CHAPTER 5
INFLUENCE OF ORTHODONTIC MINISCREW IMPLANT SIZES AND LOADING FORCES ON STRESS DISTRIBUTION: FINITE ELEMENT METHOD

5.1 Introduction

Recently, a wide variety of miniscrew implants with several sizes and designs have been developed for clinical use (Carano *et al.*, 2005; Chung *et al.*, 2004; Kyung *et al.*, 2003; Lin and Liou, 2003; Maino *et al.*, 2003). According to our previous systematic review in Chapter 2, the diameter of miniscrew implants varied from 1.0 to 2.3 mm and their lengths varied from 4.0 mm to 17.0 mm.

The diameter of the miniscrew implant has been reported to be one of the most important factors related to failure rate (Chen *et al.*, 2006a; Cheng *et al.*, 2004; Fritz *et al.*, 2004; Miyawaki *et al.*, 2003; Park *et al.*, 2006). The use of miniscrew implants of less than 1 mm in diameter did not sustain orthodontic force of 200 g and they loosened before the end of treatment, resulting in a higher failure rate than when larger sizes were used (Miyawaki *et al.*, 2003). The higher failure rate may relate to changes in the geometry of the miniscrew implant that may influence the biomechanical properties, especially load transmission, of both the miniscrew implant and the surrounding bone (Himmlova *et al.*, 2004; Holmgren *et al.*, 1998).

In a retrospective study (Ekfeldt *et al.*, 2001), designed to verify implant failures in the maxilla, it has been demonstrated that where loading problems were present, the failure rate was three times higher than it was in situations with better loading conditions. There are many methods to evaluate the loading condition, such as; the strain gauge method (Hekimoglu *et al.*, 2004), the force and bending moment method (Rangert *et al.*, 1995) and photoelastic analysis (Ochiai *et al.*, 2003). However, detailed information of the stress distribution can not be obtained and analyzed by these experimental techniques, on account of the interaction with surrounding tissue (McGuinness *et al.*, 1992) and because of the complexity of the components in a miniscrew implant-bone system. The finite element method (FEM) is an appropriate

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technique for solving the complex mechanical problem by dividing the problem domain into smaller and simpler domains (elements) (Geng *et al.*, 2001).

FEM was applied, at least three decades ago, to evaluate the influence of several parameters, such as implant design (Cook *et al.*, 1982), implant size (Himmlova *et al.*, 2004; Iplikcioglu and Akca, 2002), bone quality (Tada *et al.*, 2003), bone quality (Tada *et al.*, 2003) and number of abutments (Rangert *et al.*, 1995), on the stress/strain concentration in dental implants and surrounding bone. However, this method was first applied (Motoyoshi *et al.* in 2005), to the assessment of the influence of the abutment anchor on the miniscrew.

Buchter *et al.*, (2005) conducted a study of the influence of orthodontic load on the stability of miniscrew implants and they found that the applied load of 300 g at the 3 mm neck/bone distance level resulted in a moment of 900 cNmm and in a decrease of stability of the miniscrew implant. Moreover, Melsen and Lang (2001), concluded that excessive stress leads to destruction of surrounding tissue and decreases the bone contact area between implant and surrounding bone. However, the magnitudes of orthodontic loading force applied to miniscrew implants vary depending on the biomechanics of tooth movement. Furthermore, the effects of these loading conditions are not conclusive.

The evaluation of the effects of these parameters, size of miniscrew implant and loading condition, on the biomechanical performance of both the miniscrew implant and the surrounding bone is important. Therefore, the purpose of this study was to evaluate the influence of miniscrew implant size and orthodontic loading force on stress distribution in miniscrew implant and in surrounding bone.

5.2 Materials and methods

The finite element models were established and verified by a mathematical method at the Finite Element Method Laboratory, Department of Mechanical Engineering, Faculty of Engineering, Chiang Mai University, Thailand. Subsequently, maximum stress distribution in miniscrew implants and bone was analyzed and described. The processes of the finite element method are described in the following sequences.

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5.2.1 Preprocessing

a. Geometry of model

Twenty-five miniscrew implant models of various sizes were crated by using computer software (SolidWorks 2004, SolidWorks Corporation, U.S.). The miniscrew implant models were tapered in shape and composed of a single right-hand-thread screw, with a 0.5° taper thread, a 0.7 mm pitch distance and a 0.35 mm depth of thread (Figure 5.1, Table 5.1).

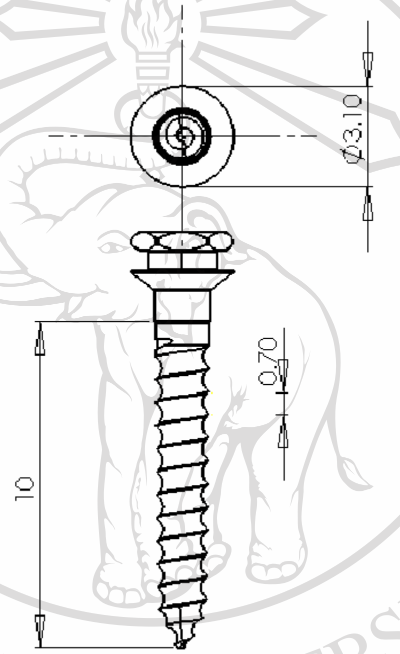

Figure 5.1 Diagram of miniscrew model design (1.6 mm body diameter, 10 mm length, 3.10 mm head diameter)

Human bone models with cortical thickness of 1.5mm and cancellous bone thickness of 15.0 mm were constructed. The overall dimensions of the bone block models were 16.5 mm in height, 8.0 mm in mesio-distal length, and 8.0 mm in bucco-lingual width. Each miniscrew implant model was inserted into a bone model, until its platform was 1.0 mm (soft tissue thickness) from the virtual cortical bone surface (Figure 5.2). This combination of virtual models is referred to as an assembly. Miniscrew implant and surrounding bone were modeled in a “no penetration” situation.

