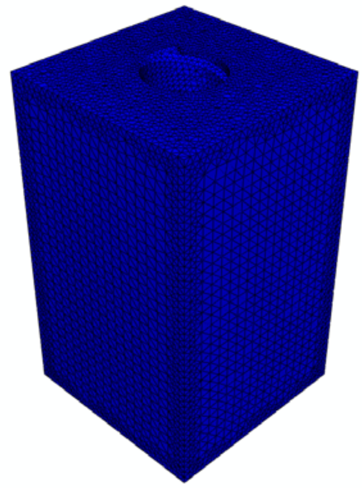


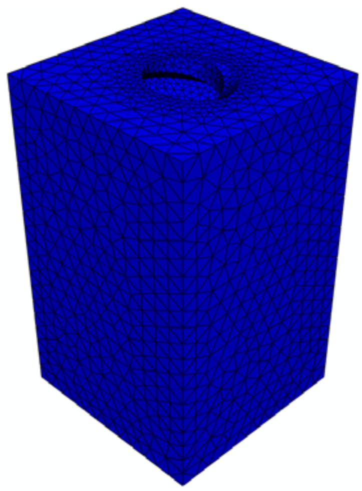
Figure 25 Pattern of the stress distribution in all cortical models



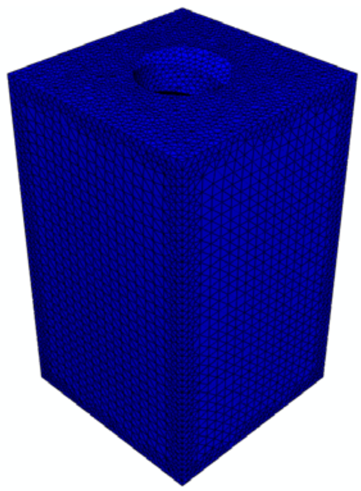
Absoanchor®



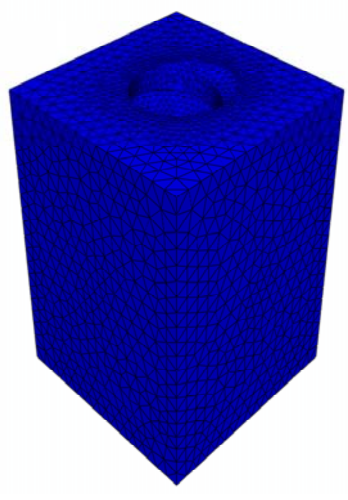
O.S.A.S®



Dual Top®



Orlus®



Remark®

Figure 26 Pattern of the stress distribution in all cancellous models

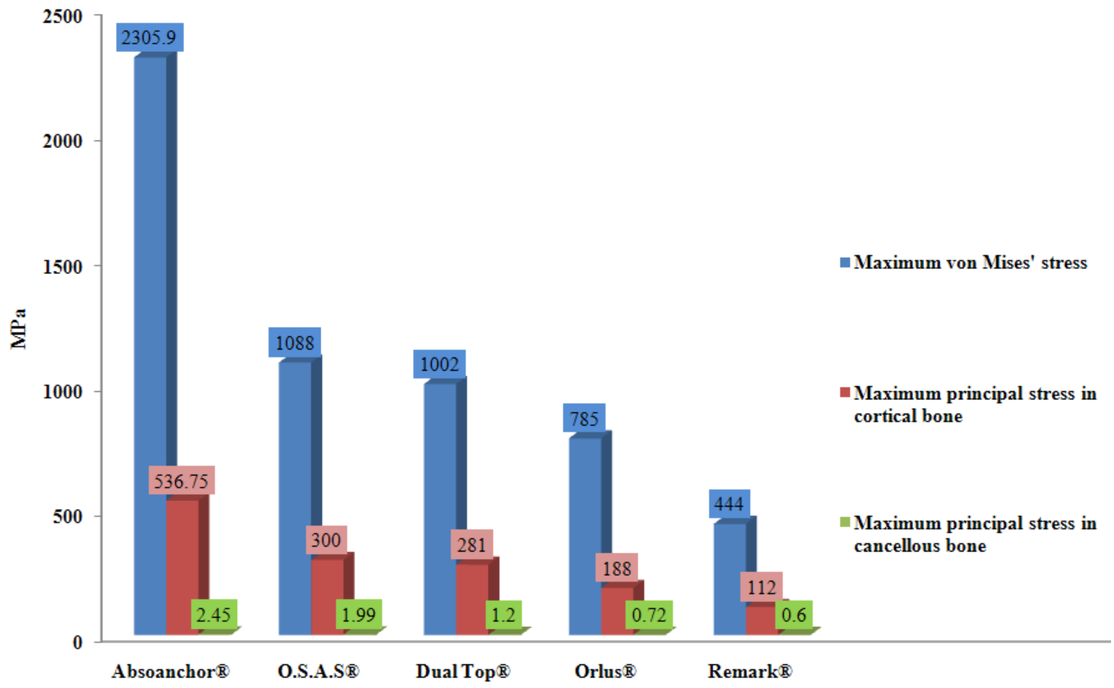


Figure 27 The comparison of the maximum von Mises' stress of the mini-screw models, the maximum first principal stress in the cortical and cancellous models in each manufacturer

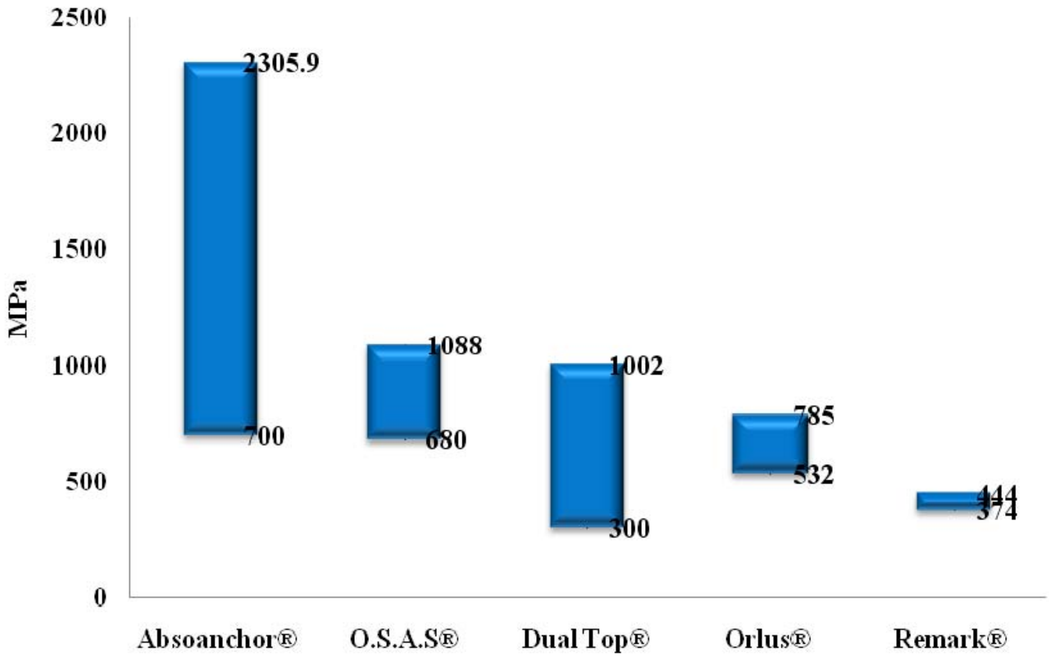


Figure 28 The comparison of the von Mises' stress range at the high stress concentration area in the mini-screw models

