FINITE ELEMENT ANALYSIS OF MINISCREW PLACEMENT IN MAXILLARY ALVEOLAR BONE WITH VARIED ANGULATION AND MATERIAL TYPE

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ABSTRACT

JASON K POLLEI: Finite Element Analysis of Miniscrew Placement in Maxillary Alveolar Bone with Varied Angulation and Material Type
(Under the Direction of Dr. Ching-Chang Ko)

Mechanical stress may be associated with orthodontic miniscrew loosening, contributing to failure. This study evaluated stresses from loads on miniscrews with varying angulation/material type combinations using Finite Element Analysis(FEA). Left posterior maxilla and TOMAS® miniscrew models were constructed with buccal miniscrew insertion between UL5/6 at 45°,60°, and 90°angulations. Titanium(Ti), stainless steel(SS) and composite(Comp) materials were used for miniscrews. After retraction load placement models were solved with ANSYS 10.0. Maximum principle(MaxPS), minimum principle(MinPS) and von Mises(vonMS) stresses were evaluated in cortical bone and miniscrew for all angulation/material combinations. Peak ultimate tensile strengths were 1/3 below bone and miniscrew MaxPS. MaxPS in bone was greatest at 45° and least at 60°. Comp miniscrew stress trends don't follow Ti and SS with varied angulation. Varied angulation likely doesn't contribute to miniscrew failure. Stress properties of SS miniscrews compare favorably with Ti miniscrews; however, Comp shows variance which could be clinically and biologically significant.

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TABLE OF CONTENTS

LIST OF FIGURES

LIST OF TABLES

Table Page

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SECTION I

LITERATURE REVIEW

Orthodontic treatment inevitably involves movement of maloccluded teeth. This requires applying forces to these teeth. In nature, an equal and opposite force results from any applied force. To counteract this opposite force (which is generally unwanted in orthodontics), attempts to minimize or alleviate these unwanted forces are collectively termed “anchorage”. Traditional means of anchorage may be extraoral (high pull, cervical, and J-hook headgear) or intraoral (elastics, undertying, distal-jet, bonding 7s, TPA, lingual arch, Nance appliances). The relatively recent advent of temporary skeletal anchorage devices in orthodontics provides an additional method of anchorage control to assist in obtaining desired tooth movement in three dimensions.(1) Currently, there are two types of temporary anchorage devices (TADs) in orthodontics: screw implants and bone plates.(2) Although bone plates have advantages, they also have disadvantages such as limited placement location. Alternatively, screw implants are able to be used widely and are used much more frequently in practice. As such, this review will focus on screw-type implants, which have also been called: *mini-implan*ts, *micro-implants*, *microscrews* and *miniscrews*.(3)

Prior to the recent popularity explosion of miniscrews, conventional endosseous dental implants were the sole means of implantable dental anchorage. Endosseous dental implants have been used in dentistry for decades.(4, 5) As an outgrowth of the successful use of these conventional implants in general dental practice, these implants began to be used as an adjunct to orthodontic treatment. For much of the late 20th century authors advocated the use of endosseous implants to facilitate orthodontic anchorage requirements and tooth movement, and their use gained some popularity. (5-9) Then in 1997 Kanomi (10) became the first during this time period to report on the use of miniscrews for orthodontic purposes. Soon thereafter Costa et al (11) reported similar use of miniscrews, and over the next few years a number of reports and studies were published which served to increase the knowledge-base of miniscrew use in orthodontics. (12-14) Yet despite the seemingly recent incorporation of miniscrews into mainstream clinical orthodontic practice, the concept of skeletal anchorage was actually introduced to orthodontics as early as 1945 by Gainsforth and Higley,(15) but went largely unnoticed for several decades following. Today, the use of implantable temporary skeletal anchorage devices afford practitioners greater control of various facets of orthodontic treatment including: increasing orthodontic anchorage; virtually eliminating patient compliance issues with regard to wearing of appliances; decreasing overall treatment time in some cases; and occasionally permitting orthodontic treatments previously thought to be impossible without surgery.(16) It truly may give orthodontists the ability to "tackle cases outside the scope of conventional mechanics".(17)

Orthodontic anchorage may be employed either directly or indirectly.(18) When serving as direct anchorage, a force is applied directly from the miniscrew in order to obtain the desired biomechanical outcome. In indirect anchorage, a miniscrew is connected to a tooth or segment of teeth to stabilize them via an archwire or ligature wire, thereby incorporating them into the anchorage unit.

The buccal surface of the posterior maxillary alveolus is a popular location for miniscrew placement. Miniscrews placed in the posterior maxilla have been used to facilitate a number of biomechanical movements, including:

1) leveling gingival contour(19);

2) en masse distalization of the maxillary arch(20);

3) retraction of anterior teeth(21);

4) mesialization of maxillary molars for class III correction(22);

5) intrusion(22-26);

6) correction of canted occlusal planes(27);

7) molar uprighting(28-30);

8) extraction space closure(13, 31);

9) midline correction(27);

These applications and others, such as extrusion (19) and de-impaction of canines and molars (32, 33), may also indicate miniscrew use in other areas of the alveolus and jaws.

With the increased clinical use of temporary skeletal anchorage and miniscrews in particular, the orthodontist's tolerance of the incidence of miniscrew failure is expected to be very slim. The success rate of traditional endosseous dental implants is as high as 95-97%.(34, 35) However, miniscrew success rates are lower and more variable with reports ranging from 70.7% (36) to 78.4% (37) to 83.8% (38) to 85.5% (39) to 86.8% (22) to 91.1% (23) to 91.6%.(40) (41) Risk of failure has been extensively reported on, and is increased when correlated with: increased peri-implant inflammation (37, 40, 42); placement in the mandible(23, 40, 41, 43, 44); small diameter (37, 45); root proximity (46); high mandibular plane angle(37); placement in the right side of the mouth(40); placement in nonkeratinized tissues(43); tightness of implant insertion(39); self-drilling miniscrew design (42); mandibular retrusion (42); uprighting movement (42); decreased bone density (42); over insertion (47); and loading within two (48) or three weeks (42). However, many reports disagree with many of the factors related to failure such as side of mouth placement,

placement location, mandibular angle. Subsequent studies that have sought to uncover the biological interaction of some of these interactions have been inconclusive. (49, 50) Varied miniscrew placement angulation is another factor that may contribute to failure directly or may impact some of the above-listed factors indirectly. However, this variable has not been studied to date and merits evaluation as it may provide understanding of why some miniscrews fail. This in turn may contribute to reducing the risk of failure in clinical practice where hundreds of miniscrews may be placed.

Although miniscrews have been placed in virtually every bony location in the mouth, one region that has been frequently used and easy to access is the posterior buccal maxilla. For applications in this region, the angle of insertion reported in the literature varies from 30^o to 90^o relative to the vertical long axis of the alveolus. (14, 27, 51) It has been postulated that insertion angle of 30^o to 70^o from the occlusal plane (i.e. 20^o to 60^o to the cortical bone long axis) (47, 52) is optimal to allow sufficient cortical bone engagement. Theoretically, this will increase the potential for maximal anchorage, while preventing miniscrew slippage along the surface of the bone during insertion. More significantly, the potential for generation of undesirable stress levels in bone exists, and altering the angulation of miniscrew placement may help to reduce stresses. Measuring stress generated in vivo at different miniscrew placement angulation is not feasible and as a result there are no published studies of such. However, is it possible to evaluate bone stress at varied miniscrew angulation with other methods and techniques, such as Finite Element Analysis (FEA).

Finite Element Analysis, described in the next paragraph, is a useful tool to evaluate the biomechanical performance of miniscrews at varying angulations and thus the potential for angulation to contribute failure. But despite their usefulness, most existing FE models are based on simplified geometry of alveolar bone. Furthermore, no model at the present time can really represent various morphological situations or predict the actual constrictions for miniscrews with various angulations of insertion.

FEA is a computer-based numerical simulation technique that is widely used for predicting the mechanical behavior of engineering structures. It can be used to calculate deflection, stress, vibration, buckling behavior and many other phenomena. It can be used to analyze either small or large-scale deflection under loading or applied displacement. It can analyze elastic deformation, or “permanently bent out of shape” plastic deformation. In the finite element method, a structure is broken down into many small simple blocks or elements.(53) It was first developed in 1943 by R. Courant, who utilized the Ritz method of numerical analysis and minimization of variational calculus to obtain approximate solutions to mechanical stress and deformation for each element of complex structures. Since the 1980’s, FEA has been rapidly implemented into computer programs. Such computer based numerical stress analysis methods allow the complex distributions of elastic and inelastic stresses in engineering components to be more routinely calculated, and allow analyses to be performed for static and dynamic loads. The process of analysis is facilitated through the processes of component drawing and reverse engineering by using Computer Aided Design (CAD) systems.

In addition to initial engineering applications, FEA has been widely used in dentistry, including Thresher’s stress analysis of human teeth in 1972 (54) and Yettram’s evaluation of restored tooth crowns in 1976. (55) Stress analysis studies of inlays, crowns, bases supporting restorations, fixed bridges, complete dentures, partial dentures and endodontic posts have been reported, as well as studies of teeth, bone, and oral tissues. FEA has also been used to model and predict the biomechanical performance of various implant designs used in dentistry and medicine. (56-61) These predictions may be used to determine the effect of clinical factors on implant success. For example, FEA has been used by a number of researchers to evaluate the interfacial stress distribution in the area where the endosseous implants contact cortical and trabecular bone. (62) FEA has been used in orthodontics applications in general (63, 64) and for miniscrews in particular. (65-67) However, these studies used simplified bone block models in pull-out tests. To date there is no data published (English) that has demonstrated FEA of a model that replicates a "clinically simulated" scenario complete with human alveolar bone, teeth, PDL, and applied forces.

Although FE offers a number of advantages in evaluating stresses that may not otherwise be determinable, there are also disadvantages. One is the significant time and effort required to generate a realistic model. Creating a completed accurate maxillary or mandibular arch, complete with enamel, dentin, pulp and PDL for each tooth, as well as lamina dura and distinct cortical and trabecular bone may take hundreds of hours. One inherent shortcoming in utilizing FEA simulation of the mechanical behavior of miniscrews (and skeletal anchorage in general) is the modeling of actual anatomy of human hard- and soft-tissues, and their response to mechanical force over time. FEA provides a "snap-shot" view of conditions within the model; it does not depict changes that occur over time, such as bone remodeling, healing, etc. Consequentially, the use of FEA in evaluating biomechanical performance of miniscrews carries with it a set of limitation and assumptions. Some of the limitations for FE use include: (1) detailed geometry of the bone and implant to be modeled, (2) material properties, (3) boundary conditions, and (4) interface between bone and implant. (62, 68)

FEA results are evaluated both visually and numerically. One of the advantages of FEA is that it allows the user to evaluate internal stress/strain of a biomedical system visually. Through illustration, FEA demonstrates the concentration and distribution of stress within particular boundaries by utilizing a continuum of color to reflect varying stress levels. Numerically, FEA allows the modeled region to be “solved”—to be numerically described. Statistical analysis is not used because only one unique solution is possible for any specific set of conditions. FEA allows inferences and impressions to be made from the single mathematical solution for each variable or scenario. Likewise, effect of ‘clinical significance’ may be made from these results when combined with other existing data.

Despite some limitations, FEA is a useful tool in many applications. It especially has the potential to provide great insight into stress levels created in bone when orthodontic miniscrews are placed at varied angulations in a clinically-representative model.

