INTRODUCTION

Mini-screw implants offer an option for high anchorage orthodontic treatments that are relatively inexpensive, easily implemented and predictable enough to be used routinely in medical practice. However, clinicians often have difficulty placing a mini-screw at the optimal anchorage site due to intra-oral space limitations, leading to unexpected traction effects on the tooth movements. Improper anatomical anchorage site positioning for mini-screw placement can bring about tooth root contact, puncture the maxillary sinus and neurovascular damage which increase the risk for mini-screw failure. Even in mini-screws without surrounding anatomical tissue damage, poor implantation angle might influence the suboptimal biomechanical stabilization, i.e. primary stability of temporary skeletal anchorage devices between the screws and bone. High implant failure risk is increased when multiple mini-screws are inserted at hazardous areas, such as open/deep bites, tooth intrusion, gummy smile and distalization situations.

Posterior tooth intrusion is essential for the treatment of vertical excess cases such as an anterior open bite. Compared with anterior tooth intrusion, posterior intrusion is harder to achieve owing to molars and bicuspids typically having more voluminous roots, which causes the alveolar bone to respond more significantly, thereby extending treatment time. One single posterior tooth intrusion, usually requires two mini-screws, one inserted at the buccal and the other at the palatal to provide a controlled vertical movement without undesirable inclinations. Improper mini-screw insertion path at the palatal site easily damages the surrounding tissue, affecting the biomechanical performance, leading to screw loosening when the insertion position is near the infrrazygomatic crest and close to the tooth roots. Controversy remains regarding the best anatomical site for the mini-screw and the force vectors to be applied. The screw position and force units will result in different moments and forces in the three spatial planes.

The use of custom-made metal surgical guides and templates has been proposed for transferring radiographic preoperative planning information to the surgical site and outlining the ideal implant axis to promote safe mini-screw placement into the dento-alveolar bone. Computed tomography (CT) images were recently used for diagnosis and treatment planning with CAD (computer-aided design) and RP (rapid prototyping) techniques integrated to fabricate surgical templates for achieving controllable precise mini-screw placement. Damage to the surrounding tissues can be avoided using CT images, but not considering mechanical factors before mini-screw insertion might lead to improper screw implantation and poor primary stability, causing screw loosening. The primary stability of a mini-screw determines its survival rate and reliability. Screw stability is dominated by the mechanical retention between the bone quality and mini-screw. Therefore, mini-screw survival rate success can be improved if biomechanical analysis accompanies custom-made surgical template fabrication to obtain the most favorable mechanical performance and screw position insertion accuracy.

This study integrates reverse engineering (RE), RP, CAD and finite element analysis (FEA) technologies to produce customized mini-screw surgical templates.
with mechanical considerations. Surgical planning with accurate insertion path is obtained using templates and existing mechanical considerations to assist dentists in increasing the success rate in multi anchor treatment for posterior molar intrusion.

MATERIALS AND METHODS

Image processing and mini-screw inserted path definition
The patient was a 26-year-old male patient, with dental mal-alignment shallow bite symptoms. Two mini-screws were planned for insertion at buccal and palatal sites to provide a controlled vertical movement with power-chain for the intruding maxillary first molar (Fig. 1a). In order to ensure that the surgical template would fit onto the patient’s teeth and soft tissue, a maxilla impression was first taken to make a plaster model. A radiographic template was fabricated using the vacuum forming technique. Aluminum wires were attached to the radiographic template as landmark points and the mucosa shape and thickness recorded. The radiographic template was then placed on the patient’s dental arch to perform the CBCT (Cone-Beam Computed Tomography, Asahi AZ3000, ASAIHROENTGEN Kyoto, Japan) scan to obtain images (Fig. 2). These images show the hard tissue positions (jawbone or teeth) and present the position and thickness of the soft tissue through aluminum wire positioning. A digital model of the patient’s teeth, jawbone shape and soft tissue was constructed using reverse engineering software (ProEngineer Wildfire 5, Parametric Technology, Needham, MA, USA) for further biomechanical FE analysis (Fig. 3a).