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Table 5.1 Details of sizes and numbers of miniscrew implant models

Number of model	Diameter (mm)	Length (mm)
1	1.0	4.0
2	1.0	6.0
3	1.0	8.0
4	1.0	10.0
5	1.0	12.0
6	1.2	4.0
7	1.2	6.0
8	1.2	8.0
9	1.2	10.0
10	1.2	12.0
11	1.4	4.0
12	1.4	6.0
13	1.4	8.0
14	1.4	10.0
15	1.4	12.0
16	1.6	4.0
17	1.6	6.0
18	1.6	8.0
19	1.6	10.0
20	1.6	12.0
21	1.8	4.0
22	1.8	6.0
23	1.8	8.0
24	1.8	10.0
25	1.8	12.0

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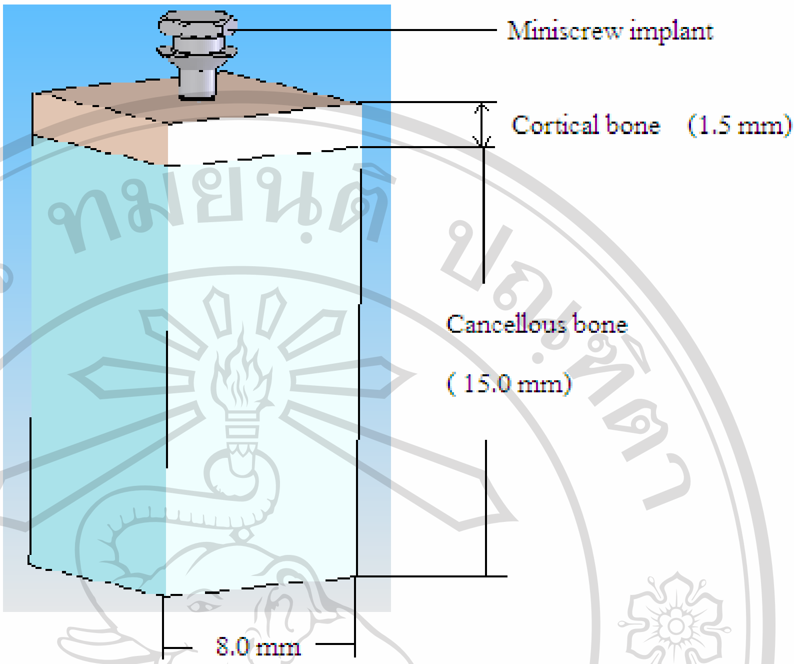

Figure 5.2 Detail of miniscrew implant and bone model dimensions, (cortical bone thickness 1.5 mm, cancellous bone thickness 15.0 mm, width of bone block 8.0 mm)

b. Properties of materials

The miniscrew implant and bone models were of isotropic, homogeneous and linearly elastic materials. The properties of the simulated materials were based on those of a previous study (Table 5.2). (Iplikcioglu and Akca, 2002)

Table 5.2 Details of properties of materials used in this study

Material	E* (GPa)	Poisson's ratio	Yield stress (MPa)	Ultimate stress (MPa)	
				Tension	Compression
Cortical bone	13.7	0.3	-	100	173
Cancellous bone	1.85	0.3	-	-	-
Titanium	110	0.35	550	-	-

*Young's modulus of elasticity

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c. Loading and constrained conditions

Loading and constrained conditions were simulated by using FEA software (COSMOSWorks 2004, SolidWorks Corporation, U.S.).

To evaluate the influence of miniscrew implant size, simulated loading forces of 50 g were applied at 90 deg. (horizontal direction) to the long axis of the miniscrew implant for all models.

To evaluate the influence of orthodontic loading forces, simulated loading forces of 50, 100, 150, 200, 250, 300, 350 and 400 g were applied for all models.

The lower part of the bone was constrained (Figure 5.2).

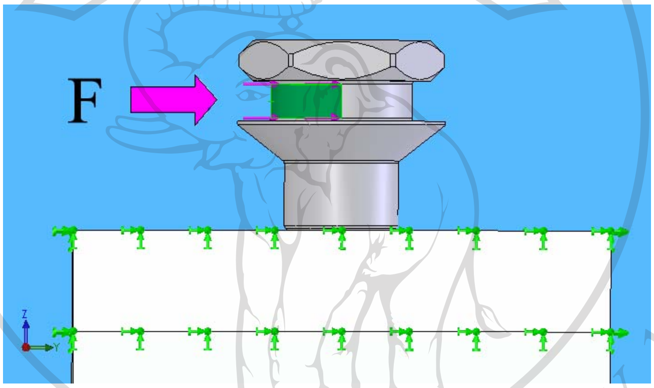

Figure 5.2 Direction and position of force applied parallel to the Y axis.

5.2.2 Processing

The models were divided into elements by means of the ten-node tetrahedral method. The *octahedral shear stress yield criterion theory* was used to calculate the maximum stresses in miniscrews and the *Coulomb-Mohr fracture criterion theory* was used to calculate the same stresses in cortical and cancellous bone, based on the recommendations of Dowling (1999).

The *octahedral shear stress yield criterion theory* is the yield criterion 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

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stresses in miniscrew implants 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*.

According to 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 can not 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. In this study, the maximum stress evaluated in bone was called *maximum first principal stress*, the highest stress at a given location.

Stress values in miniscrew implant and surrounding bone models were calculated by using FEA software (COSMOSWorks 2004, SolidWorks corporation, U.S.).

5.2.3 Post-processing

Stress distribution patterns in the miniscrew implant and surrounding bone models were described and illustrated in color scheme diagrams, in which bands of color represent the different levels of stress concentration. Maximum von Mises stresses in miniscrew implant models and maximum first principal stresses of surrounding bone models were collected and analyzed.

5.3 Results

5.3.1 Influence of miniscrew implant size

Stress distribution pattern

Stress concentration in miniscrew implant models was mainly on both sides of the cervical part of the miniscrew implant, in the area between the first and third threads. However, in miniscrew implant models with diameters of 1.0 mm and 1.2 mm, the stresses were higher and presented in a lager area than in models of other diameters (Figure 5.4 A-E).