Table 10 The correlation between the geometrical parameters of the mini-screws and the maximum von Mises's stress

Geometrical parameter	**Maximum von Mises' stress**	
	Pearson correlation	**p-value**
Thread diameter	-0.883	0.024*
Core diameter	-0.377	0.266
Pitch	-0.334	0.292

*A significant difference was detected at p-value < 0.05

The comparison of maximum von Mises' stress of the mini-screw model, the maximum first principal stress in the cortical and cancellous models in each manufacturer showed that Absoanchor® had the highest stress in all model, followed by O.S.A.S®, Dual Top®, Orlus® and Remark® respectively (Figure 27). These stresses were associated with the thread diameter of the mini-screws according to Pearson correlation coefficient indicated negative correlation between the thread diameter of the mini-screws and the maximum von Mises's stress at p-value = 0.024. The smaller thread diameter had higher stress than the larger thread diameter. Furthermore, this correlation was also tendency found in the range of the high stress concentration area of mini-screw models. The smaller diameter presented in the larger area of high stress concentration than the larger diameter. The range of high stress concentration of the mini-screw models are shown in Figure 28.

CHAPTER VI
DISCUSSION

The experiment method in this study was designed for measuring the maximum insertion torque of the mini-screws, then evaluated and compared with the finite element model. The artificial bone block composed with the E-Glass-filled epoxy sheets (a mixture of short E-Glass fibers and epoxy resin) and the solid polyurethane foam which were used as a substitute for human cortical and cancellous bone respectively. The solid polyurethane foam was used as a test medium for testing the metallic bone screw according to the American Society for Testing and Material (ASTM) F 1839-01 specification [95, 96]. The E-Glass-filled epoxy sheet had a density of 1.7 g/cc and 0.64 g/cc for the solid polyurethane foam. These material properties are in the range of human bone, based on the previous research of Misch et al [97], the density of the mandible ranges from 0.85-1.53 g/cc, with average of 1.14 g/cc. Because of the material properties are uniform and in approximate range of human bone, the amount of torque can be compared.

The mean of maximum insertion torque value of Absoanchor® mini-screws in this investigation was 20.07 N-cm. Therefore, the researches of Song et al [26] and Lim et al [84] used the different sizes of mini-screw from this study (1.5 mm diameter, 6 mm and 7 mm length) , the maximum insertion torque of the cylindrical mini-screw shape in the same artificial bone block condition were approximately to this study, 20.33 and 19.5 N-cm in sequence. Otherwise, Dual-Top® and O.S.A.S® were also the cylindrical mini-screw shape which had higher the maximum insertion torque value, more than 38 N-cm (above the digital torque guage capacity). In clinical study of Sirnok [87], the maximum insertion torque was lower than these three cylindrical manufacturer that ranged from 4.62-7.08 N-cm. The mini-screws of Srinok study were 1.4 mm diameter and 7 mm long.

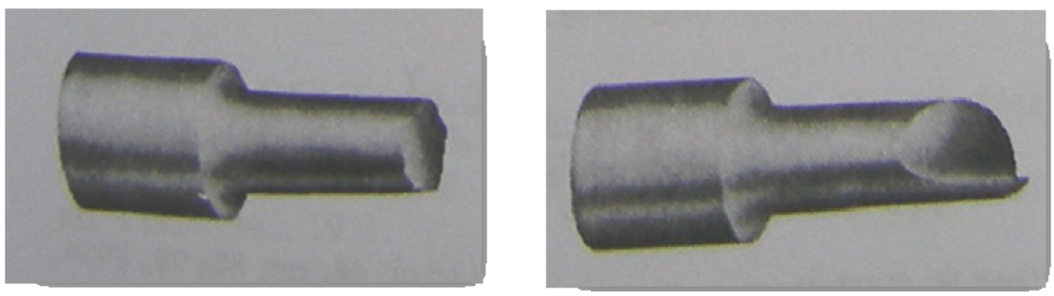
In taper type mini-screws, Remark® and Orlus® had the maximum torque more than 38 N-cm which were similarity in Song et al [26] (1.6 mm diameter, 6 mm length), 37.89 N-cm. When compared with the previous clinical study of Motoyoshi et al [11]. The mean of maximum insertion torque value were higher than it. The mean insertion torque of their study ranged from 7.2 to 13.5 N-cm. However, their mini-screws were the 1.6 mm diameter and 8 mm length.

Table 11 Summary of the past clinical and laboratory studies of mini-screws

Researcher	Type of study	Type of screw	Diameter (mm) × Length (mm))	Insertion torque (N-cm)
Song et al (2007) [26]	Laboratory	Cylinder Taper Taper	1.5×6 1.6×6 1.6×7	20.33 37.89 36.20
Lim et al (2008) [84]	Laboratory	Cylinder Cylinder Cylinder Taper	1.5×7 1.5×8 1.5×9 1.5×6	19.50 20.90 23.00 32.60
Motoyoshi et al (2006) [10]	Clinical	Taper	1.6×8	7.20-13.50
Srinok (2008) [87]	Clinical	Cylinder	1.4×7	4.62-7.08
This study	Laboratory	Cylinder - Absoanchor® - O.S.A.S® - Dual Top® Taper - Orlus® - Remark®	1.6×6	 20.07 >38 >38 >38 >38

The studies of the torque resistance using the artificial bone can be a good method of the mechanical study of the mini-screws. Nevertheless, there will be some differences of the torque value during insertion in the vital bone. In clinical studies of the maximum insertion torque were lower than this study because of the location, size, insertion method, however, there is a limitation to comparison of torque value between the different study models.