Molar intrusion was performed using two mini-screws as the anchors. The screw insertion path planning was based on the reconstructed image model according to the jawbone and teeth positions. One mini-screw was inserted between the second premolar and first molar with the horizontal direction at the buccal site (Figs. 1b and c). Another screw was placed with two optional paths at the palatal site, i.e. inserted to the molar occlusal surface inclination at an angle of 60° or 15°, defined as P60 and P15 models (Figs. 1b and c). Models with better primary stability performance between these two palatal insertion angles were based on the FE analysis results selected and exported as the customized surgical template.

FE model generation and analysis
Dual-thread mini-screws (Ti6Al4V) 1.6 mm in diameter and 8 mm in length were selected as the anchor implants inserted at the buccal and 2.0 mm in diameter and 12 mm in length at palatal sides (Bomei, Taoyuan, Taiwan), respectively. Solid models of the teeth, jawbone and mini-screws were then imported into ANSYS (ANSYS, v11.0, Swanson Analysis, Houston, PA, USA) for assembly with their corresponding relative positions to generate CAD models based on the screw insertion paths at the buccal site and its relative two optional positions at the palatal sites (Fig. 4). Different meshes of two insertion FE models (P60 and P15) were generated using quadratic ten-node tetrahedral structural solid element (Solid 187) with hyper-elastic and larger deformation capabilities after mesh convergence testing while controlling for the strain energy and displacement variations of <5% for models with three different element sizes. Nonlinear frictional contact elements (defined as surface-to-surface) were used to simulate the mini-
screw and bone adaptation for immediate placement (not-osseointegration). The numbers of total element/nodes for the P60 and P15 models were 226713/270678 and 223276/278792, respectively (Fig. 4). All materials exhibited linear elastic, homogeneous and isotropic material properties in the simulations, as adopted from the relevant literature (Table 1)\textsuperscript{29-31).} The exterior nodes at the mesial and distal alveolar bone surfaces were fixed in accordance with the boundary conditions for each model. The 100 gf traction force magnitudes in opposite directions between the buccal/palatal mini-screw heads and corresponding molar adhesive bracket positions were applied as the orthodontic force conditions. The maximum first (1st) principal strain for the surrounding bone, maximum micro-motion between the screw and bone and mini-screw head displacement were recorded to evaluate the mini-screw primary stability (Fig. 4).
Fig. 4 (a) (b) FE mesh models including bone, PDL, tooth and mini-screws for models P60 and P15 and 100 gf traction forces were applied in opposite directions between the buccal/palatal mini-screw heads and corresponding molar adhesive bracket positions; (c) Detailed mesh of mini-screws.

Table 1  Material properties simulated in this study

| Material                  | Elastic modulus (MPa) | Poisson’s ratio | Reference            |
|---------------------------|-----------------------|----------------|----------------------|
| Cortical Bone             | 13,700                | 0.3            | Lin et al. 2006      |
| Cancellous Bone           | 1,370                 | 0.3            | Lin et al. 2006      |
| Tooth                     | 18,600                | 0.31           | Barink et al. 2003   |
| PDL                       | 68.9                  | 0.45           | Poggio et al. 2006   |
| Mini-screw (Ti6Al4V)      | 110,000               | 0.33           | Sawada et al. 2013   |

Surgical template fabrication and interfacial adaption test

The customized surgical template CAD model was fabricated with a 1.8-mm-thick layer average offset based on the simulated mucosa and teeth model with better primarily stability performance between P60 and P15 models (Figs. 3a and b). Insertion holes with 1.7 mm/2.2 mm diameters (larger than mini-screw diameters with 1.5 mm/2.0 mm) and guided cylinders with 5.5 mm diameter and 6 mm height were generated according to the mini-screw insertion positions at the buccal and palatal sides on the P60 model (better performance of primary stability) (Fig. 3c). Spin-off slots on the insertion holes and guided cylinders were designed to allow taking the customized surgical template out from the patient.

Fig. 5  Customized surgical template was fitted on the plaster teeth/mucosa model of the patient to perform the interfacial adaption test. Five (a–e) buccal-palatal direction sections were obtained using a diamond saw and scanned images using a non-contact video measurement system to measure the gap sizes for 5 points with 45° separation in the counterclockwise direction on each section.
The solid surgical template model can be exported as a stereo-lithographic (STL) file that can be loaded into a RP printer with 0.1 mm resolution (V-FLASH Desktop Modeler, 3D Systems, Rock Hill, SC, USA) to duplicate the SLA (Stereolithography) plastic template (Fig. 3c).