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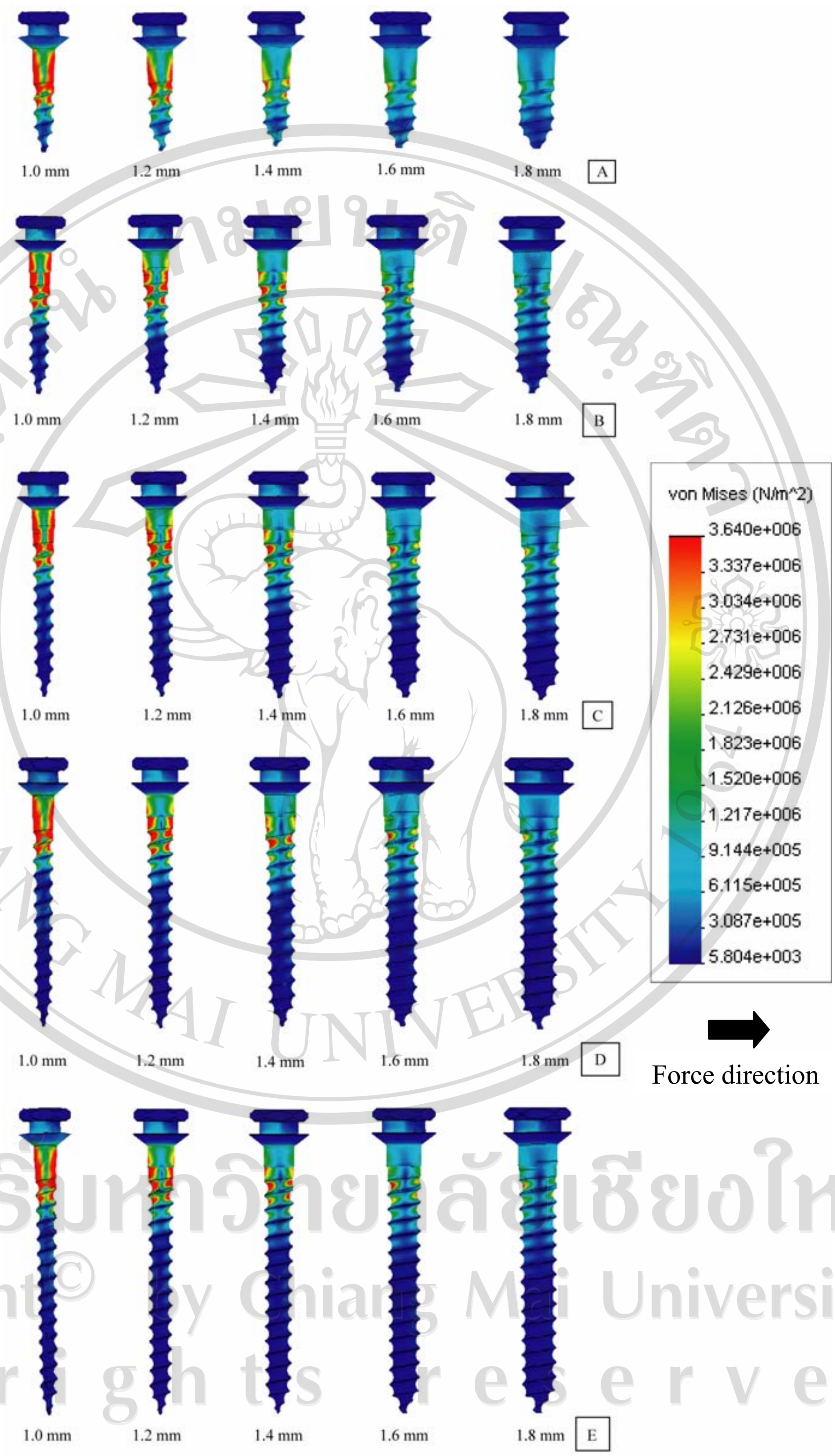

Figure 5.4 Patterns of stress distribution in miniscrew implant models with lengths of 4.0 mm (A), 6.0 mm (B), 8.0 mm (C), 10.0 mm (D), 12.0 mm (E).

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Stress concentration in cortical bone models presented in a different way. In the upper part of the cortical bone model, stress concentration was mainly on the same side as the force vector, whereas, in the lower part, stress concentration was mainly on the side opposite the force vector. The stress distribution pattern in cancellous bone was not obvious (Figure 5.5 A-E).

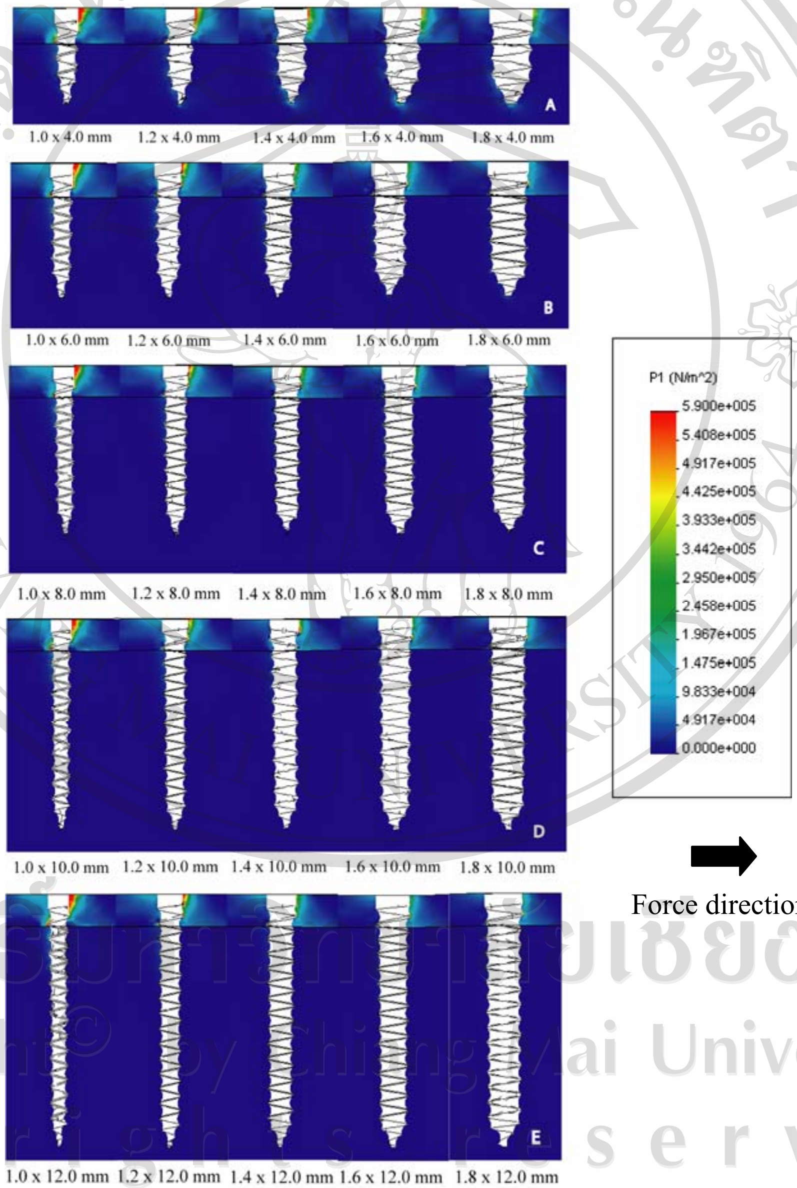

Figure 5.5 Patterns of stress distribution in cortical and cancellous bone assembly with miniscrew models with lengths of 4.0 mm (A), 6.0 mm (B), 8.0 mm (C), 10.0 mm (D), 12.0 mm (E).

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Stress value

The details of stress value in miniscrew implant and surrounding bone models are shown in Table 5.3.

Maximum von Mises stresses in miniscrew implant models decreased with increasing implant diameter (Figure 5.6A). A similar trend was shown for maximum first principal stresses in cortical bone models (Figure 5.6B). However, maximum first principal stresses in cancellous bone models presented in a different way. Models with miniscrew diameters of more than 1.2 mm and lengths of 4.0 and 6.0 mm showed slightly increased stress values in the cancellous bone models, whereas with all diameters, models with miniscrew lengths of 8.0 to 12.0 mm showed a slight decrease in stress values with increasing diameter (Figure 5.6C).

Miniscrew implant models with diameters of 1.2 to 1.8 mm presented slightly increased maximum von Mises stress values. However, stress values in miniscrew models with diameters of 1.0 and 1.2 mm were significantly higher than those in models with diameters of 1.4 to 1.8 mm (Figure 5.7A).