All of Absoanchor® mini-screws in this study were fracture. This risk was agreed with Anka suggestion that the sites with dense bone, where the final torque value can be above 20 N-cm, should be pre-drilled to avoid fracturing the mini-screw [42]. According to the type of torsion failure are divided in to two types, the shear and the tensile failure [98]. The fracture of Absoanchor® mini-screws was the torsion failure because of its fracture plane similar to a ductile metal that fails by shear along one of the planes of maximum shear stress. Generally the plane of the fracture is normal to the longitudinal axis (Figure 29a). On the other hand a brittle material fails in torsion along a plane perpendicular to the direction of the maximum tensile stress. Since this plane bisects the angle between the direction of the maximum shear stress and makes an angle of 45^{o} with the longitudinal and transverse directions, it results in a helical fracture (Figure 29b). In this study situation maybe different from the clinical situation according to the direction when implanted the mini-screw. In the clinic, the operators sometimes has bending force to implant the mini-screw that causes the mini-screw fracture in the different portion from this study.

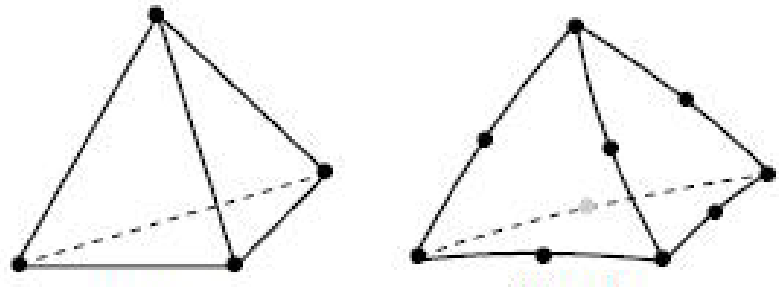


(a) (b)

Figure 29 Typical torsion failure, (a) shear (ductile) failure and (b) tensile (brittle) failure [98]

In the past two decades, FEA has become an increasing tool for the prediction of the effects of stress on the dental implant and its surrounding bone [99]. This method can obtain a detailed representation of many different factors that affect the biomechanical behavior of bone. According to previous studies of the mini-screw, they evaluated the effect of orthodontic loading to mini-screws [15-17]. Thus, these finite element model were the models for analysis the torque resistance of the mini-screws.

The element type in this study was the ten nodes tetrahedral element that will give the accuracy than the four nodes tetrahedral element [15]. Because of the mini-screws had the small size and the complexity of the design.



(a) (b)

Figure 30 The element type, (a) four nodes tetrahedral and (b) ten nodes tetrahedral

The von Mises's stress was used to evaluated the mini-screw models in this study. According to the octahedral shear stress yield criteria used for ductile material, such as titanium. Ductile material can be elongated by force before it fractures. Yielding of this type of material occurs when the shear stress on the octahedral planes reaches a critical value. The resulting octahedral shear stress yield criterion, also often called either the Von Mises or the distortion energy criterion, represents an alternative to the maximum shear criterion. Therefore, maximum stress in mini-screw made of titanium were calculated according to the octahedral shear stress yield criterion theory, and the maximum stress evaluated in this study was called the maximum von Mises' stress.

The stress in cortical and cancellous bone models were calculated following by the Coulomb-Mohr fracture criterion theory. Fracture is hypothesized to occur on a given plane in a brittle material, such as bone, when a critical combination of shear and normal stresses acts on that plane. Brittle material cannot be elongated as much as ductile material before it fractures. Fracture strength in compression of brittle material is greater than that in tension. In the simplest application of this approach, the mathematical function giving the critical combination of stresses is assumed to be a linear relationship. Thus, the maximum stress evaluated bone in this study was called maximum first principal stress.

The result of the maximum insertion torque experiment of Absoanchor® mini-screws confirmed that the damaging area of mini-screw was the same area which had the high stress concentration in finite element model. The weak area was between the thread runout and the second thread of the neck portion of the mini-screw. This stress concentration areas agree with several studies that reported the mini-screws breaking at the neck portion during the screw removal [61, 100]. The range of von Mises' stress at these area was about 400-2,300 MPa that some area were above ultimate shear strength of Ti-6Al-4V material, 760 MPa. These were the critical point for damaging of mini-screws. However, the stress distribution patterns of the other manufacturers showed that the stress was also mainly in the neck portion of the mini-screws and this risk had been found in the other cylindrical type, O.S.A.S® and Dual Top®, the range of von Mises' stress were 680-1,088 and 300-1,002 MPa respectively. In the other hand, taper type mini-screw, the range of von Mises' stress was below ultimate shear strength only in Remark® mini-screw. In cases of O.S.A.S®, Dual Top® and Orlus® that some area had the von Mises' stress more than ultimate shear strength but the fracture of the mini-screws did not occur. It can describe in term "fracture mechanics", the transfer of mechanical energy toward the creation of crack surfaces [101]. Titanium alloy is the ductile material, some plastic deformation will occur at the crack tip. Therefore, all mini-screws in this study were the same material but fracture behavior may different according to design parameters like critical load, critical crack tip opening displacement or fracture toughness [102]. The most important parameter was the thread diameter due to Pearson correlation (r = -0.883) and went a long with the past studies about the insertion torque in laboratory [84, 86] and about affected parameter associated with orthodontic load in FEA studies [15, 17, 103]. Moreover the design parameter, the distinguished factor among these manufacturers was the contact surface area between the mini-screw and its surrunding bone that associated with the diameter, pitch, length and shape of the mini-screw.

According to mini-screw models, the portion that might be corncern and further study was the countersink of the mini-screw. Because of it transmited the force from the head to the thread portion. The range of the stress distribution were vary in all mini-screw models that are shown in Figure 32.