An interfacial adaption test was performed to evaluate the interface fitness accuracy and measure the gap sizes between the surgical template and teeth/mucosa tissue at different section planes. The customized surgical template was fitted onto a plaster teeth/mucosa patient model and embedded in clear rectangular test boxes with epoxy resin (Truetime Industrial, Kaohsiung, Taiwan) to provide a stable base. The resin block was sectioned in the buccal-palatal direction from the second premolar to molar with 5 section slices using a low speed diamond saw with copious cooling (CL50 Precision Saw, Top Tech Machines, Taichung, Taiwan) (Fig. 5). The section slices were scanned using a non-contact video measurement system (SVP-2010, ARCS, Taichung, Taiwan) to measure the gap sizes at 5 points with 45° separation in the counterclockwise direction on each section. This non-contact vision measurement system is comprised of a high resolution CCD (Charge-coupled Device), monocular lens with continuous zoom magnification system, high-precision linear scales (0.5 μm) and stage. The images were obtained using 17.5 times magnification with a color CCD camera and transferred into an imaging program to perform further evaluation.

Clinical application
After the patient was given a local anesthetic the surgical template was placed intra- orally. The template seating should be confirmed with the surgical template held in place by the patient’s bite forces. Once the template was seated the patient was asked to bite the template to expose the guide surgical sites. The mini-screws were then placed with a screwdriver (Fig. 6). The surgical template can then be moved through spin-off slots. X-rays were taken after implantation to assess whether damage to the peripheral tissue occurred in order to assess mini-screw stability. Orthodontists confirmed whether inflammation occurred to the surrounding region or whether patients experienced
other discomfort. The patients were given instructions for postoperative care and antibiotics were prescribed (500 mg amoxicillin 4 times daily for 3 days). Orthodontic loading was applied 1 to 2 weeks later.

RESULTS

The simulated results showed that the maximum strains in the surrounding bone from model P15 were lower than those from model P60 regardless of buccal/palatal side (Table 2). The maximum micro-motion between the screw and bone and mini-screw head displacement from model P60 were found lower than those from model P15. The maximum micro-motion between the screw and bone increased 3.18 and 3.28 times at the palatal and buccal sides when compared to the corresponding values for models P15 to P60 (Table 2). The corresponding differences were 74.9 and 2.06 times at the palatal and buccal sides for mini-screw head displacement. The strain concentration location for the simulation models was also determined. The results showed that the strain concentration regions were located at the cortical bone cervical regions irrespective of the anchorage locations (Fig. 7).

The interfacial adaption test results showed that average gap sizes of 5 points in the different tooth sections ranged from 0.95–1.84 mm. The interfacial gap sizes at the second premolar (0.95 mm at section a) and molar (1.19 mm at section e) were lower than those for

![Fig. 7](image_url) The strain concentration location for the simulation models showed that the strain concentration regions were located at the cervical regions in cortical bone, irrespective of the anchorage locations.

Table 2 Surrounding bone maximum strains, maximum micro-motion between screw and bone and displacement of mini-screw head of models P60 and P15 and their variations

| Model | Maximum 1st principal bone strain (μ) | Maximum micro-motion between screw and bone (mm) | Variation of micro-motion (P15/P60) | Displacement of mini-screw head (mm) | Variation of displacement (P15/P60) |
|-------|--------------------------------------|-----------------------------------------------|---------------------------------|----------------------------------|----------------------------------|
| P60   | Palatal | 596 | 2.71E-3 | — | 2.67E-4 | — |
|       | Buccal | 768 | 1.41E-3 | — | 1.71E-4 | — |
| P15   | Palatal | 525 | 8.60E-3 | 3.18 | 2E-2 | 74.90 |
|       | Buccal | 471 | 4.63E-3 | 3.28 | 3.53E-4 | 2.06 |