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SECTION II

MANUSCRIPT

INTRODUCTION

Temporary anchorage devices have gained widespread popularity in orthodontics during the past decade. A broad spectrum of anchorage devices including miniscrews and on-plants have been introduced, advocated and used in both research and clinical settings. The most frequently used temporary anchorage devices are miniscrews or “TADs” as they have been nicknamed. Miniscrews are generally straightforward to place and remove, are amenable to placement in various locations in the mouth, are widely adaptable to various orthodontic anchorage requirements, and are typically constructed from biologically compatible titanium. Although their design varies among manufacturers, TADs are generally 1-2mm in width, 6-12mm in length (thread area), and have a head which serves as a point of force application to allow desired tooth movement (Figure 1).

Although miniscrews have become very popular in clinical use, their use is often substantiated as much by clinical experience as by a sound evidence base. One aspect of miniscrew use that lacks significant evidence is the evaluation of the miniscrew placement angulation and the subsequent stress generation in both miniscrew and bone when placement angulation is varied. Some researchers have advocated placement angulation that avoids tooth roots or increases cortical bone contact. Others advocate for angulation that facilitates hygiene or ease of placement. Yet the fact remains that insufficient evidence is available to state what angle of placement is preferred. The few existing attempts to evaluation angulation and stress generation have been carried out with in vitro pull-out tests and basic computer-aided design (CAD) models(1-3) rather than in vivo tests largely because in vivo human studies which accurately evaluate stress – especially in bone – are difficult to conduct.

There are challenges inherent in evaluating mechanical effects of miniscrew use on bone and miniscrew. These challenges include the inability to clinically measure stress levels in patients' bone and inability to visualize the stress patterns that are generated. In an attempt to overcome these challenges, researchers have utilized the Finite Element (FE) analysis engineering tool. This technology has been utilized in dentistry and orthodontics due to its ability to simulate and evaluate stresses of interest in CAD models.(1, 4-15) FE has particularly useful application in evaluating aspects of miniscrew use in orthodontics.(1, 2) However, most FE models reported to date have been simplified or incomplete reflections of normal human anatomy. This study sought to investigate if a more anatomically detailed, realistic FE model that previously used could be constructed.

The stresses generated around the miniscrew are a result of the structure and properties of the materials/tissue involved. Although bone, teeth, and other anatomical structures are predetermined in each patient, the material or design used in a miniscrew is potentially alterable. However, the effect of varied miniscrew material composition for orthodontic application is virtually unstudied. The potential for the use of miniscrews composed of other materials exists. Typically titanium (Ti) has been the material of choice; however, alternative materials such as stainless steel (SS) and composite (comp) may prove useful for miniscrew use if they can help reduce stress generated both within the miniscrew itself and within the cortical bone.

Despite a lack of evidence from in vivo, in vitro, or FE studies to support preferred TAD placement angle, different authors have reported or recommended a wide range of angulations for TAD placement.(16-20) Consequently, the purposes of this study were 1) to build a more realistic model using high-resolution CT data, and 2) to evaluate change in angulation with miniscrews of varied material types. The model will include all normal anatomic components, including enamel, dentin, pulp, PDL, lamina dura, cortical bone, and trabecular bone. By evaluating stresses generated in both miniscrew and bone when orthodontic TADs are placed into alveolar bone at varying angulations with varying material types, it is anticipated that additional evidence may be gained to support appropriate applications for clinical TAD use. Additionally, the effect of varied miniscrew material type will be evaluated.

MATERIALS AND METHODS

I. Tooth-Bone Model:

a. Maxilla

Three-dimensional anatomy of the human maxilla and four teeth was constructed. Using previously obtained CT data from a dentate human maxilla, incremental slices were generated with 0.5mm thickness between each slice. Following its generation, each slice was imported into the Computer-Aided Design (CAD) software program Solidworks (Solidworks Corp., Concord, MA, USA). Each imported μCT slice served as a template from which bony and dental structures were outlined in 2-dimensions. Slices were then sequentially stacked and the outlines of each separate maxillary bony component was connected to construct the 3-dimensional surfaces of cortical bone, lamina dura, and sinus. The cortical/trabecular bone interface was subsequently created by modeling a second surface offset 0.5mm internal to the external cortical surface in order to define the cortical and trabecular bone boundary.

Model boundaries were set after 3-D models of the posterior right maxilla and corresponding maxillary dentition were generated. These were established at: the interproximal region between the maxillary right canine and first premolar as the mesial boundary; the distal aspect of the maxillary tuberosity as the distal boundary; the complete coronal anatomy of all teeth as the inferior boundary; and all maxillary structures (including sinus and zygoma) up to 15mm superior to tooth apices as the superior boundary (Figure 2).

b. Tooth

The maxillary right first premolar through second molar teeth were each constructed from μCT data (μCT40, Scanco Medical, Bassersdorf, Switzerland; courtesy of Paulo Cattaneo, Dept. of Orthodontics, Royal Dental College, University of Aarhus) in the same manner as described above. Dimensions of the enamel, dentin and pulp chamber for each tooth were determined from the μCT slices, and each structure was constructed accordingly in Solidworks (Figure 3). Each tooth was then assembled with the proper pulp, dentin and enamel (Figure 4). PDL and lamina dura were then constructed around each tooth, with the thickness of both set at 0.5mm.

c. Assembly

Following construction of the maxilla and teeth, the modeled teeth (including associated PDL and lamina dura) were inserted into the maxillary bone model. Coronal interproximal contacts were ensured, and interradicular distances were similar to those reported as adequate by Poggio et al.(21) Teeth and bone were combined with Boolean operations. Each bony and dental structure was then created by subtracting all other model components and saved for later model assembly. Lamina dura remained merged with cortical bone as once contiguous component. After each component was saved individually, all components were merged into a final maxillary model (Figure 5). Young's modulus and Poisson's coefficients assigned for each component type were similar to those reported in the literature(5, 22, 23) and were applied isotropically (Table 1).

II. Miniscrew Model:

Orthodontic miniscrews were created using Solidworks CAD software. A commercially available miniscrews produced by TOMAS® was chosen to be modeled due to its broad clinical use in private practice and in the University of North Carolina Orthodontic clinic at the time of project commencement. Dimensions for the 8mm long, 1.6mm diameter TOMAS miniscrew were provided by the manufacturer, and included: taper, thread pitch, thread depth, minor diameter, thread root, head size and design, etc. (Figure 6).

The miniscrew outline was created using the sketch function in Solidworks and revolved into three dimensions. The helical sweep function was used to create a continuous, spiral thread. Subtraction cuts were used to create the appropriate head configuration after hexagon ring placement (Figure 7).

III. Miniscrew-Maxillae Models:

As a parametric study design, the effects of varying both miniscrew angulation and material type were modeled. To evaluate miniscrew angulation, Following 3D maxilla and miniscrew model construction, each miniscrew type was inserted separately into the maxillary model from the buccal surface between the second premolar and first molar using Solidworks. The miniscrew was inserted at angles of 90°, 60° and 45° relative to the surface of the cortical bone (Figure 8), and was placed so that the miniscrew neck/thread interface was coincident with the external contour of the cortical bone (Figure 9). For each angulation, the point of intersection between the cortical bone surface and the central axis of the miniscrew was maintained constant to ensure reproducibility and consistency between models. Boolean operations were performed and a completed model assembly was created at each angulation (Figure 10).

Variation due to material property difference was also evaluated. Subsequent models were created where miniscrew material was varied from Titanium (Ti) to Composite (Comp) and Stainless Steel (SS) in the 45°, 60° & 90° TOMAS models.

IV. Analysis of Stresses:

Three different types of stress were evaluated in this study (Table 2). They are:

-maximum or1st principle stress [tensile stress, MaxPS; $\sigma_{max}=(\sigma_x+ \sigma_y)/2+\tau_{max}$];

-minimum or 3rd principle stress [compressive, MinPS; $\sigma_{min}=(\sigma_x+ \sigma_y)/2-\tau_{max}$], and

-von Mises stress [equivalent, vonMS; $\sigma_v= \sqrt{([(\sigma_1-\sigma_2)^2+(\sigma_2-\sigma_3)^2+(\sigma_3-\sigma_1)^2]/2)}$].

MaxPS is the peak value at which point the tensile stress in a material is exceeded. MinPS is the peak measure of compressive stress resulting from a load or force applied to a material. The greater the negative value of the stress, the greater the compressive load. Von Mises stress is a measure of the elasticity of a material, and represents the point at which the elastic limit is exceeded and permanent deformation results.

The IGES format file of each finished model was exported to ANSYS 10.0 Workbench (Swanson Analysis Inc., Huston, PA, USA) and FE models with 10-node tetrahedral h-elements were generated for each assembly. The final FE mesh generated for each model contained approximately 91,500 elements, which was sufficient to obtain solution convergence. Following FE mesh generation, the model was fixed at the palatal, mesial, and superior boundaries. A 150gm loading force(24, 25) from the mesial (coordinates of X= -0.7N; Y= -1.2787N; Z= -0.2N) was then applied to the miniscrew for each model to simulate distalization of anterior teeth (Figure 11). All materials were linear and isotropic, and the miniscrew/bone interface was assumed to be rigidly bonded. Each model was mathematically solved and each solution saved as a .db format file. This allowed for visualization of different stresses discussed below.

Stress analysis was performed both visually and numerically (Figure 12). Visual color mapping depicts stress location and intensity: areas corresponding to greatest stress are bright red and areas of least stress are dark blue. Intermediate stress values are progressively colored along a rainbow red to blue. Stress values were scaled and deformation was standardized to zero for consistent visual comparison among models. Stress patterns and distributions among simulations were compared to evaluate differences of stress location across models.

Numerical MaxPS, MinPS and vonMS values were obtained. Two-way ANOVA was performed to evaluate for significance between angulations and material types and significance was set at $p<0.05$. Certain regions of interest (ROIs) where particular attention was given to stress levels include: the mesial and distal surfaces of the miniscrew neck, the buccal cortical bone region between the second premolar and first molar, and all cortical and trabecular bony surfaces along the length of the miniscrew/bone interface. The neck of the miniscrew was evaluated due to the high degree of stress concentration that is usually found in that area in traditional endosseous implants using FEA.

The yield strength of the materials in the model (bone and titanium) was equated with those extracted from existing studies. This data was compared with the maximal stress levels found in the simulations run in this study to determine if there is a significant chance for clinical failure of the implant or of the bone based upon the excessive stress levels.

RESULTS

Peak stress values for each model simulation are located in (Table 3). Stress patterns on both cortical bone and miniscrew from each simulation were captured (Figures 13-18). When mean average stress values for the three miniscrew material types at each angulation were compared with the other angulations, statistically significant stress differences were noted for miniscrews placed at 45° compared to 60° and 90° for all stress types analyzed (MaxPS $p=0.012$, Min PS $p=0.011$, vonM $p=0.014$). Stresses were localized to the second premolar/first molar area immediately around the implant/bone interface (Figure 12).

Greatest peak MaxPS on the miniscrew itself was noted when angle placement was 45° for all three miniscrew material types (89.3MPa - Ti, 82.99MPa - Comp, 82.75MPa - SS). Lowest peak MaxPS is found at the 60° placement angle for Ti (40.31MPa) and SS (47.53MPa) and at 90° for Comp (27.91MPa). Peak MaxPS in cortical bone is greatest at 45° angulation for Ti (17.93MPa) and Comp (39.94MPa) miniscrews, and 90° for SS (14.96MPa) miniscrews. There is no significant difference in cortical bone stress at any angulation. There is a noticeable ($p=0.052$) difference between material types with Comp miniscrews having a higher average maximum principle stress than Ti or SS (Table 4). In each angulation, the location of greatest maximum principle stress is located at the distal aspect of the miniscrew/cortical bone interface. Comparison of stress on cortical bone with a Ti miniscrew at 45°, 60° and 90° is illustrated in Figure 13.