Maximum first principal stress values in cortical bone models containing miniscrew implants with diameters of 1.2 to 1.8 mm presented slightly increased stress values with increasing miniscrew implant length. However, stress values in cortical bone models containing miniscrews with diameters of 1.6 and 1.8 mm were significantly lower than in those containing miniscrews with diameters of from 1.0 to 1.4 mm (Figure 5.7B).

Maximum first principal stress values in cancellous bone models decreased slightly with increasing miniscrew implant length (Figure 5.7C).

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Table 5.3 Stress values in different sizes of miniscrew implant models with 50 g loading force

Diameter (mm)	Length (mm)	Maximum von Mises stress (MPa)	Maximum first principal stress (MPa)	
			Cortical bone	Cancellous bone
1.0	4.0	12.10	2.65	0.18
	6.0	15.16*	2.18*	0.19
	8.0	14.46*	2.18*	0.16
	10.0	11.88	2.69	0.16
	12.0	11.50	2.85	0.18
1.2	4.0	9.38	1.55	0.21
	6.0	10.06	1.65	0.22
	8.0	10.98	1.52	0.16
	10.0	11.02	1.93	0.25
	12.0	11.13	1.81	0.21
1.4	4.0	5.16	1.03	0.45
	6.0	7.40	1.35	0.32
	8.0	7.51	1.31	0.15
	10.0	6.59	1.40	0.19
	12.0	7.28	1.29	0.12
1.6	4.0	4.31	0.74	0.41
	6.0	5.09	0.77	0.19
	8.0	5.12	0.79	0.10
	10.0	5.26	0.98	0.13
	12.0	4.94	0.80	0.11
1.8	4.0	2.99	0.63	0.33
	6.0	3.55	0.75	0.12
	8.0	3.66	0.68	0.12
	10.0	3.80	1.00	0.12
	12.0	3.64	0.59	0.07

*Error stress values

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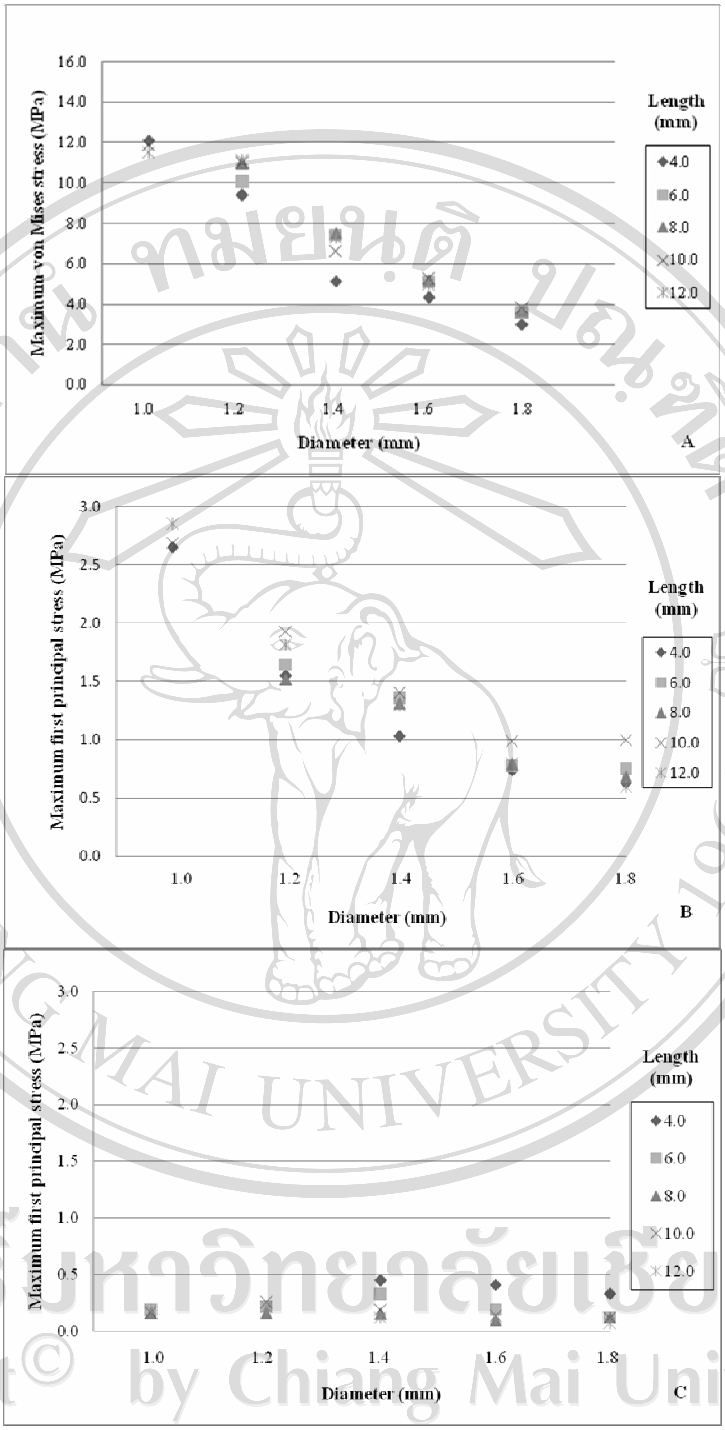

Figure 5.6 Maximum von Mises stresses in miniscrew implant models (A), maximum first principal stresses in (B), cortical, and (C) cancellous bone models in relation to miniscrew implant diameter.