Table 3 Gap sizes of the interfacial adaption test base on using piece of aluminum wires as landmark points and attached on the radiographic template

| Section | Placement | Point (unit: mm) | Average (unit: mm) | SD |
|---------|-----------|-----------------|--------------------|----|
| a       | 2nd Premolar | 0.52 0.62 1.34 1.32 0.93 | 0.95 | 0.34 |
| b       | 1st Molar  | 0.53 0.64 2.20 2.67 0.61 | 1.33 | 0.92 |
| c       | 1st Molar  | 0.53 1.63 2.25 1.69 0.61 | 1.34 | 0.67 |
| d       | 1st Molar  | 0.57 2.27 2.01 3.30 1.04 | 1.84 | 0.96 |
| e       | 2nd Molar  | 0.68 1.17 1.91 1.08 1.12 | 1.19 | 0.40 |
| Total Average | 0.52 0.62 1.34 1.32 0.93 | 0.95 | 0.34 |
the first molar (1.33±0.92–1.84±0.96 mm at sections b–d) (Table 3).

DISCUSSION

Integrating CT images and CAD techniques to fabricate surgical templates have been used widely for implant and prostheses design. The accuracy of this approach has been shown in many studies14,16,22,32-34. Similar to implants, mini-screws placed in limited interradicular spaces need high accuracy. CBCT image reconstruction must be used for measuring interradicular spaces for accurate and reproducible mini-screw placement. This process also offers greater and more reliable deeper 3D anatomical structure assessment. Poggio et al., Park and Carano et al. measured the distances between the roots of adjacent teeth using different methods. They all demonstrated that the interradicular space for mini-screws is limited, necessitating that mini-screws must be accurately placed32,35,36. In this study, with two possible paths at the palatal site, i.e. inserted to the molar occlusal surface inclination at an angle of 60° or 15° can be achieved for optional orthodontic treatment. However, an obvious difference was found in the primary stability performance between these two palatal insertion angles based on the FE analysis results. We therefore suggest that biomechanical evaluation always be considered in customized surgical template fabrication. Identifying the safe zone using the CBCT image reconstruction model is more reliable.

Excessive loads on the implant lead to high mechanical stimulus at the interface and transfer stress to the bone causing original bone modeling/remodeling equilibrium, producing premature failure over time is well known as the “mechanostat theory”37-42. Strain is accepted as the mechanical stimuli for bone remodeling around dental implants. Bone remodeling is initiated at some critical strain level (4,000 μ). Studies pointed out that a newly formed bone area is proportional to the induced strain magnitude37,38. High micro-motion might be caused by early loading during treatment. The literature suggests that there is a critical micro-motion threshold smaller than 150 μm that ensures implant stability43. Bone strain value, interfacial micro-motion between mini-screw and bone and mini-screw head displacement were therefore selected as the criteria for evaluation. The simulation results showed that the maximum strain to the surrounding bone in models P15 and P60 were far lower than the bone remodeling critical value, i.e. 4,000 micro-strain and did not damage the surrounding bone. However, maximum micro-motion between the screw and bone increased 3.18 and 3.28 times at the palatal and buccal sides when the corresponding values for models P15 to P60 were compared. The corresponding differences were 74.9 and 2.06 times at the palatal and buccal sides for mini-screw head displacement. This result implied that model P60, i.e. mini-screw inserted into the molar occlusal surface inclination at an angle of 60° provided better primary stability performance than a 15° angle on the palatal side. This result indicated that the screw loosening risk can be decreased when biomechanical analysis can be combined with CT image and CAD technique as complementally tools for safety considerations in surgical template planning procedures.

The stability and inherent support of the surgical template is a crucial factor. The template in this study was supported using 3 surfaces-occlusal teeth and buccal/palatal mucosa. The bite forces on the template provided support to keep it stable. Accuracy and security were evaluated using interface adaption testing to measure the gap sizes between the surgical template and teeth/mucosa tissue. A large average gap size for all sections was found as 1.33±0.33 mm that might cause insufficient retention through inducing surgical template instability, influencing the guiding accuracy. This phenomenon occurred was because fitting errors might accumulate during soft tissue image reconstruction owing to aluminum wire used as the landmark points to identify the soft tissue shape and thickness. Although this approach can locate the soft tissue positions, accurate geometry must be obtained in order to determine the actual soft tissue model reconstruction from unclear scan images using manual effort.