Peak MinPS is lowest on the miniscrew at 60° for Ti and SS TADs, and similar at 60° and 90° for Comp. MinPS is greatest at 45° for all three materials. Peak MinPS is approximately the same in cortical bone for all three miniscrew materials at 90° (range -10.29 to -11/93MPa) but at 45° and 60° MinPS in cortical bone is higher for Comp than Ti or SS (-33.26 & -29.83MPa respectively for Comp vs. -11.68/-10.05MPa & -16.23/-12.62MPa for Ti/SS). When comparing the MinMS pattern generated for 45°, 60°, and 90° angulations, the Comp miniscrew does not mimic the Ti or SS pattern. Rather, MinMS is substantially lower in both miniscrew and bone at 90° than Ti or SS (Table 5).

Peak vonMS was lowest on the miniscrew at 60° for all three miniscrew materials relative to the other angulations. As with MinPS, the vonMS for the Comp miniscrew differs from the Ti and SS pattern generated for 45°, 60°, and 90° angulations and is substantially lower in both miniscrew and bone at 90° than Ti or SS (Table 6).

DISCUSSION

One of the primary areas of interest in the present study related to the construction of a human model that was both realistic and of sufficient detail to obtain results that may be of value clinically. Comparison of the results in this study to those of other orthodontic studies using FEA is made difficult due to many differences between the model created in this study and that used in other studies. Many of the studies available in the literature do not models anatomical settings in humans,(1, 2) are not 3-dimensional,(26) or do not demonstrate comparably detailed resolution.(9) One study which modeled a scenario similar to this study in detail was published by Cattaneo et al. They evaluated orthodontic tooth movement and resulting periodontal stresses, with particular attention given to the PDL and surrounding alveolar bone. (5) The model in their study was three-dimensional and similar in detail to the model used in this study, although their model consisted of only two teeth and adjacent bony structures. Furthermore, Cattaneo et al simulated both linear and non-linear PDL mechanical properties, a focus not often evaluated in FE studies. In a different study by Motoyoshi, peak bone vonMS in their model was reported to be between 4-33MPa. (1) These stress levels were within the same range as peak vonMS levels in the current study (9.43-15.13MPa) However, Motoyoshi used a 2N (~203gm) force applied obliquely at 45°, different in both magnitude and orientation from that in this study.

A second area of focus in the present study was the effect of miniscrew placement angulation on stress production. Placing miniscrews at an angle to the cortical bone surface may potentially increase the amount of bone the TAD interfaces with, thereby increasing the stability of the miniscrew and decreasing the likelihood of failure. Deguchi et al used CT imaging to measure cortical bone thickness at 30°, 45° and 90° angulations in 10 patients.(27) They found significantly greater cortical thickness in 30° over 45° and 90°, and then in 45° over 90°. Although increasingly more acute angulation may increase the potential for miniscrew retention due to increased cortical bone thickness, too acute of an angle may also result in miniscrew slippage along the cortical bone surface during insertion (16), especially if a guide hole or pilot hole is not used.

In the present study, direct comparison of peak stresses generated in this model to reported stresses in maxillary bone were difficult due to the lack of published data for maxillary cortical bone. Some data from the orthopedic literature may allow for comparison of peak stresses in this study to other bones such as femurs. The highest peak cortical bone stress in this study was also at 45° with a Comp miniscrew (39.94MPa) which is well below the lowest reported tensile strength in cortical bone of ~100MPa.(28, 29) In comparing compression values in the present study to those of Burstein, the peak compression (MinPS) generated in cortical bone was 33.26MPa by the Comp miniscrew at 45°, well below the 120MPa yield stress reported. Looking further, the compressive values of Ti miniscrew (i.e. those being used in clinical practice today) are even lower than those of Comp, with peak MinPS being -16.23 MPa at 60°. When evaluating peak stress in the miniscrew with any angulation/material combination, the vonMS value in the 45° Comp miniscrew model (110.09MPa) is far below the >800MPa tensile strength of titanium.(30) Using the engineering safety factor equation $\sigma_w = \sigma_y / k$ allow comparison of stress levels generated in this study with the peak levels reported. In this study, the resultant stress in all models fell several orders of magnitude below the stress levels of bone and miniscrew.

From the results of the present study, an angulation of 60° is more favorable than either 45° or 90° for all three stress types generated relative to the stress on the miniscrew. Conversely, all three stress types have levels at 60° which are comparable to 45° and 90° in bone, so varied angulation within the range evaluated in the present study may not have a marked effect on the bone. However, miniscrews at 60° do have slightly higher MinPS (compressive) values than either 45° or 90°, which could have an effect on the rate or extent of biological activity and remodeling.

A third area of focus in the present study was the effect of using different miniscrew materials by comparing stress levels generated by currently-used titanium miniscrews with rarely- or unused stainless steel and composite miniscrews. Although no studies were found that compare Ti and/or SS miniscrews to Comp miniscrews, one published study compared some of the mechanical properties of Ti and SS miniscrews.(31) Carano et al reported that

Ti and SS miniscrews could both safely be used as skeletal anchorage, and that Ti and SS miniscrew bending is >0.02mm at 1.471N (150gm)…the load applied in the present study. There was deformation of >0.01mm noted in the present study. However, the data in the study by Carano et al was otherwise not comparable to that in the present study.

There are no studies available which compare Ti and SS stresses or stress pattern generation. Therefore, to evaluate stresses generated from the use of Comp to Ti and SS in the present study, the modulus of the miniscrew in each Ti model was varied to reflect that of SS and Comp, with a subsequent test at each angulation. The fact that the general stress pattern for Comp is dissimilar to Ti and SS when the miniscrew angulation is changed may be of clinical importance (see tables 4-6). Because Comp has a much lower modulus than Ti or SS, it may be that stress shielding does not happen as much in Comp miniscrews, and therefore more stress is transferred to the adjacent cortical bone in both compression and tension scenarios. As a result of increased compressive and tensile forces, especially in the 45° model, biological activity related to remodeling may be increased relative to other models with lower stress levels, and therefore have a more significant impact on long-term miniscrew stability and success. Another potential undesirable effect of using Comp miniscrews in place of either Ti or SS is the increased deformation and distortion that is inherent due to the decreased modulus of composite relative to titanium or steel (Figures 19 & 20). Mechanical or design improvements need to be made to allow for Comp miniscrews to be a viable alternative in clinical practice.

One area of interest noted in the present study that is undiscussed in any other study was the MinPS that was generated on the mesial surface of the lamina dura of the upper left first molar (Figure 21). The advantage the computer model that was used in this study is that trabecular bone is able to be removed from the model so that stress patterns on the UL6 may be visualized. Although the stress levels seen were moderate relative to peak MinPS levels noted at other locations, it is possible that biological remodeling could be altered in this part of the lamina dura. Of additional interest is the relative magnitude and direction of both tensile and compressive stresses visualized in this area through vector plotting (Figures 22 & 23). Liou et al(32) reported that when miniscrews were used for en masse distalization of anterior teeth, the tip of the miniscrew moved distally in some patients in their study. Although measurements were made on cephalometric radiographs, and the magnitude of movement that was reported to be 1mm or less in each patient, it is possible that the areas of compression seen in the lamina dura of the present study are related to the miniscrew tip even though the miniscrews Liou et al used were 17mm in length – longer than those modeled in the present study. Consequently, the amount of miniscrew tip noted would likely be less with shorter miniscrews.

In situations such as those modeled in this study, the goal of maximum retraction of maxillary anterior teeth and minimization of anchorage loss is a common rational for the use of TADs clinically. Multiple studies have reported that miniscrews result in significantly less anchorage loss than other conventional anchorage maintenance methods in patients with normal facial patterns.(33-35) Miniscrew use also resulted in significantly less anchorage loss in normal/hypodivergent patients than in the same type of patients where other methods of anchorage are used.(36)

Numerous studies have been published that employ FEA to predict orthodontic changes, including tooth loading(37), tooth movement(4, 38, 39), skeletal changes(12, 40), and biomechanical effects.(41) However, very few FE studies focus on miniscrews, and none specifically evaluate the effect of changed miniscrew angulation in a model of the level of detail presented in the present study.

While miniscrews have been placed in many intraoral locations, one region that remains frequently used and is easy to access is the posterior buccal maxilla. For this placement location, the reported angle of insertion varies from 30^o to 90^o relative to the long axis of the alveolus.(18, 19, 42) Liou and Kravitz separately postulated that insertion angle of 30^o to 70^o from the occlusal plane (i.e. 20^o to 60^o to the cortical bone long axis) is optimal to allow sufficient cortical bone engagement.(16, 17). However, no studies to date have evaluated the effects of miniscrew placement angulation change and its relationship to failure resulting from stress levels generated. Measuring stress generated with in vivo methods at different miniscrew placement angulation is unfeasible. One study by Pickard et al(3) used shear tests to evaluate pull-out strength of miniscrews placed at 45^o and 90^o in human cadaver mandibles. They found that miniscrews angled at 45^o toward the line of force application (i.e. horizontally angled in a scenario such as that in the present study) had the highest force at failure; however, in clinical practice the operator may be limited in his or her ability to place a miniscrew between upper teeth at a 45^o angle horizontally due to root position.

CONCLUSIONS

1) Because peak MPS levels generated did not exceed the UTS of cortical bone in any model after accounting for the safety factor, variation in miniscrew angulation or material (within the range simulated) is not sufficient alone to cause failure of the miniscrew due to forces generated. This is given the current assumptions and parameters of this model.

2) Other placement factors should be given consideration during placement of miniscrews. If these variables indicate appropriateness of miniscrew placement, angulation may be varied to facilitate treatment objectives, ease of placement, or mechanics.

3) Currently used titanium miniscrews produce stress patterns and magnitudes similar to those of stainless steel miniscrews. Composite miniscrews, however, result in higher stress values in bone than either Ti or SS, and may not be as suitable for clinical use.