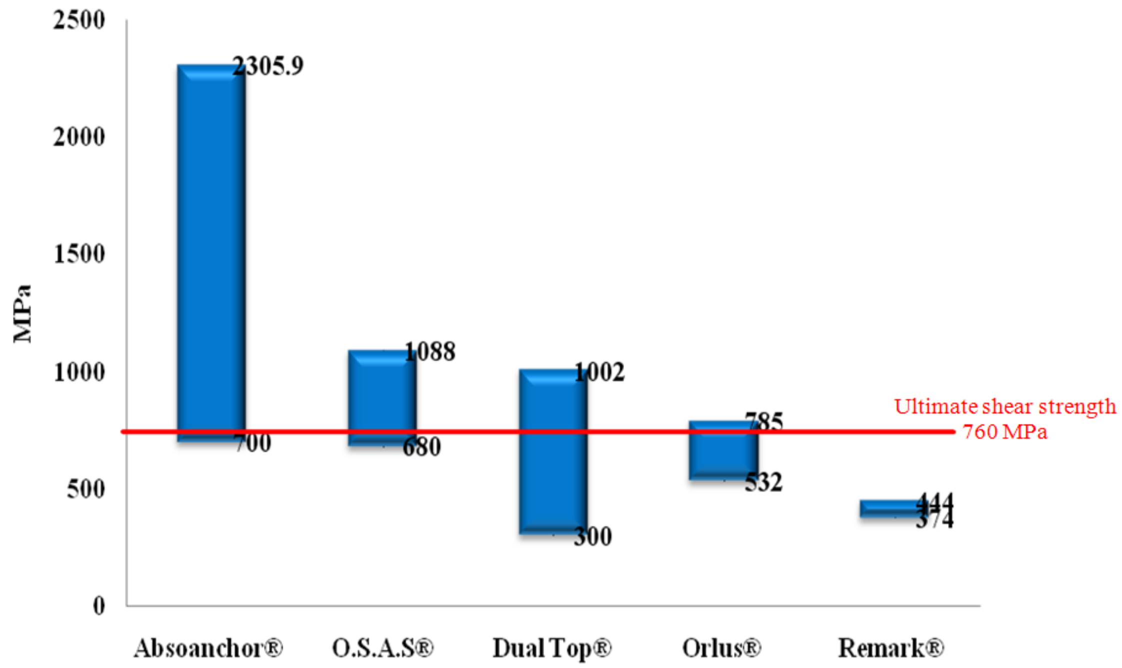


Figure 31 The range of the von Mises' stress of the mini-screw models between the thread runout and the second thread compared with the ultimate shear strength

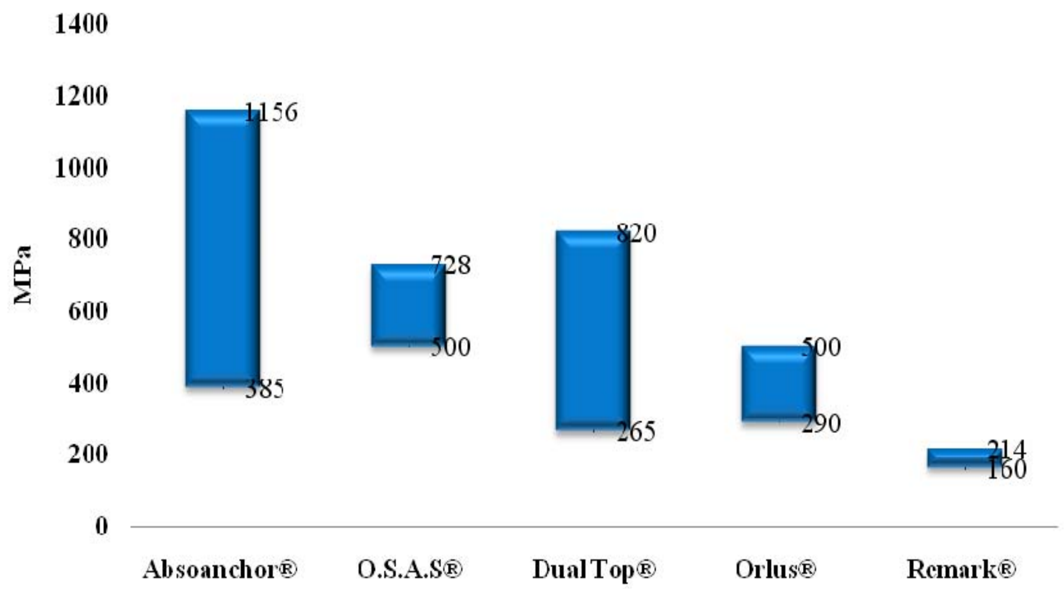


Figure 32 The comparison of the von Mises' stress range at the countersink stress distribution area in mini-screw models

The stress distribution patterns of bone models showed that the stress concentrated in the cortical bone more than in the cancellous bone, especially in the upper part of cortical bone. Additionally, the stress values in mini-screws were the highest, followed by in the cortical and cancellous bone models, respectively. The cortical bone showed the stress values approximately 150 to 260 times higher than in the cancellous bone. Therefore, the results of this study suggest that the cortical bone is the most important critical part of bone which against to force application [103].

According to a role of strain on bone physiology [104], loads on bone cause bone strains that generate signals that some cells can detect and to which they or other cells can respond. Without such strains, disuse-mode remodeling tends to remove a callus while modeling tends to stay off, so bone healing can retard or fail. Excessive strains (gross motion) can usually prevent bony union. The naturally permissible strains might lie in the 100-2000 microstrain span; about 2 MPa-40 MPa (Table 12). One thousand microstrain equals a 0.1 %stretch or shortening.

Table 12 Set point values for bone's thresholds and ultimate strength (in microstrain, stress and unit-load terms)

MESr, 50-100 microstrain; ~ 1-2 MPa, or ~0.1 kg/mm2
MESm, 1000- 1500 microstrain; ~ 20 MPa, or ~2 kg/mm2
MESp , ~ 3000 microstrain; ~ 60 MPa, or ~6 kg/mm2; This also approximately equals bone's yield point.
Fx, ~ 25,000 microstrain; ~120 MPa or ~12 kg/mm2 in healthy young adult mammals.

MESr, bone's genetically determined disuse-mode threshold strain range, below which the maximum disuse-mode activity occurs and above which it deigns to declined or turn off

MESm, bone's genetically determined modeling threshold range strain, in and above which modeling usually turns on to strengthen a bone

MESp, bone's genetically determined operational microscopic damage in bone threshold strain range, in and above which unrepaired microscopic damage can begin to accumulate

Fx, a bone's fracture strength or ultimate strength

The updated bone physiology suggests that the design of load-bearing endoprotheses should keep typical peak strains in the bone supporting implants below the bone's microscopic damage threshold but let those strains exceed bone' MESr, and perhaps exceed its MESm but much less than its MESp. This strain makes modeling strengthen the supporting bone but help to keep disuse-mode remodeling from removing it. When the strains exceed MESp, then bone's microdamage damage accumulation would usually occur and lead to nontraumatic and stress fractures.

An experimental study of Melsen and Lang [105] about the excessive force that applied to the dental implants using for the orthodontic anchorage, the range of functional strain between 3,400-6,700 µ strain was higher than Frost's study. They reported that functional strain was to maintain a normal bone remodeling rate, whereas the strain above this range caused a high percentage of bone resorptive surface.