Image superimposition using laser scan and CBCT images was proposed to improve the surgical template

Table 4  Gap sizes of the interfacial adaption test base on using image superimposes through laser scan and CBCT images

| Section | Placement     | Point (unit: mm) | Average (unit: mm) | SD  |
|---------|---------------|------------------|--------------------|-----|
|         |               | 1    | 2    | 3    | 4    | 5    |       |       |
| a       | 2nd Premolar  | 0.27 | 0.20 | 0.38 | 0.50 | 0.23 | 0.31  | 0.11  |
| b       | 1st Molar     | 0.43 | 0.32 | 0.37 | 0.35 | 0.35 | 0.36  | 0.04  |
| c       | 1st Molar     | 0.31 | 0.24 | 0.41 | 0.24 | 0.29 | 0.30  | 0.06  |
| d       | 1st Molar     | 0.38 | 0.37 | 0.40 | 0.37 | 0.37 | 0.38  | 0.01  |
| e       | 2nd Molar     | 0.48 | 0.41 | 0.38 | 0.39 | 0.31 | 0.39  | 0.06  |
| **Total Average** | | | | | | | **0.35±0.04** |
interfacial fitting adaptation problem. The plaster model geometry was acquired from the patient using a laser scanning system (3Shape Scanners, 3D Scan, Rock Hill, SC, USA). These images were obtained from the patient’s dental arch plaster model with a good degree of similarity and may enhance the holding power and stability of the surgical template. Laser scanning plaster model and CBCT hard tissue (teeth and bone) images for FE model analysis can then be superimposed to identify the mini-screw insertion path positions. Insertion holes, guided cylinders and spin-off slots can be designed based on the superimposed image and exported to generate the SLA plastic surgical template. Interfacial adaption testing was performed again to compare the performances of this method. The average gap sizes in the different tooth sections were found to be smaller than 0.5 mm (0.30±0.06–0.39±0.06 mm) (Table 4). Differences in average gap size for all sections using the aluminum wire landmark and image superimposed methods were found as 74% [(1.33–0.35)/1.33]. These results confirmed that the image superimposed method can indeed improve surgical template suitability correction.

Using the RP (3D printing) technique makes the surgical template feasible. However, the detailed surgical template structure cannot be formed around the insertion hole/slot/guide hole, leading surgical templates with improper function to affect the guiding accuracy when the machine print resolution is not accurate. We suggest using a high precision 3D printing machine to fabricate the surgical template with complete detailed structure.

We presented a new method integrating CT image, CAD system, FE analysis and RP techniques to fabricate an accurate customized surgical template for orthodontic mini-screws. The customized surgical template was generated using biomechanical evaluation to increase the clinical survival rate. The clinical application in this study did not reflect the inflammation, screw loosening or other symptoms of discomfort that could occur one week after surgery. The study results verified the template’s accuracy and security. However, more clinical application cases are needed in long-term studies to verify the feasibility of the proposed method.

ACKNOWLEDGMENTS
This study is one part of the Master Thesis of Yu-Tzu WANG (2014) at the Dept. of Biomedical Engineering at National Yang-Ming University, Taiwan and supported in part by NSC project 100-2628-E-010-003-MY3 of the National Science Council, Taiwan.