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Figure 1: Two different miniscrew designs: Mondeal (left) and IMTEC (right)

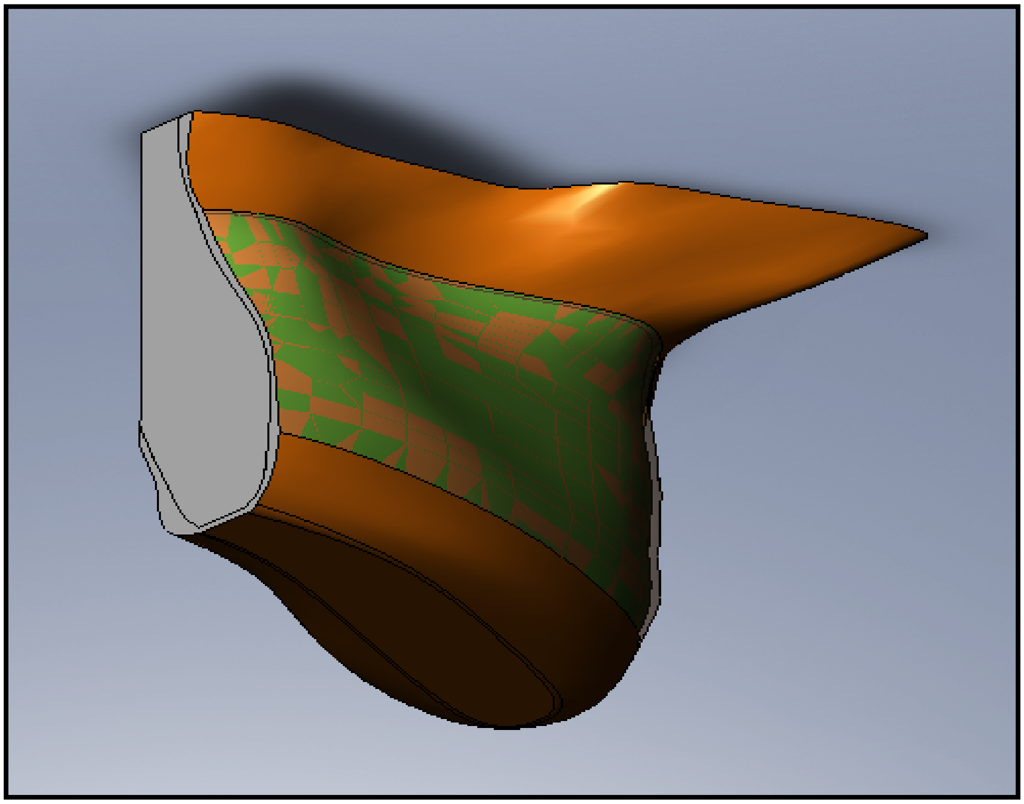


Figure 2: Maxillary model boundaries

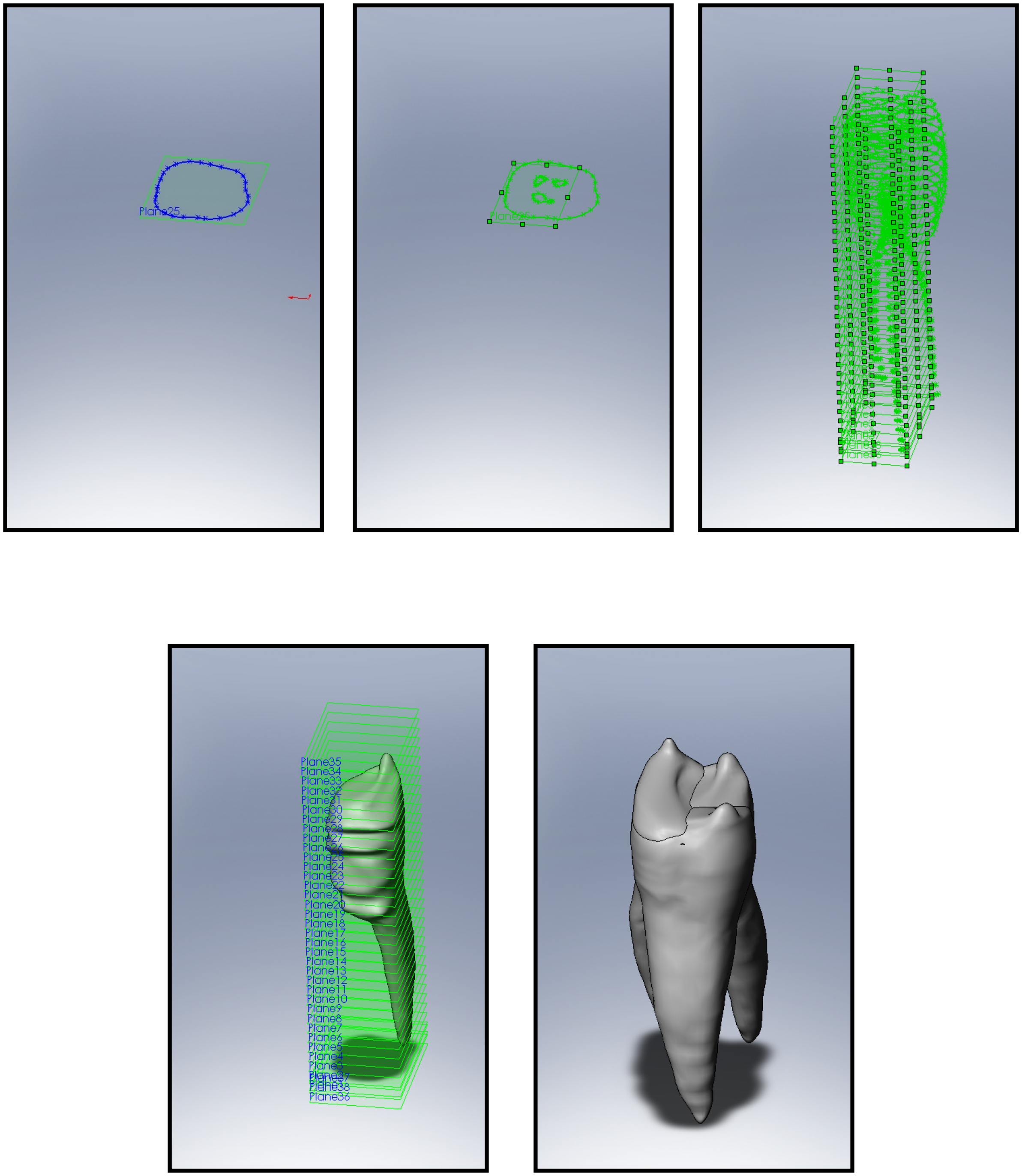


Figure 3: Solidworks construction of maxillary left first molar (UL6) dentin

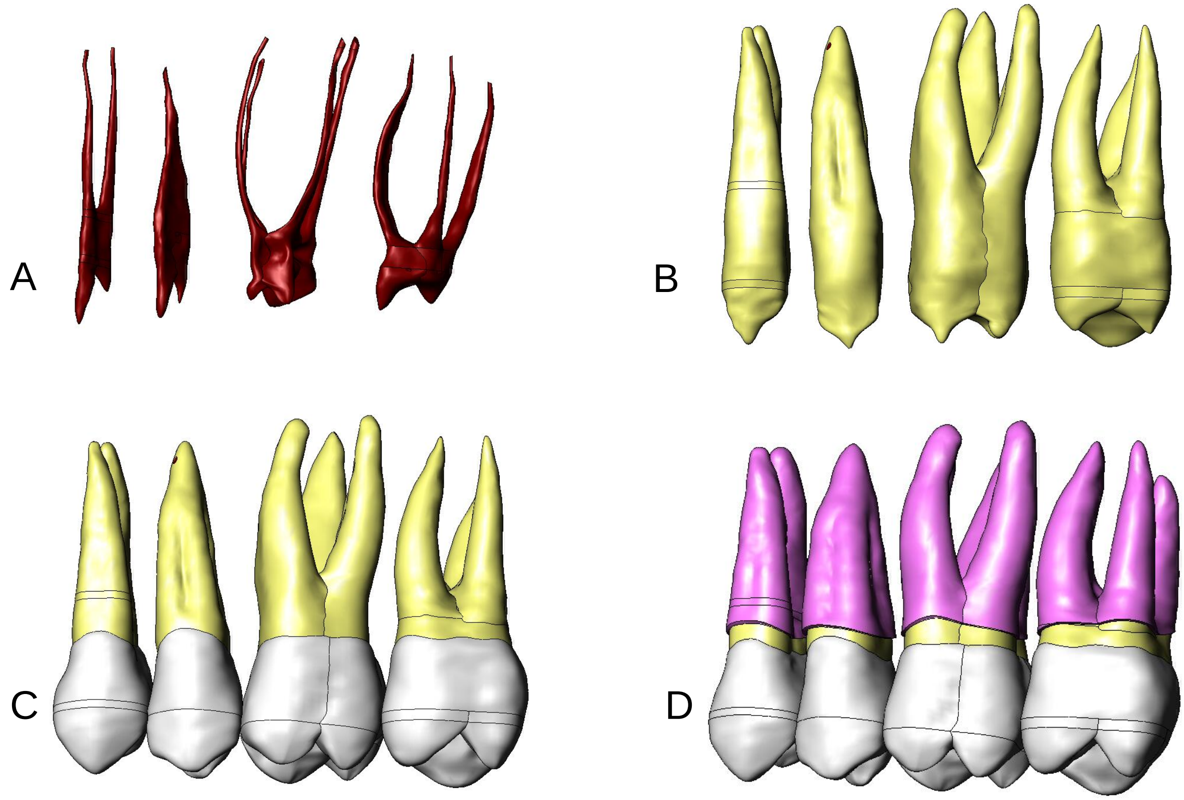


Figure 4: Views of sequential construction of teeth from (A) pulp *(angled view)*; (B) addition of dentin; (C) addition of enamel; (D) addition of PDL *(angled view)*

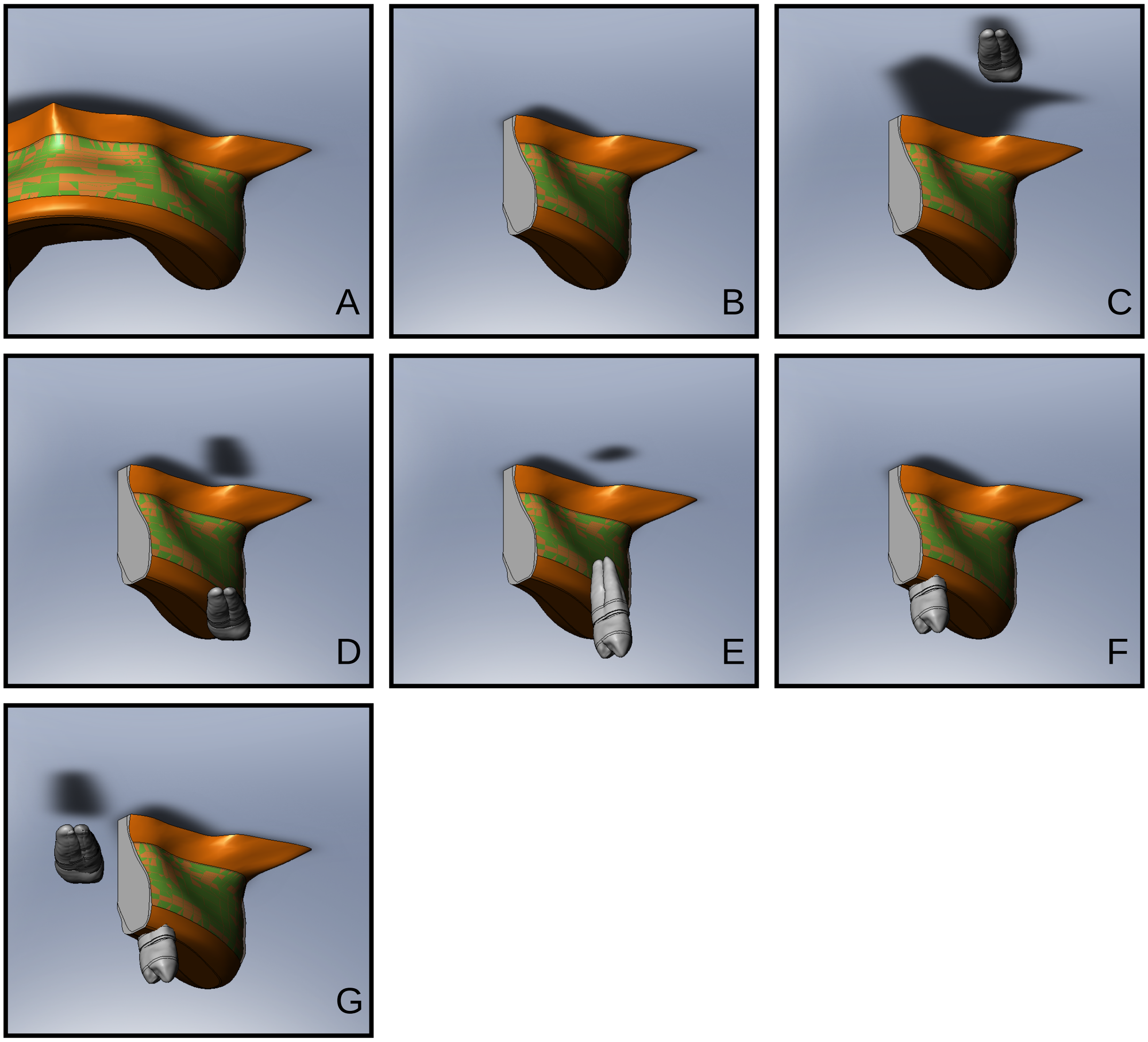


Figure 5– Maxilla construction with addition of teeth by orienting rotationally then locationally