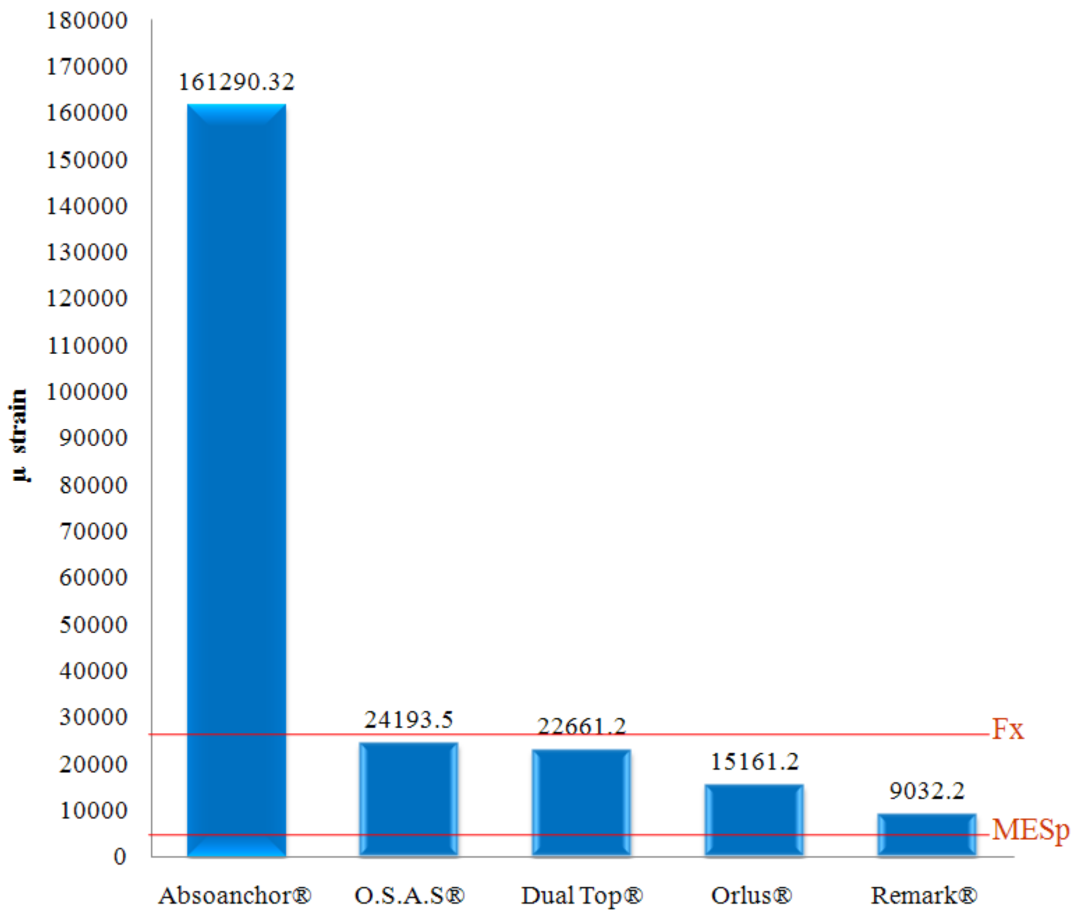


Figure 33 The maximum strain in the bone models compared with MESp and Fx of Frost's study

The strain value in all cortical and cancellous bone of this investigation which referred from the maximum first principal stress value, were excessive strain that identified by Frost [104] and Melen and Lang [105], the maximum strain value was 161,290.32, 24,193.50, 22,661.20, 15,161.20 and 9,032.2 μ strain in the cortical bone of Absoanchor®, O.S.A.S®, Dual Top®, Orlus® and Remark® respectively (Figure 33). Therefore, the surrounding bone of all manufacturers were harmful from the microdamaging and the high percentage of bone resorption with the torsional force 20.07 N-cm applied to the mini-screws. This adverse effect to supporting bone should not disregard to consider the optimal design of the mini-screw.

CHAPTER VII
CONCLUSIONS

1. The maximum insertion torque of the mini-screws in the artificial bone block could determine in only Absoanchor®, 20.07 N-cm, according to the limitation of the digital torque gauge capacity.

2. The finite element method could explain the interaction between the mini-screw and its surrounding bone with torsional force application. The stress distribution patterns of the mini-screw models showed that the neck potion, from the thread runout to the second threads, was the weak point of the body. Furthermore, the stress concentration area was similar to the fracture portion of Absoanchor® mini-screws. The stress distribution patterns of the bone models showed that the stress concentrated in the cortical bone more than in the cancellous bone, especially in the upper part of the cortical bone. Additionally, the stress values in the mini-screws were highest, followed by in the cortical and cancellous bone models, respectively.

3. The design parameter that had correlation with the maximum von Mises' stress in the mini-screws was the thread diameter, r = -0.883. Moreover, it could be implied to the contact surface area between the mini-screw and its surrounding bone was the important factor related with the stress concentration.

4. The clinical implications from this study, the most concerning area is the neck portion of the mini-screw and when implanted the mini-screw, the operator should be avoid the bending force to the mini-screw. It will decrease risk of the mini-screw fracture in the different portion from this study. In addition to the high insertion torque value above 20 N-cm, pre-drilled the mini-screw is also the optional method to avoid the mini-screw fracture.

5. The suggestions for further studies are to investigate the changing in each parameter using finite element method for the optimized design of the mini-screw including biomechanical consideration of its surrounding bone.

REFERENCES

[1] Proffit WR, Fields HW, Ackerman JL, Bailey LJ, Tulloch JFC. The biological basis of orthodontic therapy. Contemporary orthodontics. 3rd ed. St. Louis: Mosby 2000:295-325.

[2] Kyung HM, Park HS, Bae SM, Sung JH, Kim IB. Development of orthodontic micro-implants for intraoral anchorage. J Clin Orthod. 2003 Jun;37(6):321-8; quiz 14.

[3] Roberts WE, Nelson CL, Goodacre CJ. Rigid implant anchorage to close a mandibular first molar extraction site. J Clin Orthod. 1994 Dec;28(12):693-704.

[4] Motoyoshi M, Matsuoka M, Shimizu N. Application of orthodontic mini-implants in adolescents. Int J Oral Maxillofac Surg. 2007 Aug;36(8):695-9.

[5] Kuroda S, Yamada K, Deguchi T, Hashimoto T, Kyung HM, Takano-Yamamoto T. Root proximity is a major factor for screw failure in orthodontic anchorage. Am J Orthod Dentofacial Orthop. 2007 Apr;131(4 Suppl):S68-73.

[6] Chen YJ, Chang HH, Huang CY, Hung HC, Lai EH, Yao CC. A retrospective analysis of the failure rate of three different orthodontic skeletal anchorage systems. Clin Oral Implants Res. 2007 Dec;18(6):768-75.