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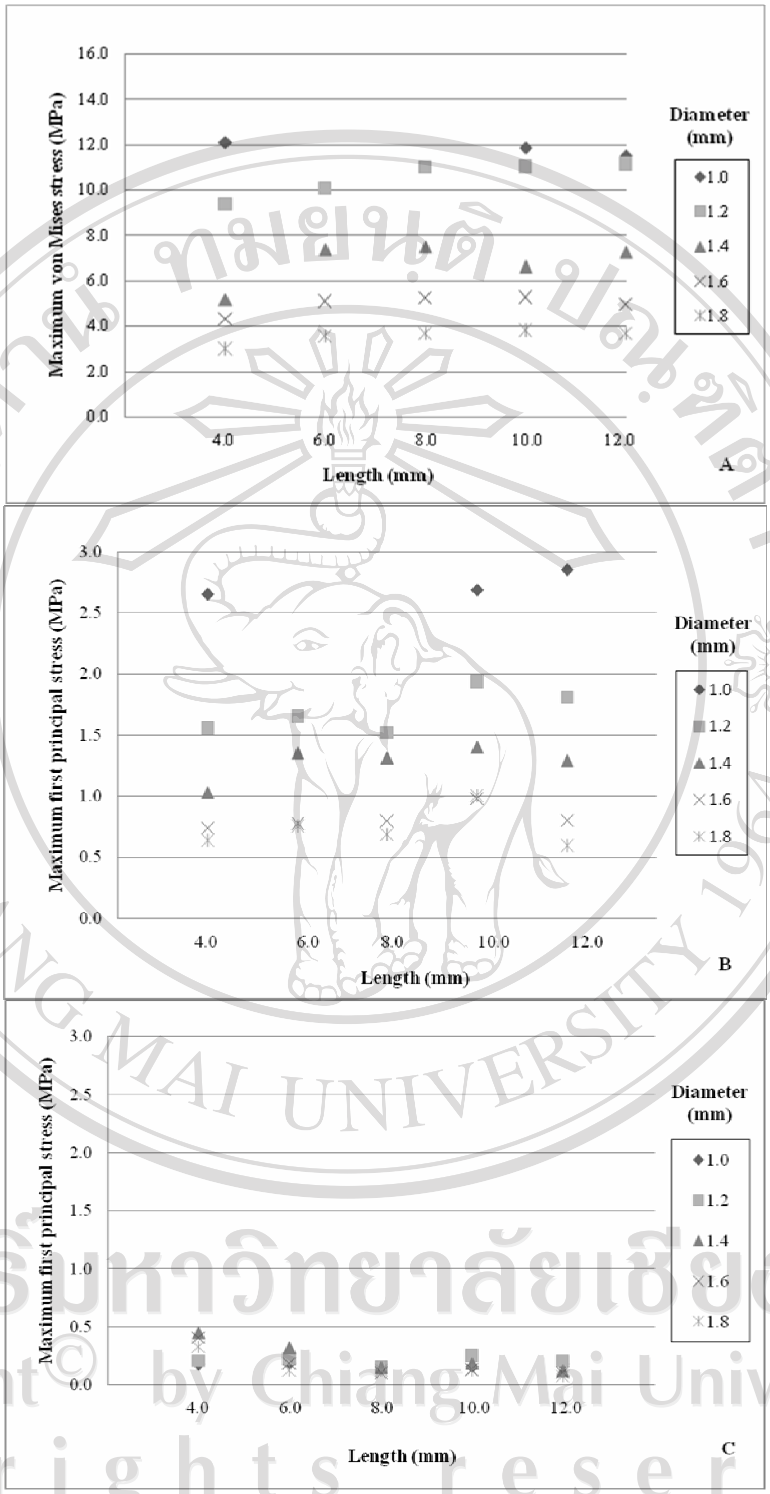

Figure 5.7 Maximum von Mises stresses in miniscrew implant models (A), maximum first principal stresses in (B), cortical, and (C) cancellous bone models in relation to miniscrew implant length.

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5.3.2 Influence of loading

Maximum von Mises stress values in all miniscrew models increased when the magnitude of loading force increased. Interestingly, the stress values in miniscrew implant models with diameters of 1.0 and 1.2 mm were remarkably higher than the values in models with a diameter of 1.4 mm. On the other hand, implant models with diameters of 1.6 and 1.8 mm presented the lowest stress values for all lengths (Figure 5.8 A-E).

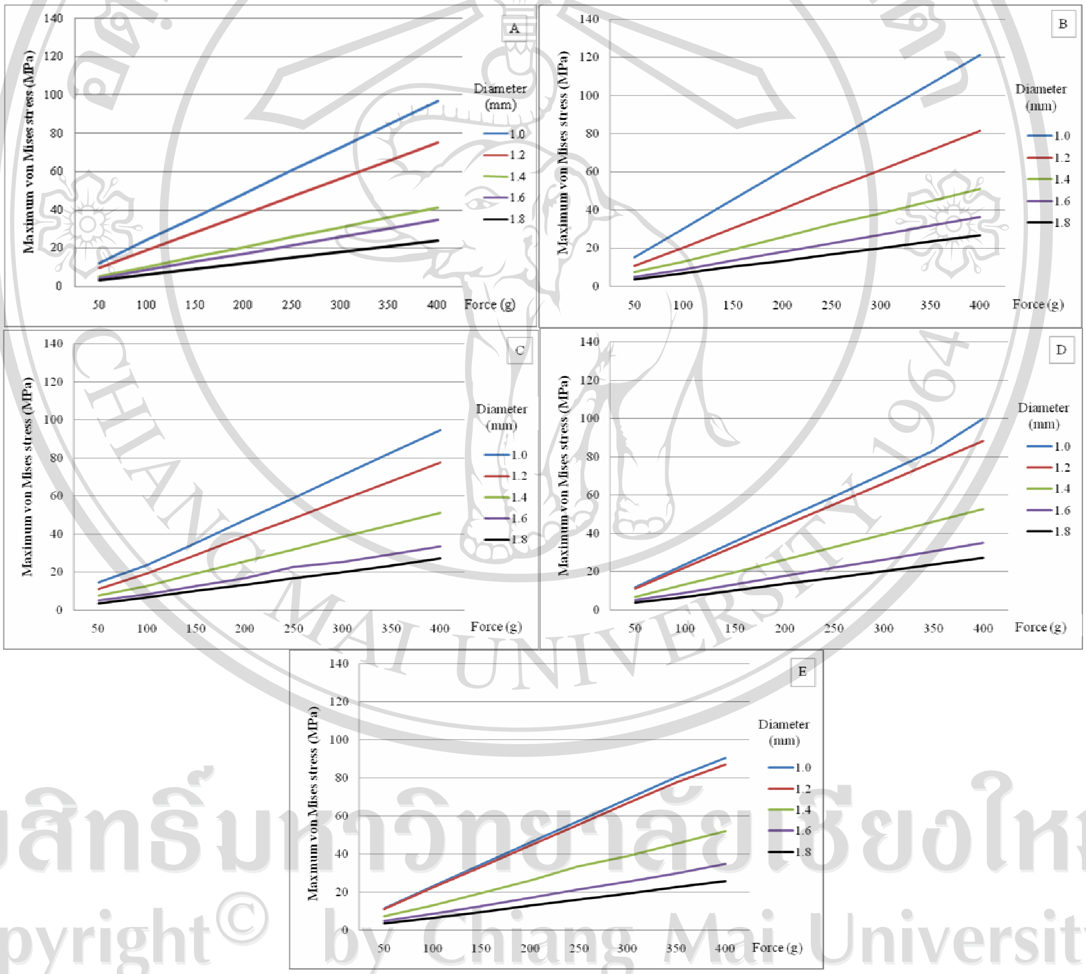

Figure 5.8 Maximum von Mises stresses in miniscrew implant models with lengths of 4.0 mm (A), 6.0 mm (B), 8.0 mm (C), 10.0 mm (D), 12.0 mm (E) shown in relation to magnitude of loading force.

Maximum first principal stress values in cortical bone models also increased with increasing magnitude of loading force. The stress values in cortical bone models

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containing miniscrews with diameters of 1.0 and 1.2 mm were significantly higher than the values in models containing miniscrew implants with diameters of 1.4 to 1.8 mm, especially with a diameter of 1.0 mm. On the other hand, stress values in cortical bone models were lowest in models containing miniscrews with diameters of 1.6 and 1.8 mm (Figure 5.9A-E).