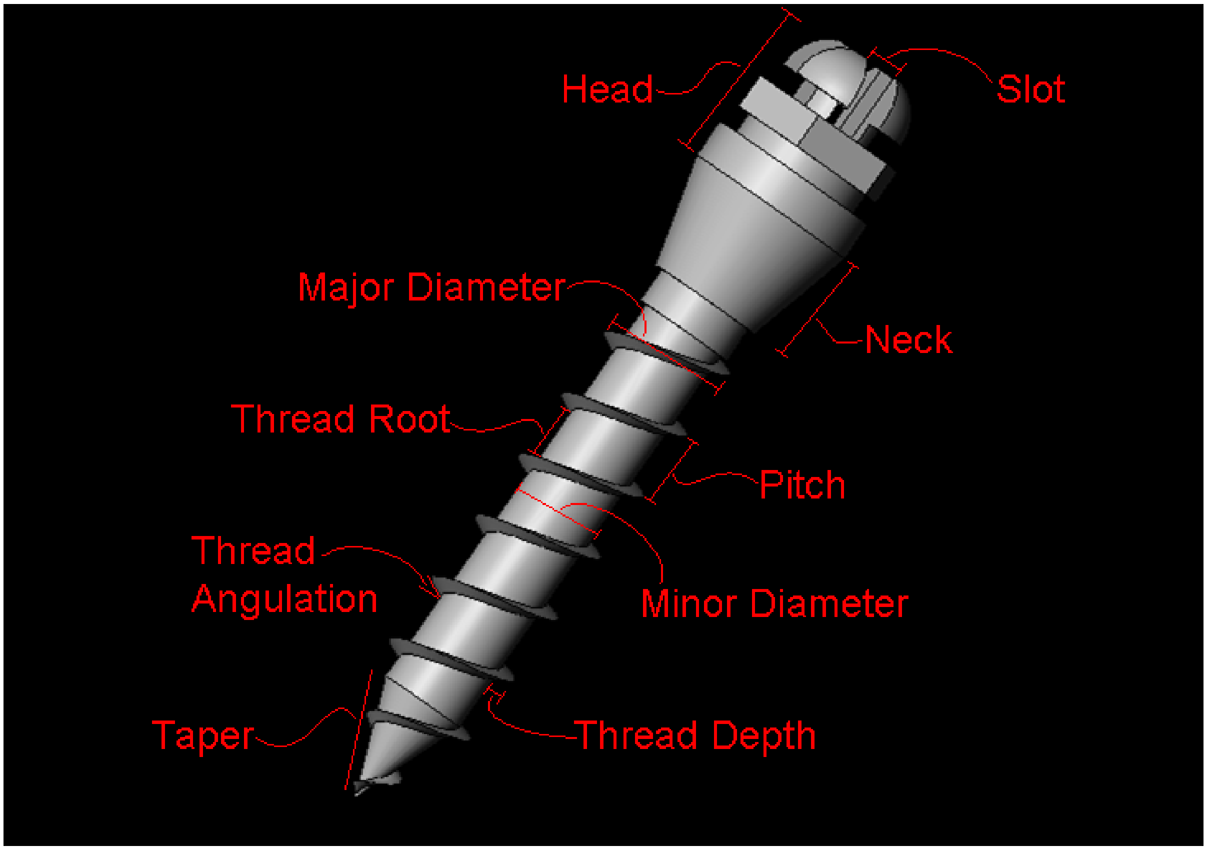


Figure 6: Orthodontic miniscrew specifications (TOMAS miniscrew depicted)

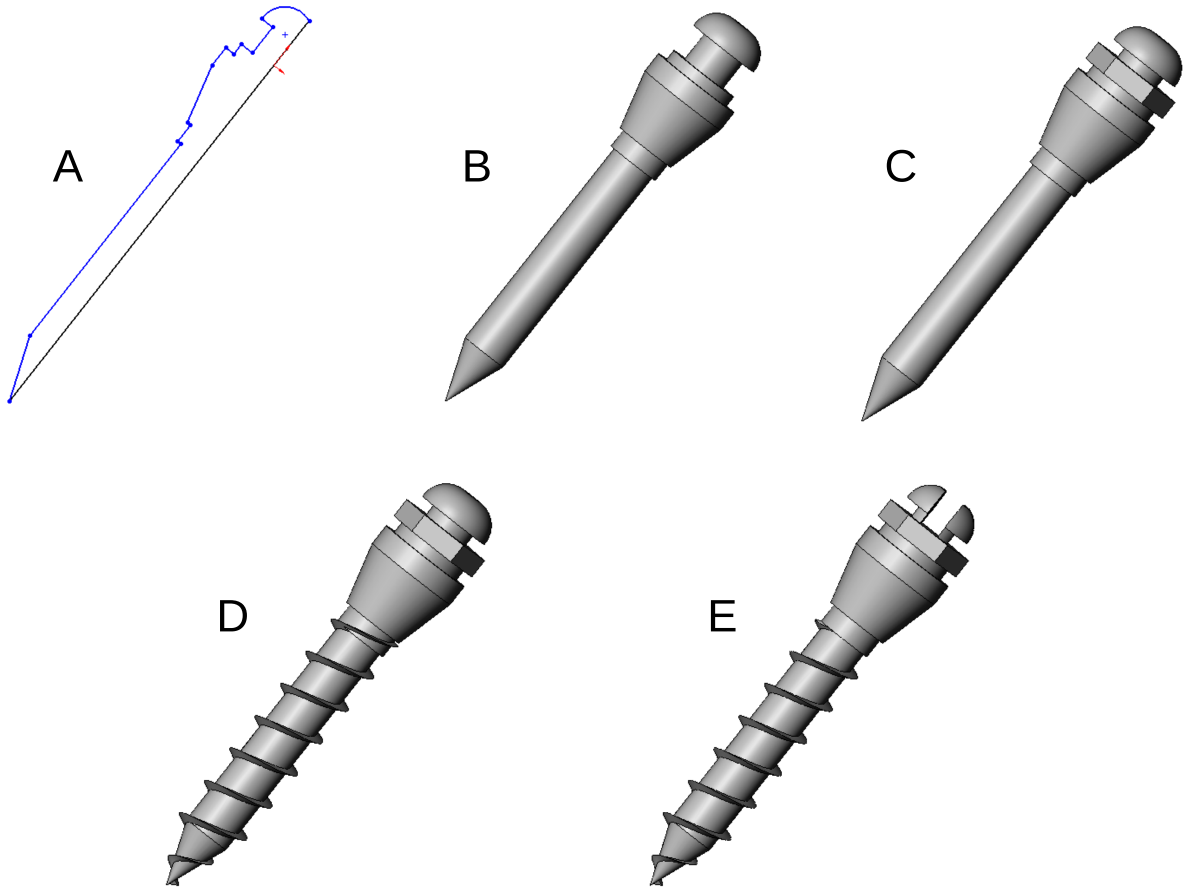


Figure 7: Solidworks construction of the TOMAS miniscrew. (A) Initial hemi-sectioned outline; (B) 3D revolution of outline; (C) Addition of hexagonal head segment; (D) Thread addition via helical sweep function; (E) Removal of head segment material and excess thread to create final miniscrew

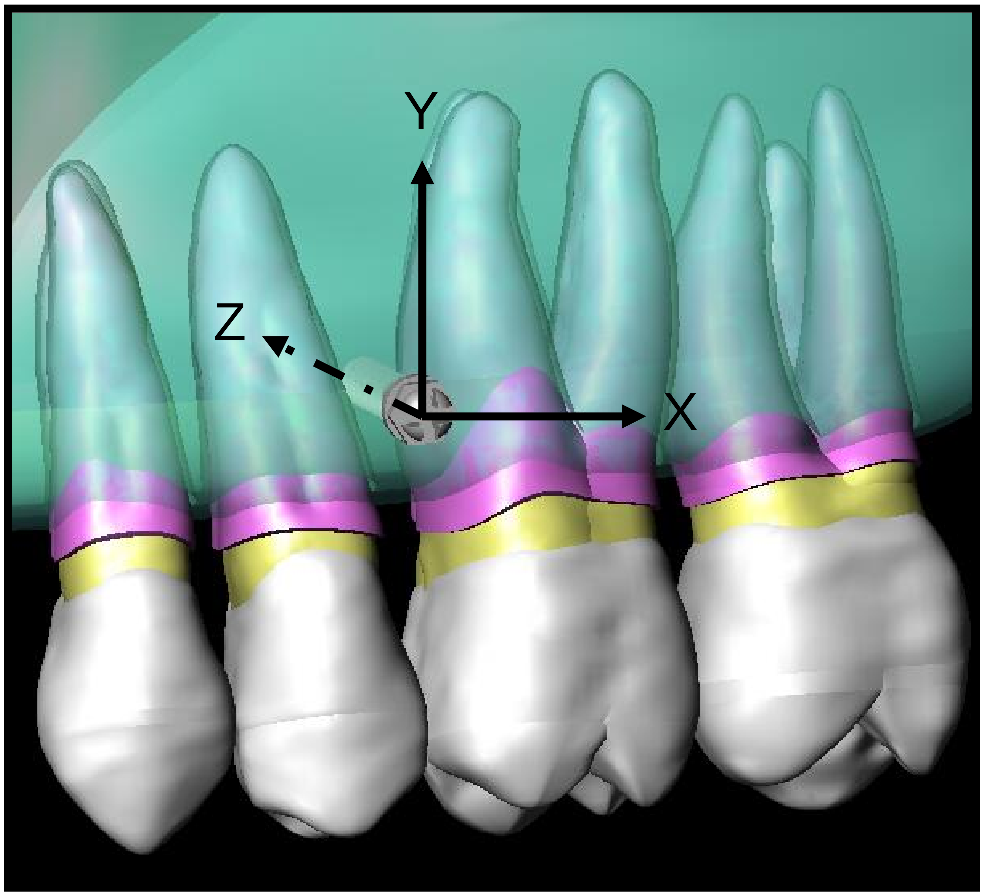


Figure 8: Coordinate planes depicting vectors of potential angulation change: X = mesial/distal movement of miniscrew head or tip; Y = occlusal/apical movement of miniscrew head or tip; Z = in/out movement relative to cortical bone surface