[7] Chen CH, Chang CS, Hsieh CH, Tseng YC, Shen YS, Huang IY, et al. The use of microimplants in orthodontic anchorage. J Oral Maxillofac Surg. 2006 Aug;64(8):1209-13.

[8] Cheng SJ, Tseng IY, Lee JJ, Kok SH. A prospective study of the risk factors associated with failure of mini-implants used for orthodontic anchorage. Int J Oral Maxillofac Implants. 2004 Jan-Feb;19(1):100-6.

[9] Miyawaki S, Koyama I, Inoue M, Mishima K, Sugahara T, Takano-Yamamoto T. Factors associated with the stability of titanium screws placed in the

posterior region for orthodontic anchorage. Am J Orthod Dentofacial Orthop. 2003 Oct;124(4):373-8.

[10] Motoyoshi M, Yoshida T, Ono A, Shimizu N. Effect of cortical bone thickness and implant placement torque on stability of orthodontic mini-implants. Int J Oral Maxillofac Implants. 2007 Sep-Oct;22(5):779-84.

[11] Motoyoshi M, Hirabayashi M, Uemura M, Shimizu N. Recommended placement torque when tightening an orthodontic mini-implant. Clin Oral Implants Res. 2006 Feb;17(1):109-14.

[12] Zhang Y, Zhang D, Feng CJ. A three-dimensional finite element analysis for the biomechanical characteristics of orthodontic anchorage micro-implant. Shanghai Kou Qiang Yi Xue. 2005 Jun;14(3):281-3.

[13] Himmlova L, Dostalova T, Kacovsky A, Konvickova S. Influence of implant length and diameter on stress distribution: a finite element analysis. J Prosthet Dent. 2004 Jan;91(1):20-5.

[14] Holmgren EP, Seckinger RJ, Kilgren LM, Mante F. Evaluating parameters of osseointegrated dental implants using finite element analysis--a two-dimensional comparative study examining the effects of implant diameter, implant shape, and load direction. J Oral Implantol. 1998;24(2):80-8.

[15] Fongsamootr T, Seehawong N. Parametric analysis of miniscrew implant on stress distribution in the bone using Finite Element Method. 20th ME-NETT Conference; 2006; Nakhon Ratchasima; 2006.

[16] Motoyoshi M, Yano S, Tsuruoka T, Shimizu N. Biomechanical effect of abutment on stability of orthodontic mini-implant. A finite element analysis. Clin Oral Implants Res. 2005 Aug;16(4):480-5.

[17] Lee J. Contact Non-linear Finite Element Model Analysis of immediately loaded orthodontic mini implant. Seoul, Korea: Yonsei University; 2005.

[18] Kravitz ND, Kusnoto B. Risks and complications of orthodontic miniscrews. Am J Orthod Dentofacial Orthop. 2007 Apr;131(4 Suppl):S43-51.

[19] Esposito M, Hirsch JM, Lekholm U, Thomsen P. Biological factors contributing to failures of osseointegrated oral implants. (II). Etiopathogenesis. European journal of oral sciences. 1998 Jun;106(3):721-64.

[20] Esposito M, Hirsch JM, Lekholm U, Thomsen P. Biological factors contributing to failures of osseointegrated oral implants. (I). Success criteria and epidemiology. European journal of oral sciences. 1998 Feb;106(1):527-51.

[21] Egolf RJ, BeGole EA, Upshaw HS. Factors associated with orthodontic patient compliance with intraoral elastic and headgear wear. Am J Orthod Dentofacial Orthop. 1990 Apr;97(4):336-48.

[22] Ohmae M, Saito S, Morohashi T, Seki K, Qu H, Kanomi R, et al. A clinical and histological evaluation of titanium mini-implants as anchors for orthodontic intrusion in the beagle dog. Am J Orthod Dentofacial Orthop. 2001 May;119(5):489-97.

[23] Kanomi R. Mini-implant for orthodontic anchorage. J Clin Orthod. 1997 Nov;31(11):763-7.

[24] Costa A, Raffainl M, Melsen B. Miniscrews as orthodontic anchorage: a preliminary report. Int J Adult Orthodon Orthognath Surg. 1998;13(3):201-9.

[25] Justens E, De Bruyn H. Clinical outcome of mini-screws used as orthodontic anchorage. Clin Implant Dent Relat Res. 2008 Sep;10(3):174-80.

[26] Song YY, Cha JY, Hwang CJ. Mechanical characteristics of various orthodontic mini-screws in relation to artificial cortical bone thickness. Angle Orthod. 2007 Nov;77(6):979-85.

[27] Park YC, Lee SY, Kim DH, Jee SH. Intrusion of posterior teeth using mini-screw implants. Am J Orthod Dentofacial Orthop. 2003 Jun;123(6):690-4.

[28] Bae SM, Park HS, Kyung HM, Kwon OW, Sung JH. Clinical application of micro-implant anchorage. J Clin Orthod. 2002 May;36(5):298-302.

[29] Park HS, Bae SM, Kyung HM, Sung JH. Micro-implant anchorage for treatment of skeletal Class I bialveolar protrusion. J Clin Orthod. 2001 Jul;35(7):417-22.

[30] Park HS, Kwon TG, Sung JH. Nonextraction treatment with microscrew implants. Angle Orthod. 2004 Aug;74(4):539-49.

[31] Chang YJ, Lee HS, Chun YS. Microscrew anchorage for molar intrusion. J Clin Orthod. 2004 Jun;38(6):325-30; quiz 33.

[32] Cope BJ. Temporary anchorage devices in orthodontics: A paradigm shift. Seminars in orthodontics. 2005;11(1):3-9.

[33] Melsen B, Costa A. Immediate loading of implants used for orthodontic anchorage. Clin Orthod Res. 2000 Feb;3(1):23-8.

[34] Gainforth BL, Higley LB. A study of orthodontic anchorage possibilities in basal bone. Am J Orthod Oral Surg. 1945;31:406-17.

[35] Linkow LI. The endosseous blade implant and its use in orthodontics. Int J Orthod. 1969 Dec;7(4):149-54.

[36] Branemark PI, Breine U, Adell R, Hansson BO, Lindstrom J, Ohlsson A. Experimental studies on intra-osseous anchorage of dental prosthesis. Arsb Goteb Tandlak Sallsk. 1970:9-25.

[37] Roberts WE, Smith RK, Zilberman Y, Mozsary PG, Smith RS. Osseous adaptation to continuous loading of rigid endosseous implants. Am J Orthod. 1984 Aug;86(2):95-111.