REFERENCES
1) Baumgaertel S. Temporary skeletal anchorage devices: the case for miniscrews. Am J Orthod Dentofacial Orthop 2014; 145: 558-564.
2) Telma Martins de A, Mauro Henrique AN, Fernanda Catharino MF, Marcos Alan VB. Tooth intrusion using mini-implants. Dental Press J Orthod 2008; S: 36-48.
3) Araujo TM, Nascimento MHA, Bezerra F, Sobral MC. Ancoragem esqueletica on Ortodontia com miniimplantes. R Dental Press Ortopd Ortop Facial 2006; 42: 126-156.
4) Bae SM, Kyung HM. Mandibular molar intrusion with miniscrew anchorage. J Clin Orthod 2006; 40: 107-108.
5) Lin JC, Liu ED, Yeh CL, Evans CA. A comparative evaluation of current orthodontic miniscrew systems. World J Orthod 2007; 8: 136-144.
6) Kravitz ND, Kusnuto B. Risks and complications of orthodontic miniscrews. Am J Orthod Dentofacial Orthop 2007; 131: s43-s51.
7) Lee JH, Choo H, Kim SH, Chung KR, Giannuzzi LA, Ngan P. Replacing a failed mini-implant with a miniplate to prevent interruption during orthodontic treatment. Am J Orthod Dentofacial Orthop 2011; 139 : 849-857.
8) Lin YS, Yu JH, Chang YZ, Lin CL. Biomechanical evaluation of an orthodontic miniimplant used with revolving (translation and rotation) temporary anchorage device by finite element analysis and experimental testing. Implant Dent 2013; 22: 77-82.
9) Sawada K, Nakahara K, Matsunaga S, Abe S, Ide Y. Evaluation of cortical bone thickness and root proximity at maxillary interradicular sites for mini-implant placement. Clin Oral Implants Res 2013; A100: 1-7.
10) Motoyoshi M, Hirabayashi M, Uemura M, Shimizu N. Recommended placement torque when tightening an orthodontic mini-implant. Clin Oral Implants Res 2006; 17: 109-114.
11) Ottone JM, Oliveira ZF, Mansini R, Cabral AM. Correlation between placement torque and survival of single-tooth implants. Int J Oral Maxillofac Implants 2005; 20: 769-776.
12) Zhao L, Xu Z, Yang Z, Wei X, Tang T, Zhao Z. Orthodontic mini-implant stability in different healing times before loading: a microscopic computerized tomographic and biomechanical analysis. Oral Surg Med Oral Pathol Oral Radiol Endod 2009; 108: 196-202.
13) Schouman T, Murcier G, Goudet P. The key to accuracy of zygoma repositioning: Suitability of the Synpliciti customized guide-plates. J Craniomaxillofac Surg 2015; 43: 1942-1947.
14) Romeo A, Esteves M, Garcia V, Bermudez J. Movement evaluation of overerupted upper molars with absolute anchorage: an in-vitro study. Med Oral Patol Oral Cir Bucal 2010; 15: 930-935.
15) Liu H, Liu DX, Wang G, Wang CL, Zhao Z. Accuracy of surgical positioning of orthodontic miniscrews with a computer-aided design and manufacturing template. Am J Orthod Dentofacial Orthop 2010; 728: 1-10.
16) Suzuki KY, Suzuki B. Accuracy of miniscrew implant placement with a 3-dimensional surgical guide. J Oral Maxillofac Surg 2008; 66: 1245-1252.
17) Kitai N, Yasuda Y, Takada K. A stent fabricated on a selectively colored stereolithographic model for placement of orthodontic mini-implants. Int J Adult Orthod Orthognath Surg 2002; 17: 264-266.
18) Morea C, Dominguez GC, Wuo Ado V, Tortamano A. Surgical guide for optimal positioning of mini-implants. J Clin Orthod 2005; 39: 317-321.
19) Cousley RR, Parkberry DJ. Surgical stents for accurate miniscrew insertion. J Clin Orthod 2006; 40: 412-417.
20) Martin W, Heffernan M, Ruskin J. Template fabrication for a midpalatal orthodontic implant: Technical note. Int J Oral Maxillofac Implants 2002; 17: 720-722.
21) Bae MJ, Kim JY, Park JT, Cha JY, Kim HJ, Yu HS, Hwang CJ. Accuracy of miniscrew surgical guide assessed from cone-beam computed tomography and digital models. Am J Orthod Dentofacial Orthop 2013; 143: 893-901.
22) Sarment DP, Sukovic P, Clinthorne N. Accuracy of implant placement with a stereolithographic surgical guide. Int J Oral Maxillofac Implants 2003; 571: 7-18.
23) Ruppin J, Popovic A, Straus M, Spüntrup E, Steiner A, Stoll C. Evaluation of the accuracy of three different computer-aided surgery systems in dental implantology: optical tracking vs. stereolithographic splint systems. Clin Oral Implants Res 2008; 7: 709-716.