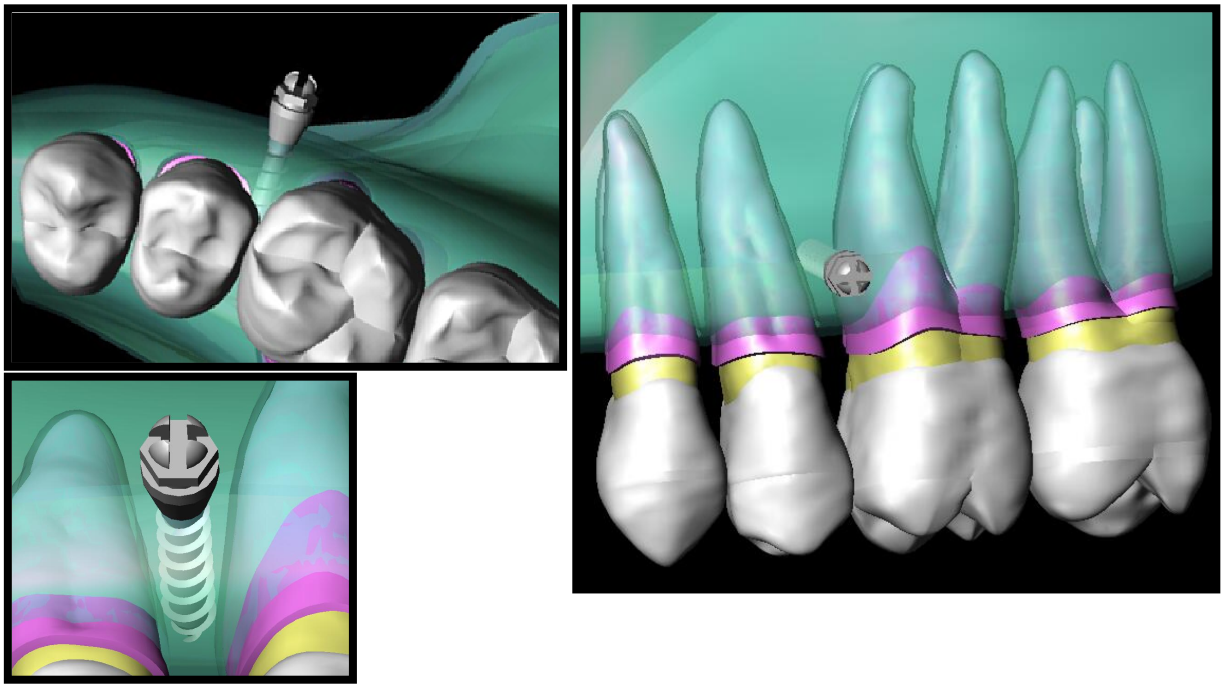


Figure 9: Views of miniscrew placement (90° relative to cortical bone) between upper left 2nd premolar and 1st molar on buccal ~2-3mm apical to alveolar crest.

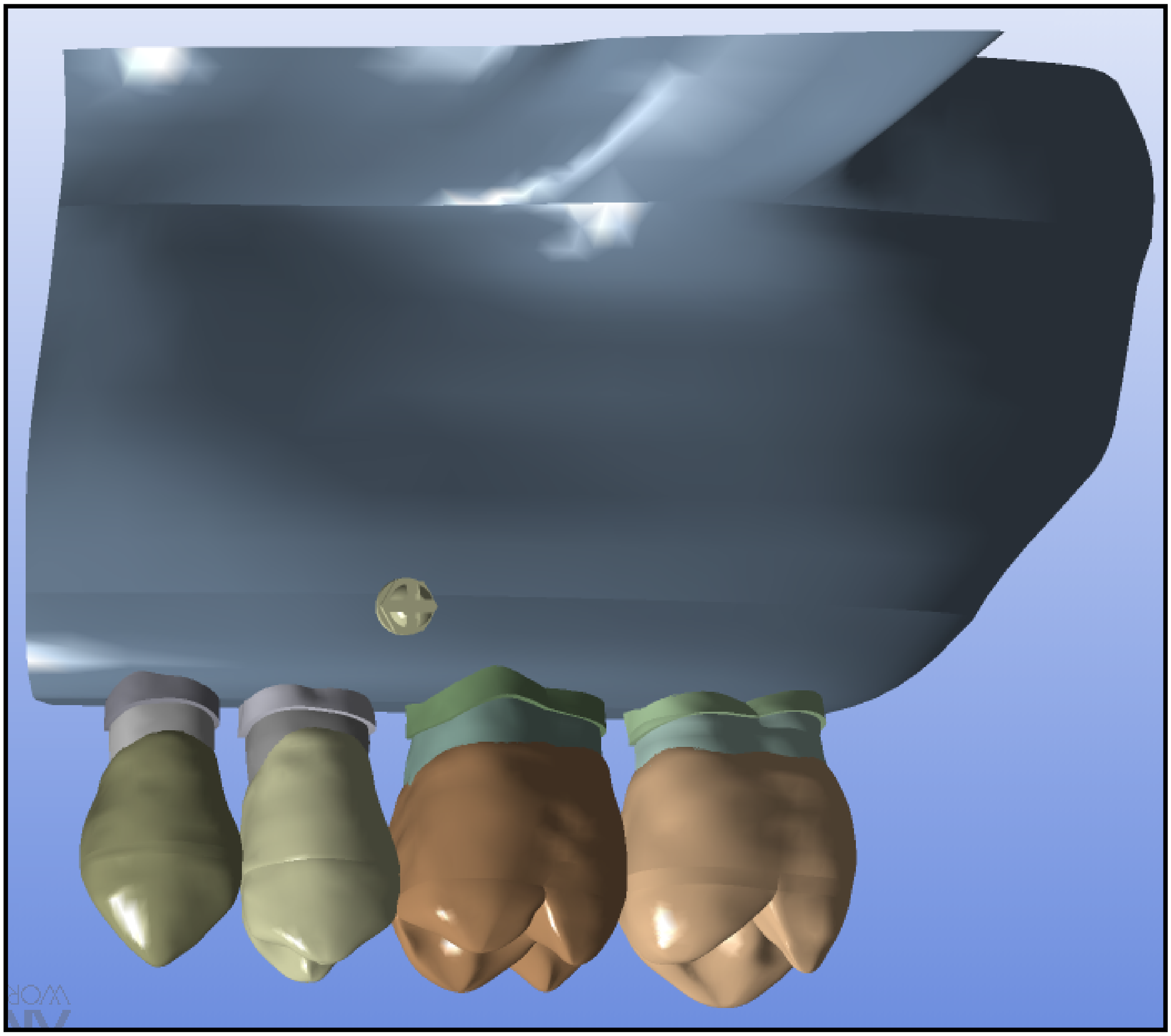


Figure 10: Completed maxillary model (Solidworks image)

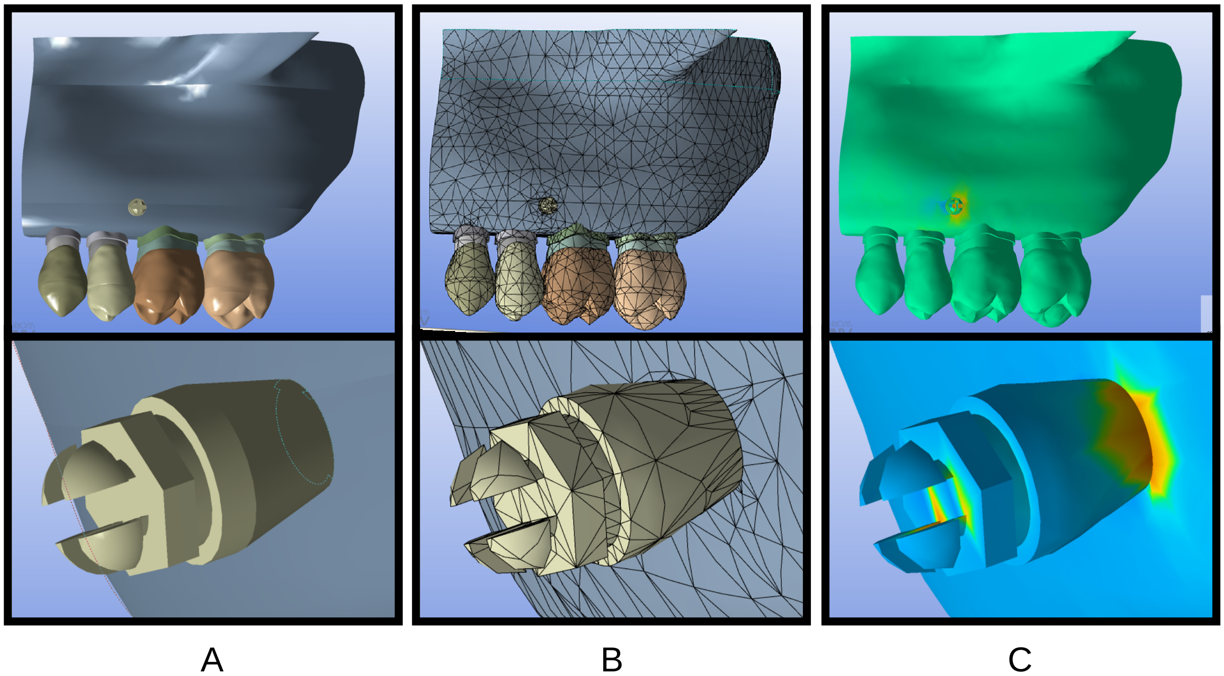


Figure 11: Reconstructed complete maxillary and miniscrew models (A); models with mesh overlay (B); models after force application solved to depict tress generation (C)