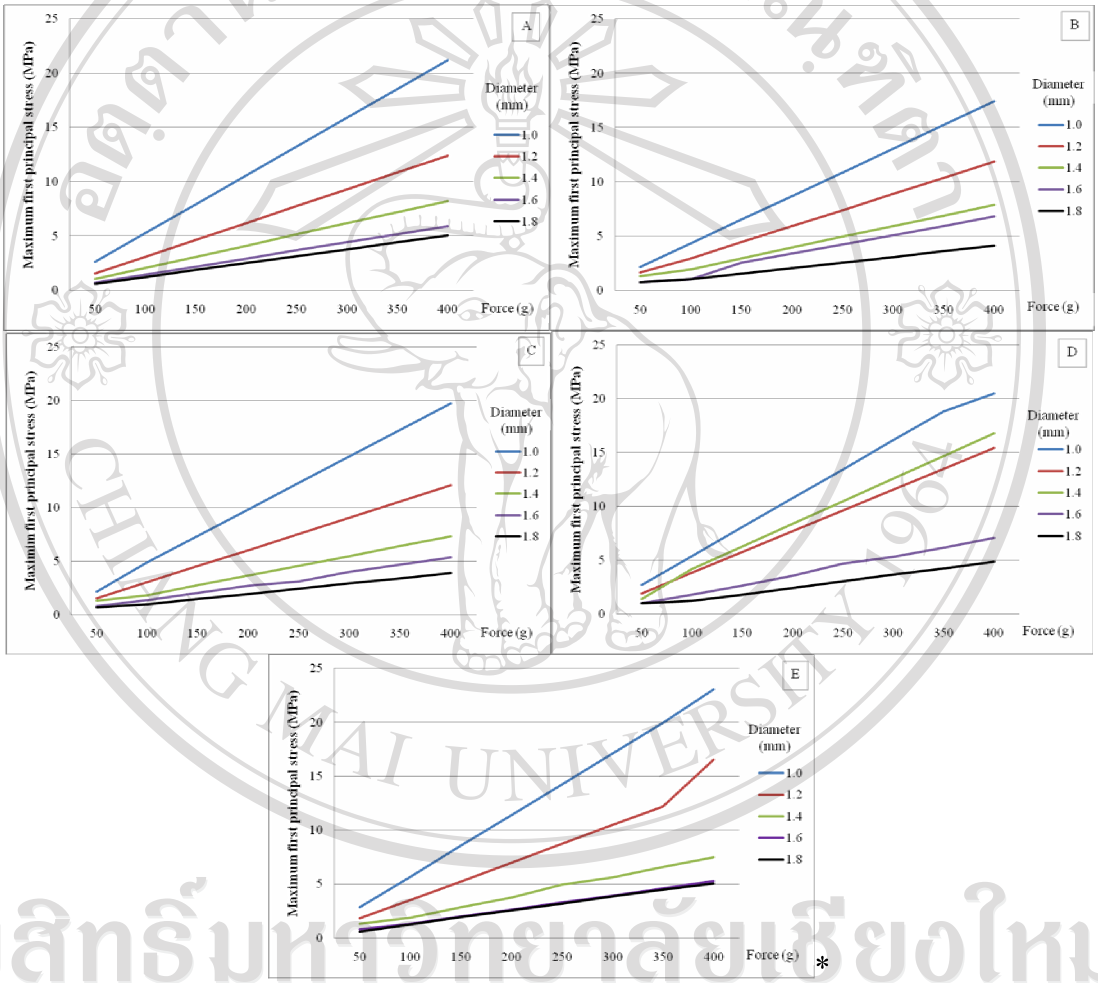

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Figure 5.9 Maximum first principal stresses in cortical bone models containing miniscrew implants with lengths of 4.0 mm (A), 6.0 mm (B), 8.0 mm (C), 10.0 mm (D), 12.0 mm (E) shown in relation to magnitude of loading force.

*Error values in 1.4 mm x 10.0 mm model

Maximum first principal stress values in cancellous bone models increased slightly with increasing magnitude of loading force in all models. The stress values for various diameters of miniscrew implant models were not significantly different for all lengths (Figure 5.10A-E).