24) Rudolph H, Luthardt RG, Walter MH. Computer-aided analysis of the influence of digitizing and surfacing on the accuracy in dental CAD/CAM technology. Comput Biol Med 2007; 37: 579-587.

25) Friberg B, Senneryby L, Meredith N. A comparison between cutting torque and resonance frequency measurements of maxillary implants: a 20-month clinical study. Int J Oral Maxillofac Surg 1999; 28: 297-303.

26) Wilmes B, Su YY, Drescher D. Insertion angle impact on primary stability of orthodontic mini-implants. Angle Orthod 2008; 78: 1065-1070.

27) Miyamoto I, Tsuboi Y, Wada E, Suwa H, Iizuka T. Influence of cortical bone thickness and implant length on implant stability at the time of surgery-clinical, prospective, biomechanical, and imaging study. Bone 2005; 37: 776-780.

28) Katranji A, Misch K, Wang HL. Cortical bone thickness in dentate and edentulous human cadavers. J Periodontol 2007; 78: 874-878.

29) Asmussen E, Peutzfeldt A, Sahafi A. Finite element analysis of stresses in endodontically treated dowel restored teeth. J Prosthet Dent 2005; 94: 321-329.

30) Lin CL, Chang YH, Cheng MH. Evaluation of a reinforced slot design for CEREC system to restore extensively compromised premolars. J Dent 2006; 34: 211-219.

31) Barink M, Van der Mark PC, Pennis WM, Kuijs RH, Kreulen CM, Verdonschot N. A three-dimensional finite element model of the polymerization process in dental restorations. Biomaterials 2003; 24: 1427-1435.

32) Poggio PM, Incorvati C, Velo S, Carano A. “Safe zones”: a guide for miniscrew positioning in the maxillary and mandibular arch. Angle Orthod 2006; 76: 191-197.

33) Ohtani T, Kusumoto N, Wakabayashi K, Yamada S, Nakamura T, Kumazawa Y, Yatani H, Sohmura T. Application of haptic device to implant dentistry —accuracy verification of drilling into a pig bone. Dent Mater J 2009; 28: 75-81.

34) Choi JY, Choi JH, Kim NK, Kim Y, Lee JK, Kim MK, Lee JH, Kim MJ. Analysis of errors in medical rapid prototyping models. Int J Oral Maxillofac Surg 2002; 31: 23-32.

35) Park HS. An anatomical study using CT images for the implantation of micro-implants. Korean J Orthod 2002; 32: 435-441.

36) Carano A, Velo S, Incorvati C, Poggio P. Clinical applications of the mini-screw-anchorage-system (M.A.S.) in the maxillary alveolar bone. Prog Orthod 2004; 5: 212-235.

37) Frost HM. Wolff’s Law and bone’s structural adaptations to mechanical usage: an overview for clinicians. Angle Orthod 1994; 64: 175-188.

38) Cehreli M, Sahin S, Akca K. Role of mechanical environment and implant design on bone tissue differentiation: current knowledge and future contexts. J Dent 2004; 32: 123-132.

39) Chang SH, Lin CL, Haue SS, Lin YS, Huang SR. Biomechanical analysis of the effects of implant diameter and bone quality in short implants placed in the atrophic posterior maxilla. Med Eng Phys 2012; 34: 153-160.

40) Mellal A, Wissott HW, Botsis J, Scherrer SS, Belser UC. Stimulating effect of implant loading on surrounding bone. Comparison of three numerical models and validation by in vivo data. Clin Oral Implants Res 2004; 15: 239-248.

41) Abreu CW, Nishioka RS, Balducci I, Consani RL. Straight and offset implant placement under axial and nonaxial loads in implant-supported prostheses: Strain gauge analysis. J Prosthodont 2012; 21: 533-539.

42) Baek SH, Cha HS, Cha JY, Moon YS, Sung SJ. Three-dimensional finite element analysis of the deformation of the human mandible: a preliminary study from the perspective of orthodontic mini-implant stability. Korean J Orthod 2012; 42: 159-168.

43) Szmucler-Moncler S, Salama S, Reingewirtz Y, Dubrullel JH. Timing of loading and effect of micro-motion on bone-implant interface. J Biomed Mater Res 1998; 43: 192-203.