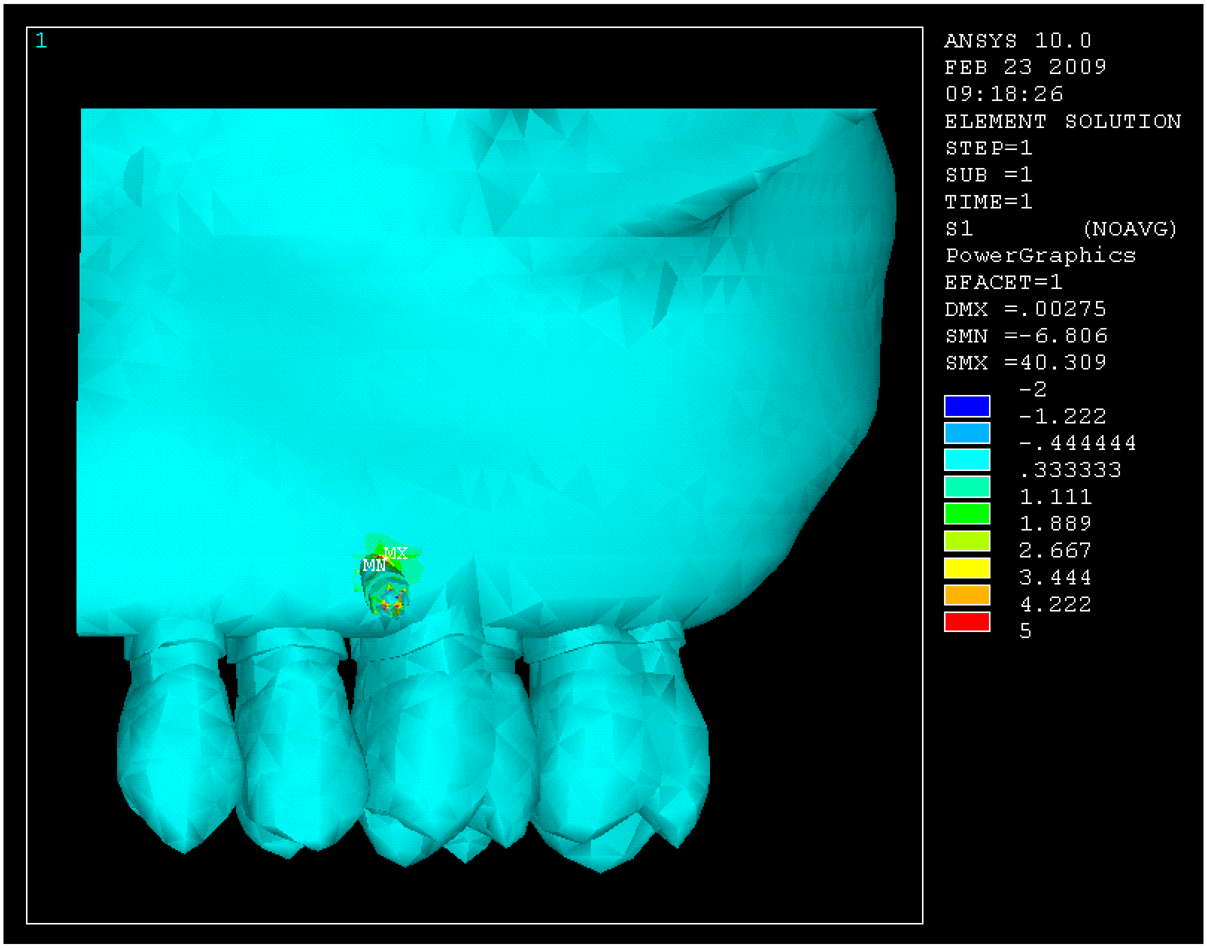


Figure 12: Stress Analysis Results, depicted both numerically and visually

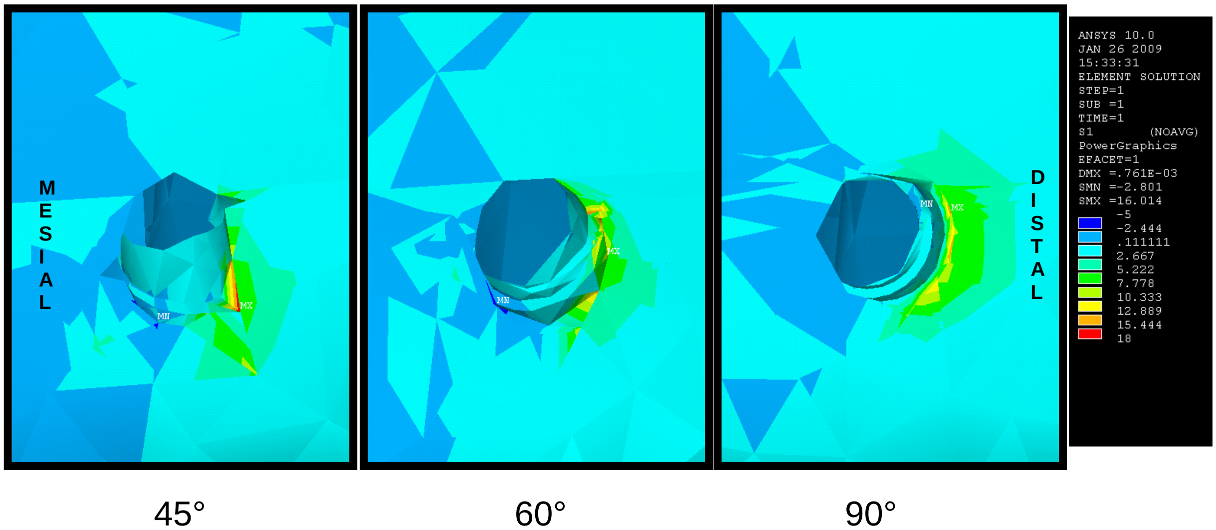
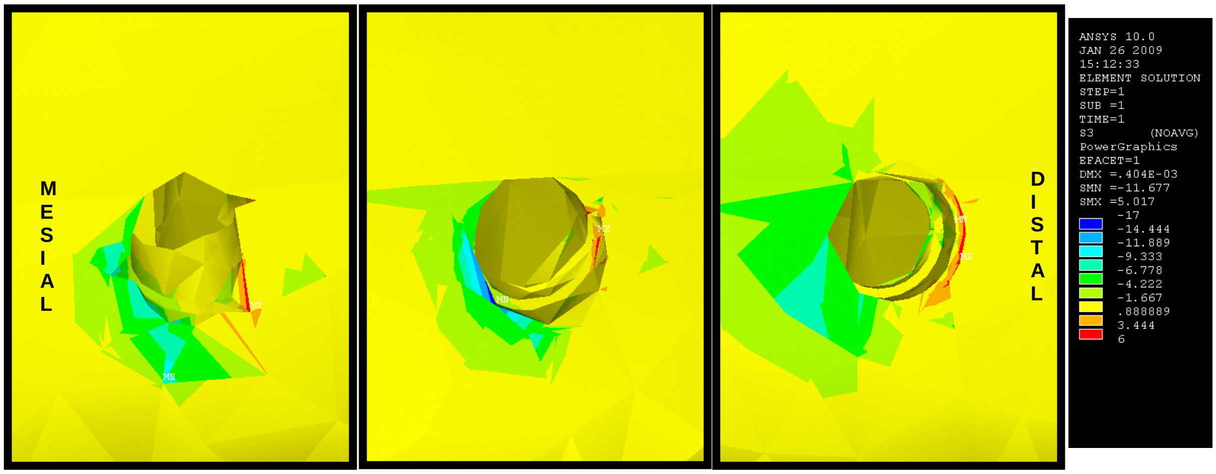
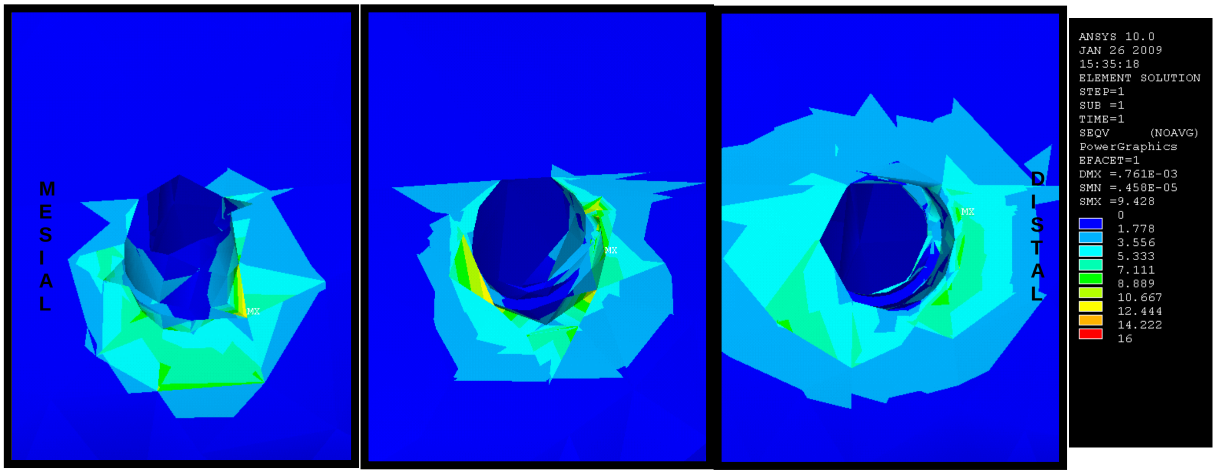


Figure 13: Maximum (1st) Principle Stress patterns generated in cortical bone using TOMAS Titanium miniscrews



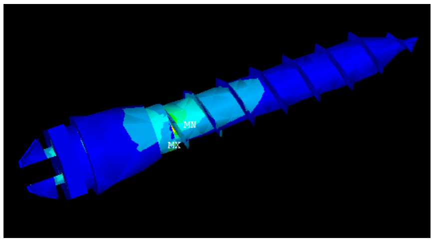
45° 60° 90°

Figure 14: Minimum (3rd) Principle Stress patterns generated in cortical bone using TOMAS Titanium miniscrews

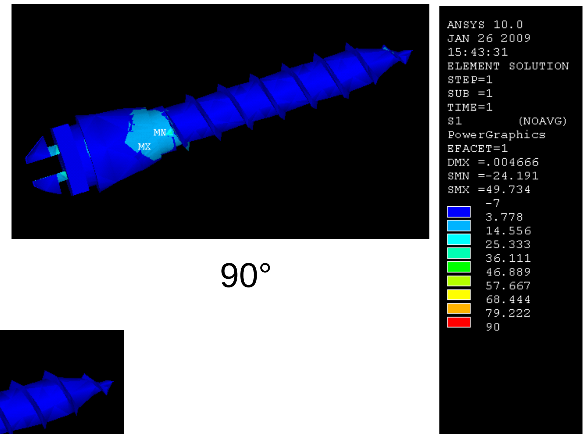


45° 60° 90°

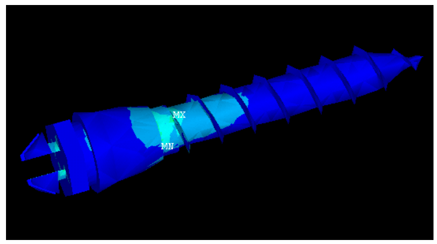
Figure 15: von Mises (Equivalent) Stress patterns generated in cortical bone using TOMAS Titanium miniscrews



45°



90°



60°

Figure 16: Maximum (1st) Principle Stress patterns generated on TOMAS miniscrew (viewed from distal)

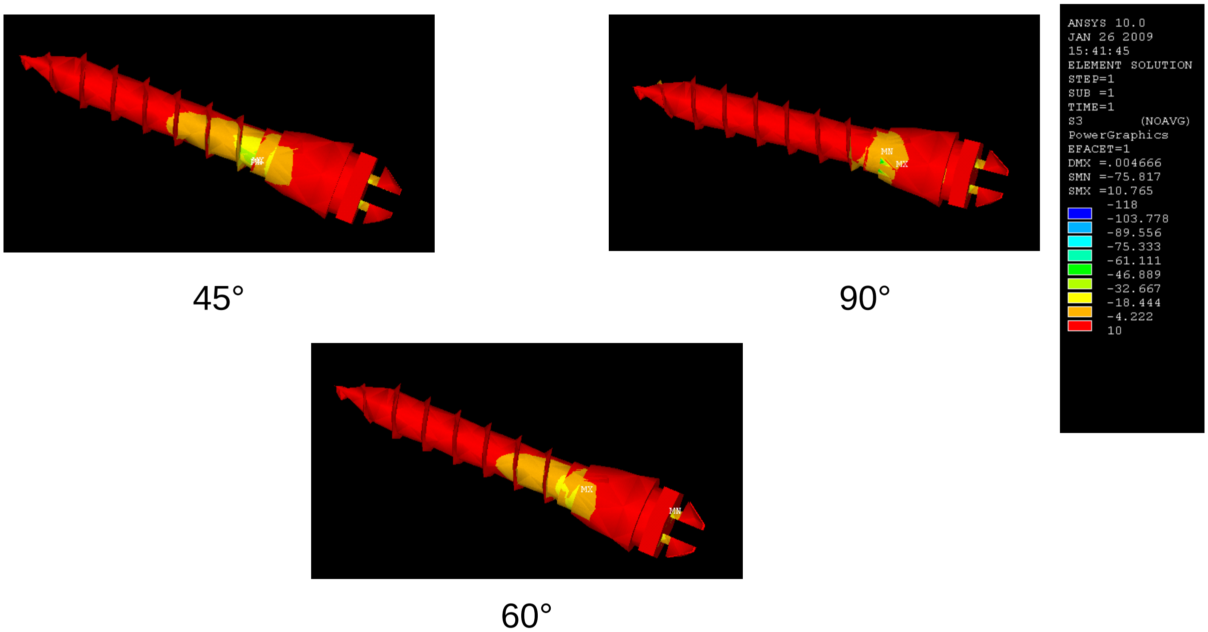


Figure 17: Minimum (3rd) Principle Stress patterns generated on TOMAS miniscrew (viewed from mesial)