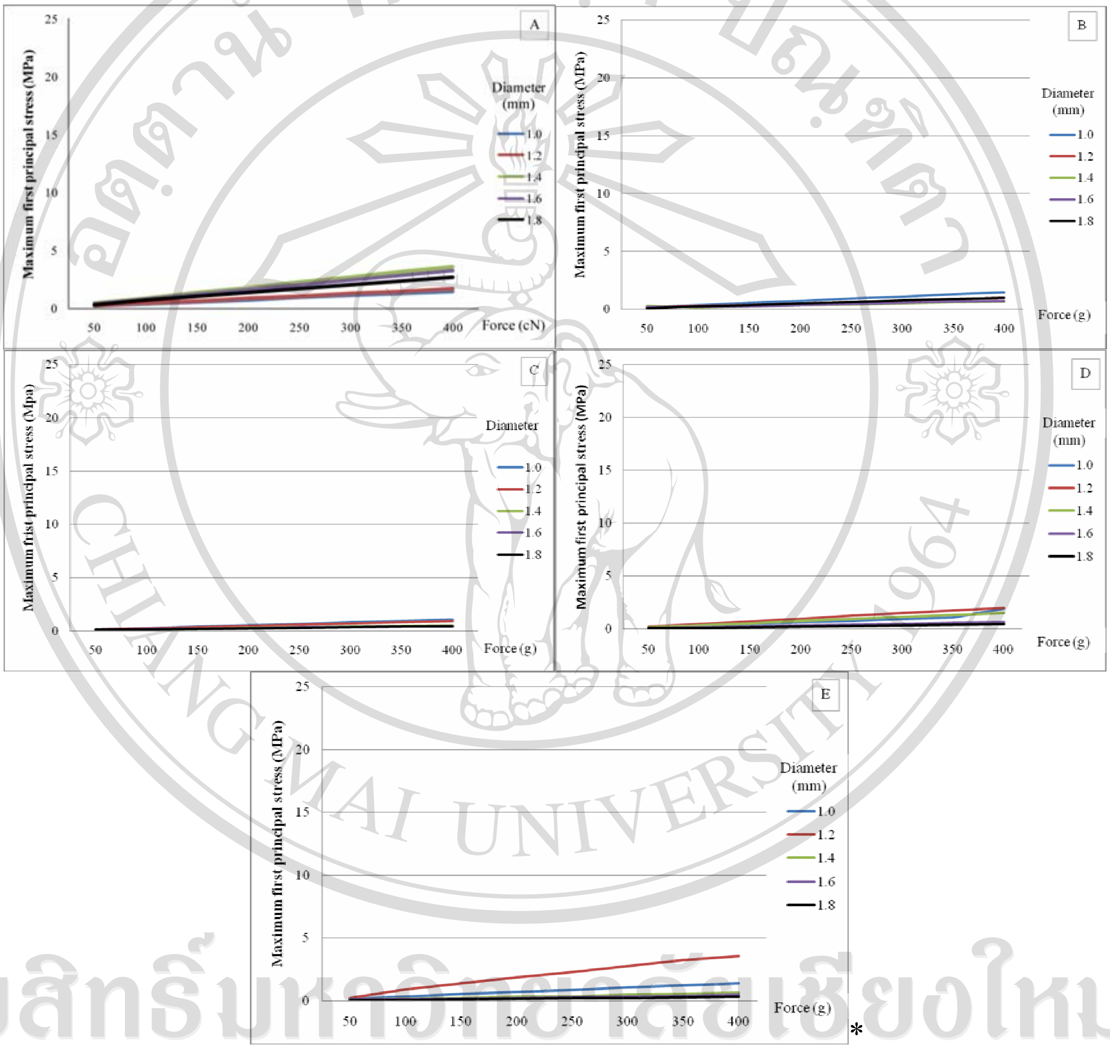

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Figure 5.10 Maximum first principal stresses in cancellous bone models containing miniscrew implants with lengths of 4.0 mm (A), 6.0 mm (B), 8.0 mm (C), 10.0 mm (D), 12.0 mm (E) shown in relation to magnitude of loading force.

*Error values in 1.2 mm x 12.0 mm model

5.4 Discussion

According to previous knowledge about conventional skeletal anchorage, such as dental implants, inappropriate stress distribution in dental implants and bone is an important factor leading to bone resorption and implant failure (Melsen and Lang, 2001). Because bone has mechanical properties that differ from those of dental implants, such as lower Young's modulus of elasticity and tensile stress, if excessive loads are applied, the bone surface fails first, rather than the implant surface. The distribution of stress in the dental implant itself can cause fatigue and failure inside the material (Skalak, 1983). Moreover, Sanches-Garces *et al.* (2004) concluded that excessive mechanical stress and microbial infection lead to peri-implantitits followed by implant failure.

However, mini-screw implants differ from dental implants in various aspects. Firstly, mini-screw implants are attached to the bone mainly by mechanical locking, whereas the main retention of dental implants is obtained from the osseointegration (Cope, 2005). Secondly, mini-screw implants have reduced diameter compared to dental implants (Ohashi *et al.*, 2006). Thirdly, the magnitude of orthodontic loading forces applied to mini-screw implants is much lower than the occlusal forces applied to dental implants. Fourthly, only one orthodontic force direction is applied to miniscrew implants, whereas there are at least two directions of force, vertical and oblique, applied to dental implants (Bozkaya *et al.*, 2004; Ohashi *et al.*, 2006; Ren *et al.*, 2003). Finally, the nature of force rhythm applied to miniscrew implants and dental implants is different. Most of the force applied to miniscrew implants is light, continuous force, whereas force applied to dental implants is heavy, discontinuous force from chewing action (Ohashi *et al.*, 2006). Because of these differences in clinical applications and in force applications, the biomechanical properties of surrounding bone that are influenced by dental implant parameters can not be applied to miniscrew implants.

The finite element method (FEM) is a powerful tool for simulating and predicting the mechanical behavior of biological tissues. The benefit of this technique is that it can obtain a detailed representation of many different factors that affect the biomechanical behavior of bone: mechanical properties, shape, loading configuration, and boundary conditions. However, in order to control the validity of numerical

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techniques, numerical results should be verified (Natali, 2003). In this study, all finite element models were verified with a mathematical method as described in a previous study (Seehawong, 2006). A limitation of this study was the unfavorable results in some calculations because of characteristics of the software. Even though there were some result errors from the COSMOSWorks program used in this study, the majority of results were appropriate.

5.4.1 Influence of miniscrew implant sizes

Stress concentrations in all miniscrew implant models were mainly located in the area below the platform of the miniscrew implant, between the first and third threads, on both sides of the miniscrew implant. A probable reason for the result is that this area was the first part of the implant where the diameter was reduced. According to previous studies, the stress concentrations in dental implant models where horizontal force was applied, were also around the implant neck at the first thread (Cattaneo *et al.*, 2007; Clelland *et al.*, 1991; Gallas *et al.*, 2005).

The stress concentration areas shown in the present study agree with those in several studies that reported miniscrews used for orthodontic anchorage breaking at the cervical portion during screw removal (Jeon *et al.*, 2006; Park *et al.*, 2006; Wilmes *et al.*, 2006). Consequently, fractures of miniscrew implants can be expected to occur in the cervical portion, between the head and body. This area is a weak point of the body.

However, the stress concentration areas narrowed with increasing miniscrew implant diameter, whereas increasing in miniscrew implant length did not show remarkable change in stress distribution pattern. Therefore, increasing the diameter of miniscrew implants improves the pattern of stress distribution in the miniscrew implant.

In general, stress distribution patterns showed that stress was mainly concentrated in cortical bone more than in cancellous bone, especially in the upper part of the cortical bone. This result refers to the nature of the force being transferred from miniscrew implant to the first contact surface. In addition, stress was concentrated in the same direction as the force vector, and acted as a cantilever beam (Rees, 1997).

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Increasing miniscrew implant diameter resulted in decreasing maximum von Mises stress values. Additionally, stress values in miniscrew implants were highest, followed by those in cortical and cancellous bone models, respectively. Cortical bone presented stress values approximately ten times higher than those in cancellous bone. Therefore, the results of this study suggest that cortical bone is the most important critical part of the bone which against which to apply force and consequently, the most important for mechanical retention of miniscrew implants.

Increasing the diameter of miniscrew implant models resulted in decraesed stress values in cortical and cancellous bone. A possible explanation for this result is that the wider miniscrews increase the surface contact area. Similar results have been reported by İplikçioğlu and Akça (2002), Yacoub *et al.* (2002) and Himmlová *et al.* (2004). However, these were dental implant studies with implant diameters of 2.9 - 6.5 mm. In contrast, Holmegren *et al.* (1998) found that increasing the diameter of dental implants did not influence stress value. However, their simulations used two dimensional models that provided less accuracy than can three dimensions model (Himmlova *et al.*, 2004).

From a biomechanical perspective, the optimum choice was a miniscrew implant with the maximum possible diameter allowed by the anatomy. The limitation of miniscrew implant diameter size can be explained mostly by the anatomical limitations encountered in the dentoalveolar bone (Costa *et al.*, 2005; Deguchi *et al.*, 2006; Ishii *et al.*, 2004; Poggio *et al.*, 2006). Since the preferred site for miniscrew implant placement is in the dentoalveolar bone, the position of the roots of adjacent teeth and, consequently, the amount of interradicular bone, plays an important role in the selection of the appropriate diameter to be used.