[38] Block MS, Hoffman DR. A new device for absolute anchorage for orthodontics. Am J Orthod Dentofacial Orthop. 1995 Mar;107(3):251-8.

[39] Umemori M, Sugawara J, Mitani H, Nagasaka H, Kawamura H. Skeletal anchorage system for open-bite correction. Am J Orthod Dentofacial Orthop. 1999 Feb;115(2):166-74.

[40] Sugawara J. Dr. Junji Sugawara on the skeletal anchorage system. Interview by Dr. Larry W. White. J Clin Orthod. 1999 Dec;33(12):689-96.

[41] Chung KR, Kim YS, Linton JL, Lee YJ. The miniplate with tube for skeletal anchorage. J Clin Orthod. 2002 Jul;36(7):407-12.

[42] Anka G. Use of miniscrews as temporary anchorage devices in orthodontic practice. I--Introduction. Aust Orthod J. 2006 Nov;22(2):131-9.

[43] Lin JC, Liou EJ. A new bone screw for orthodontic anchorage. J Clin Orthod. 2003 Dec;37(12):676-81.

[44] Huang LH, Shotwell JL, Wang HL. Dental implants for orthodontic anchorage. Am J Orthod Dentofacial Orthop. 2005 Jun;127(6):713-22.

[45] Manopatanakul S. Titanium mini-implant for orhodontic anchorage. J Tha A O. 2004;3:41-6.

[46] ASTM. Standard specification for wrought titanium-6 aluminium-4 ELI (Extra low interstitial) Alloy (R56401) for surgical implant applications, Designation: F 136-96. American Society of Testing and Material. 1996:1-3.

[47] Park JB, Lakes RS. An introduction. Biomaterials. 2 ed. New York: Plenum 1992.

[48] McCracken M. Dental implant materials: commercially pure titanium and titanium alloys. J Prosthodont. 1999 Mar;8(1):40-3.

[49] Glatzmaier J, Wehrbein H, Diedrich P. Biodegradable implants for orthodontic anchorage. A preliminary biomechanical study. Eur J Orthod. 1996 Oct;18(5):465-9.

[50] Tansalarak R. Comparison of the mechanical properties of othodontic mini-screws in commercial use. Bangkok, Thailand: Mahidol University; 2008.

[51] ISO5832-3. Implants for surgery-Metallic materials-Parts 3: Wrought titanium 6-aluminium 4-vanadium alloy. 1996.

[52] Perry CR, Gilula LA. Basic principles and clinical uses of screws and bolts. Orthop Rev. 1992 Jun;21(6):709-13.

[53] Uhl RL. The biomechanics of screws. Orthop Rev. 1989 Dec;18(12):1302-7.

[54] Lee JS, Kim JK, Park Y-C, Vanarsadall RL. Design and Function of New, Screw-Type Orthodontic Mini-Implants. Application of Orthodontic Mini-Implants. Hanover Park: Quintessence Publishing Co 2007:29-50.

[55] Saka B. Mechanical and biomechanical measurements of five currently available osteosynthesis systems of self-tapping screws. Br J Oral Maxillofac Surg. 2000 Feb;38(1):70-5.

[56] Perren SM. Force measurements in screw fixation. J Biomech. 1976;9(11):669-75.

[57] Bahr W. Pretapped and self-tapping screws in the human midface. Torque measurements and bone screw interface. Int J Oral Maxillofac Surg. 1990 Feb;19(1):51-3.

[58] Misch CE, Strong JT, Bidez MW. Specific rationale for dental implant design. Contemporary Implant Dentistry. 3rd ed. St. Louis: Mosby 2008:200-29.

[59] Branemark PI, Hansson BO, Adell R, Breine U, Lindstrom J, Hallen O, et al. Osseointegrated implants in the treatment of the edentulous jaw. Experience from a 10-year period. Scand J Plast Reconstr Surg Suppl. 1977;16:1-132.

[60] Misch CE. Implant design considerations for the posterior regions of the mouth. Implant Dent. 1999;8(4):376-86.

[61] Park HS, Jeong SH, Kwon OW. Factors affecting the clinical success of screw implants used as orthodontic anchorage. Am J Orthod Dentofacial Orthop. 2006 Jul;130(1):18-25.

[62] Brunski JB, Puleo DA, Nanci A. Biomaterials and biomechanics of oral and maxillofacial implants: current status and future developments. Int J Oral Maxillofac Implants. 2000 Jan-Feb;15(1):15-46.

[63] Motoyoshi M, Inaba M, Ono A, Ueno S, Shimizu N. The effect of cortical bone thickness on the stability of orthodontic mini-implants and on the stress distribution in surrounding bone. Int J Oral Maxillofac Surg. 2009 Jan;38(1):13-8.

[64] Moroni A, Vannini F, Mosca M, Giannini S. State of the art review: techniques to avoid pin loosening and infection in external fixation. J Orthop Trauma. 2002 Mar;16(3):189-95.

[65] Gapski R, Wang HL, Mascarenhas P, Lang NP. Critical review of immediate implant loading. Clin Oral Implants Res. 2003 Oct;14(5):515-27.

[66] Simon H, Caputo AA. Removal torque of immediately loaded transitional endosseous implants in human subjects. Int J Oral Maxillofac Implants. 2002 Nov-Dec;17(6):839-45.

[67] Choi BH, Zhu SJ, Kim YH. A clinical evaluation of titanium miniplates as anchors for orthodontic treatment. Am J Orthod Dentofacial Orthop. 2005 Sep;128(3):382-4.

[68] Costa A, Maric M, Danesino P. Comparison between two orthodontic skeletal anchorage devices: osseointegrated implants and miniscrews - Medical-Legal Considerations. Prog Orthod. 2006;7(1):24-31.

[69] Carano A, Lonardo P, Velo S, Incorvati C. Mechanical properties of three different commercially available miniscrews for skeletal anchorage. Prog Orthod. 2005;6(1):82-97.

[70] Kim JW, Ahn SJ, Chang YI. Histomorphometric and mechanical analyses of the drill-free screw as orthodontic anchorage. Am J Orthod Dentofacial Orthop. 2005 Aug;128(2):190-4.

[71] Heidemann W, Terheyden H, Gerlach KL. Analysis of the osseous/metal interface of drill free screws and self-tapping screws. J Craniomaxillofac Surg. 2001 Apr;29(2):69-74.

[72] Eriksson AR, Albrektsson T. Temperature threshold levels for heat-induced bone tissue injury: a vital-microscopic study in the rabbit. J Prosthet Dent. 1983 Jul;50(1):101-7.