The results of this study show that changes in miniscrew implant length from 4 to 12 mm slightly increase stress values in miniscrew implants and cortical bone, but slightly decrease stress values in cancellous bone. A possible explanation for these results is that longer miniscrew implants obtained more surface contact and effected a lower degree of rotation (ϴ) than did shorter miniscrew implants (Figure 5.11 and Table 5.4). That means that the miniscrew is less flexible as length increase. This reduced flexibility probably causes increasing bending stress in the miniscrew

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implant and in the upper part of the cortical bone. But it probably slightly decreases the stress in the cancellous bone because of increasing contact area.

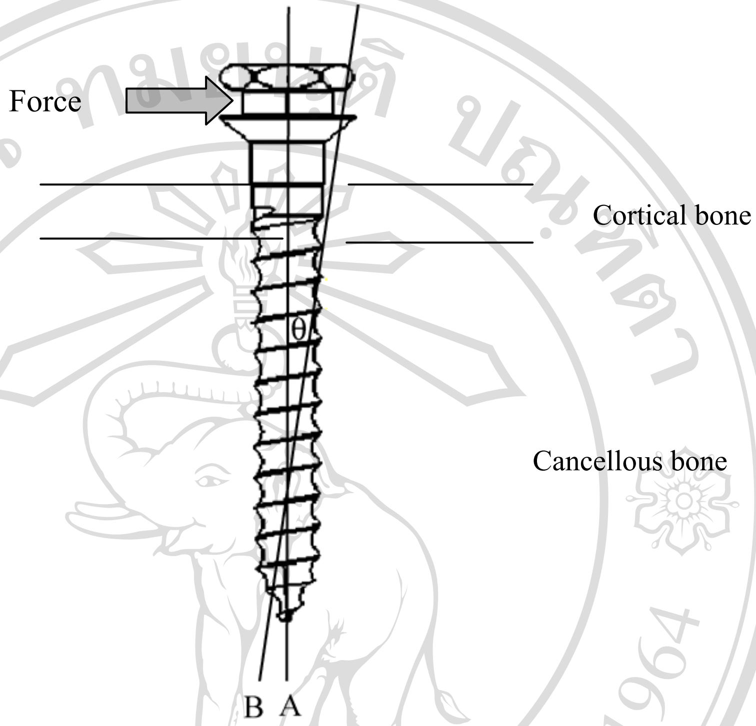

Figure 5.11 Diagram presenting the degree of rotation (Θ)
Note: "A" refers to longitudinal axis of miniscrew implant before loading force, "B" refers to longitudinal axis of miniscrew implant after loading force

Table 5.4 The degree of rotation in miniscrew implant diameter 1.2 mm

Length (mm)	Degree of miniscrew movement (Θ)
6.0	0.00928°
8.0	0.00712°
10.0	0.00650°
12.0	0.00467°

Although this study did not present any obvious trend of stress values influenced by miniscrew implant length, clinical studies may present different values resulting from individual bone quality.

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5.4.2 Influence of orthodontic loading force

In this investigation, the loading forces were applied in the horizontal direction at the neck of the miniscrew implant model to simulate orthodontic loadings in sliding mechanics. Since the most frequently-used amount of orthodontic loading force is 200 to 250 g, this study simulated loads in the range of 50 to 400 g. Increasing orthodontic loading forces in the present study resulted in increasing stress values.

The maximum stress value in the miniscrew implant models, 121.2 MPa, was found in a miniscrew model of size 1.0 x 6.0 mm. with a loading force of 400 g. The yield strength of miniscrew implant modeling in this study was 550 MPa. Therefore, the true factor of safety for miniscrew implant models in this study was 4.12 (Rothbart, 1996). According to Mott (2004), the design factor, or factor of safety, for ductile material under dynamic loading should be 2.0 to 2.5; consequently, miniscrew implants under loading of 50 to 400 g were safe.

The maximum stress value for cortical bone with loading of 400 g was 23.08 MPa in a miniscrew implant model of size 1.0 x 12.0 mm. This maximum first principal stress value was referred to 1685 μ strain. The maximum stress value for cancellous bone with loading of 400 g was 3.567 MPa in a miniscrew implant model of size 1.2 x 10.0 mm. This maximum first principal stress value was referred to 1928 μ strain. According to an experimental study by Melsen and Lang (2001), excessive force applied to dental implants used for orthodontic anchorage resulted in an increase in bone resorption rate. Functional strain between 3,400 – 6,700 μ strain was reported to maintain a normal bone remodeling rate, whereas strain above this range has been reported to cause a high percentage of bone resorptive surface. Strain values in the present study were lower than the excessive level identified by Melsen and Lang (2001). Therefore, surrounding bone is safe from excessive resorption with loading forces of from 50 to 400 g applied to miniscrew implants of all sizes.

According to the results of this study, at the same length of miniscrew implant, the stress values in miniscrew implant and cortical bone decreased with increasing miniscrew diameter, whereas in miniscrew implants with the same diameter the stress values increased slightly with increasing miniscrew implant length. Additionally, at loading force of 50 g, the miniscrews with diameters of 1.6

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and 1.8 mm were safer for cortical bone than were miniscrews with diameters of less than 1.4 mm because stress values in these sizes were lowest (Figure 5.7 B). The 4.0 mm-long miniscrew showed a significantly higher stress value in cancellous bone than did miniscrews of other lengths. However, increasing miniscrew implant length did not remarkably influence stress values. Moreover, the survival rates in an experimental study (Freire *et al.*, 2007) were higher in 10.0 mm-long screws than in 6.0 mm-long screws, and the success rates in a prospective clinical study (Chen *et al.*, 2006a), for 8.0 mm-long miniscrews were higher than for 6.0 mm-long miniscrews. Furthermore, removal torque increases with increasing screw length(Chen *et al.*, 2006b). Therefore, biomechanically, the appropriate size of miniscrew implant should be 1.6 to 1.8 mm in diameter with length of more than 4.0 mm.

Loading force was not only parameter effecting the biomechanical properties of surrounding bone. Van Oosterwyck *et al*. (1998) evaluated the influence of bone mechanical properties and implant fixation upon bone loading around dental implants and found that bone loading patterns were highly sensitive to bone properties, bone anatomy, prosthetic design and implant loading conditions. Meyer *et al.* (2001) evaluated bone loading pattern around implants in average and atrophic edentulous maxillae and reported that overloading transmitted from implants to bone seemed to mainly depend on bone quality.

5.5 Conclusions

The stress distribution patterns of miniscrew implant models showed that the cervical portion, form the first to the third threads was a weak point of the body. Increases in the diameter of miniscrew implant models improve the biomechanical properties of miniscrew implant and surrounding bone models, whereas increases in length of miniscrew implant models did not. Biomechanically, recommended sizes of miniscrew implants should be 1.6 to 1.8 mm in diameter with lengths more than 4.0 mm. Finally, miniscrew implant and surrounding bone modeling with all sizes were safe with loading forces of 50 to 400 g.

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