[73] Misch CE, Bidez MW, Sharawy M. A bioengineered implant for a predetermined bone cellular response to loading forces. A literature review and case report. J Periodontol. 2001 Sep;72(9):1276-86.

[74] Lekholm U, Zarb G. Patient selection and preparation. In: Branemark PI, Zarb GA, Albrektsson T (eds): Tissue-Integrated Prostheses: Osseointegration in Clinical Dentistry. Chicago: Quintessence 1985.

[75] Misch C. Contemporary Implant Dentistry. 2nd ed. Chicago: Mosby 1999.

[76] Friberg B, Sennerby L, Roos J, Lekholm U. Identification of bone quality in conjunction with insertion of titanium implants. A pilot study in jaw autopsy specimens. Clin Oral Implants Res. 1995 Dec;6(4):213-9.

[77] Adell R, Lekholm U, Rockler B, Branemark PI. A 15-year study of osseointegrated implants in the treatment of the edentulous jaw. Int J Oral Surg. 1981 Dec;10(6):387-416.

[78] Norton MR, Gamble C. Bone classification: an objective scale of bone density using the computerized tomography scan. Clin Oral Implants Res. 2001 Feb;12(1):79-84.

[79] Molly L. Bone density and primary stability in implant therapy. Clin Oral Implants Res. 2006 Oct;17 Suppl 2:124-35.

[80] Deguchi T, Nasu M, Murakami K, Yabuuchi T, Kamioka H, Takano-Yamamoto T. Quantitative evaluation of cortical bone thickness with computed

tomographic scanning for orthodontic implants. Am J Orthod Dentofacial Orthop. 2006 Jun;129(6):721 e7-12.

[81] Ono A, Motoyoshi M, Shimizu N. Cortical bone thickness in the buccal posterior region for orthodontic mini-implants. Int J Oral Maxillofac Surg. 2008 Apr;37(4):334-40.

[82] Beer A, Gahleitner A, Holm A, Tschabitscher M, Homolka P. Correlation of insertion torques with bone mineral density from dental quantitative CT in the mandible. Clin Oral Implants Res. 2003 Oct;14(5):616-20.

[83] Ikumi N, Tsutsumi S. Assessment of correlation between computerized tomography values of the bone and cutting torque values at implant placement: a clinical study. Int J Oral Maxillofac Implants. 2005 Mar-Apr;20(2):253-60.

[84] Lim SA, Cha JY, Hwang CJ. Insertion Torque of Orthodontic Miniscrews According to Changes in Shape, Diameter and Length. Angle Orthod. 2008;78(2):234-40.

[85] Mailath G, Stoiber B, Watzek G, Matejka M. Bone resorption at the entry of osseointegrated implants--a biomechanical phenomenon. Finite element study. Zeitschrift fur Stomatologie (1984). 1989 Jun;86(4):207-16.

[86] You ZH, Bell WH, Schneiderman ED, Ashman RB. Biomechanical properties of small bone screws. J Oral Maxillofac Surg. 1994 Dec;52(12):1293-302.

[87] Srinok C. Association between insertion torque and the success of minii-screws used as orthodontic anchorage. Bangkok, Thailand: Mahidol University; 2008.

[88] Eriksson RA, Albrektsson T. The effect of heat on bone regeneration: an experimental study in the rabbit using the bone growth chamber. J Oral Maxillofac Surg. 1984 Nov;42(11):705-11.

[89] Heidemann W, Gerlach KL, Grobel KH, Kollner HG. Influence of different pilot hole sizes on torque measurements and pullout analysis of osteosynthesis screws. J Craniomaxillofac Surg. 1998 Feb;26(1):50-5.

[90] Heidemann W, Gerlach KL, Grobel KH, Kollner HG. Drill Free Screws: a new form of osteosynthesis screw. J Craniomaxillofac Surg. 1998 Jun;26(3):163-8.

[91] Chen F, Terada K, Hanada K, Saito I. Anchorage effect of osseointegrated vs nonosseointegrated palatal implants. Angle Orthod. 2006 Jul;76(4):660-5.

[92] Gallas MM, Abeleira MT, Fernandez JR, Burguera M. Three-dimensional numerical simulation of dental implants as orthodontic anchorage. Eur J Orthod. 2005 Feb;27(1):12-6.

[93] Freudenthaler JW, Haas R, Bantleon HP. Bicortical titanium screws for critical orthodontic anchorage in the mandible: a preliminary report on clinical applications. Clin Oral Implants Res. 2001 Aug;12(4):358-63.

[94] ASTM. Standard specification and test methods for metallic medical bone screw, Designation: F 543-02. American Society of Testing and Material. 2002:114-32.

[95] ASTM. Standard specification for rigid polyurethane foam for use as a standard material for testing orthopaedic devices and intruments, Designation:F 1839-01. American Society of Testing and Material. 2007:861-6.

[96] Vasquez M, Calao E, Becerra F, Ossa J, Enriquez C, Fresneda E. Initial stress differences between sliding and sectional mechanics with an endosseous implant as anchorage: a 3-dimensional finite element analysis. Angle Orthod. 2001 Aug;71(4):247-56.

[97] Misch CE, Qu Z, Bidez MW. Mechanical properties of trabecular bone in the human mandible: implications for dental implant treatment planning and surgical placement. J Oral Maxillofac Surg. 1999 Jun;57(6):700-6; discussion 6-8.

[98] Dieter GE. Mechanical metallurgy. Singapore: McGraw-Hill, Inc. 1988.

[99] Geng JP, Tan KB, Liu GR. Application of finite element analysis in implant dentistry: a review of the literature. J Prosthet Dent. 2001 Jun;85(6):585-98.

[100] Wilmes B, Rademacher C, Olthoff G, Drescher D. Parameters affecting primary stability of orthodontic mini-implants. J Orofac Orthop. 2006 May;67(3):162-74.

[101] Laird II G, Epstein JS. Fracture mechanics and finite element analysis. Mechanical engineering. 1992:69-73.

[102] Kulkarni DM, Prakash R, Kumar AN. Experimental and finite element analysis of fracture criterion in general yeilding fracture mechanics. Sadhana. 2002;27:631-42.

[103] Torut S. Influence of orthodontic miniscrew implant size and loading forces on stress distribution: finite element method. Chiang Mai, Thailand: Chiang Mai University; 2008.

[104] Frost HM. A 2003 update of bone physiology and Wolff's Law for clinicians. Angle Orthod. 2004 Feb;74(1):3-15.

[105] Melsen B, Lang NP. Biological reactions of alveolar bone to orthodontic loading of oral implants. Clin Oral Implants Res. 2001 Apr;12(2):144-52.

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