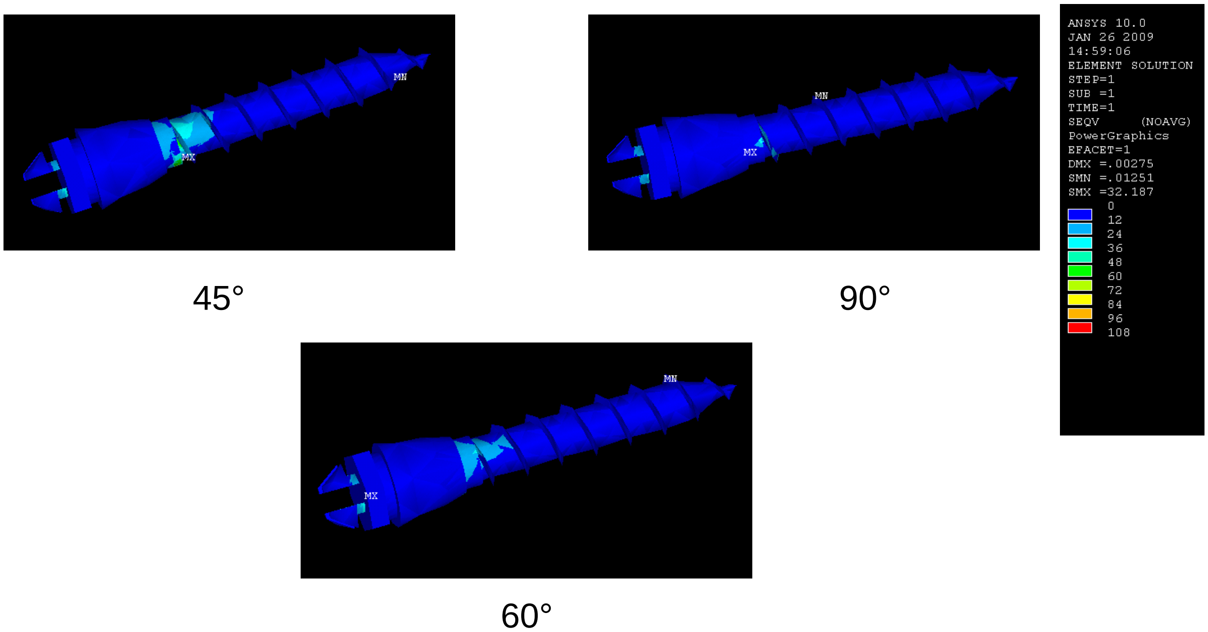


Figure 18: von Mises (equivalent) Stress patterns generated on TOMAS miniscrew (viewed from distal)

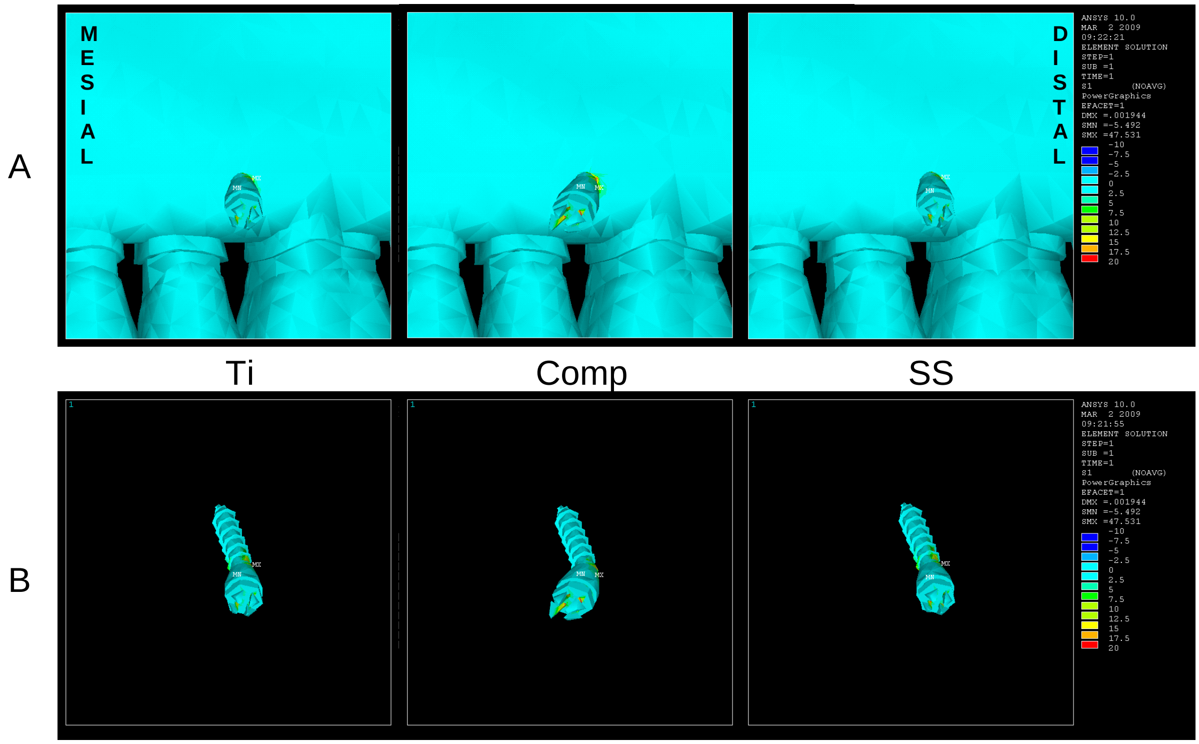


Figure 19: TOMAS miniscrew deformation at 60°: Entire model (A); bone and teeth subtracted (B)

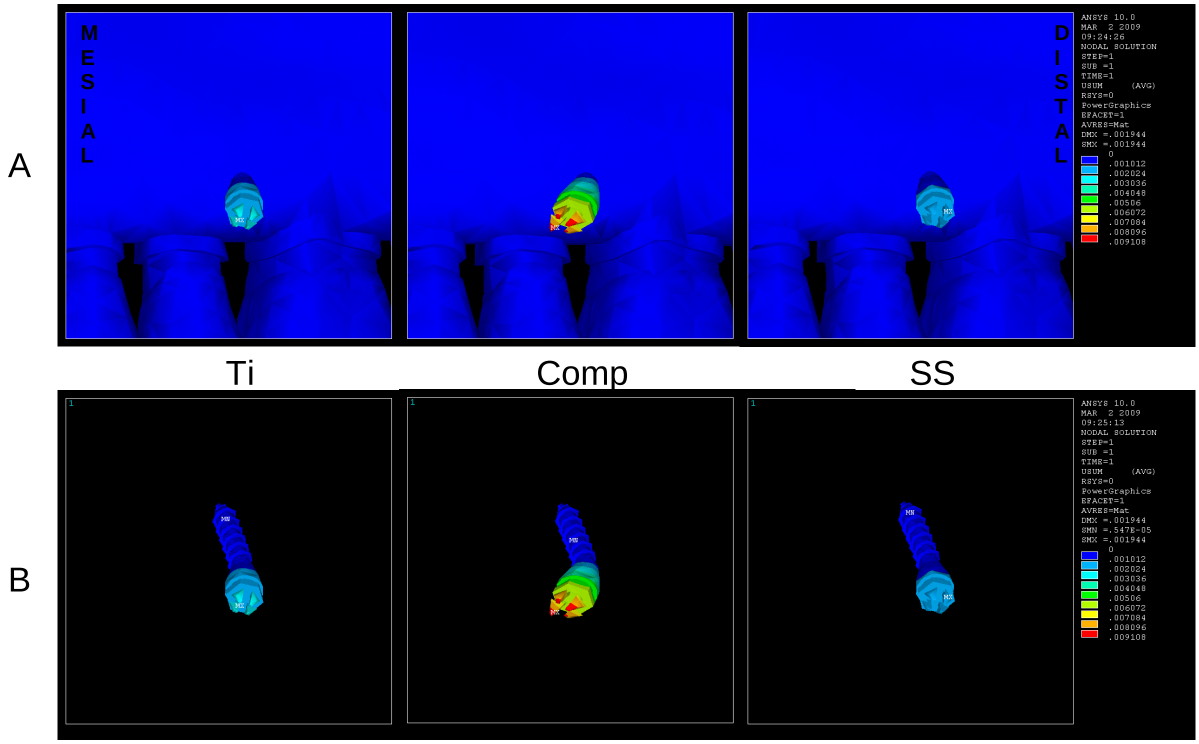
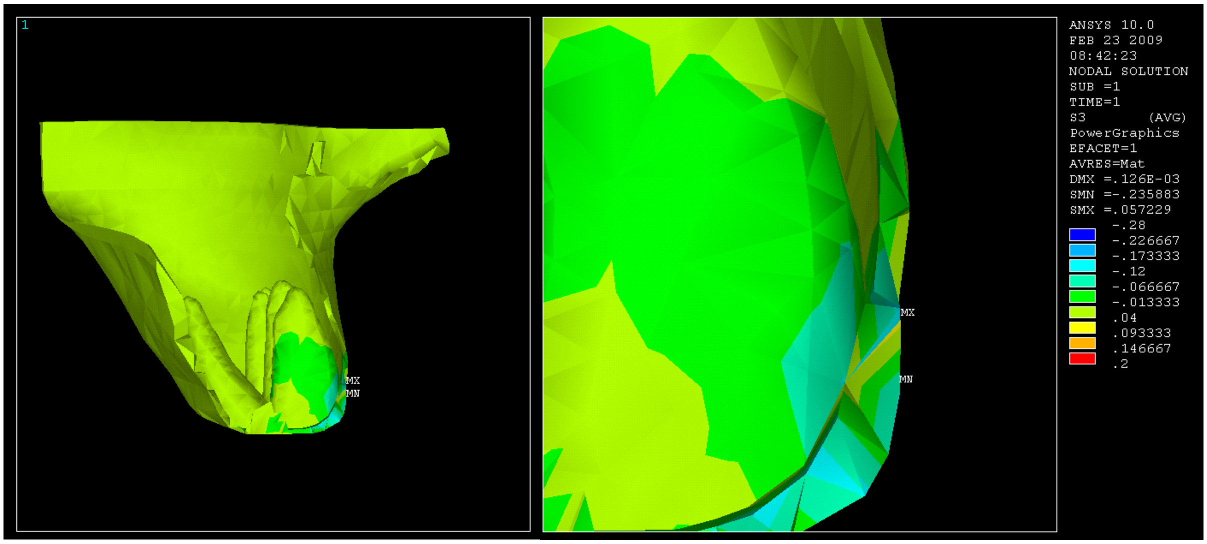


Figure 20:TOMAS miniscrew displacement at 60°: Entire model (A); bone and teeth subtracted (B)



A B

Figure 21: Cut view visualization of MinPS on lamina dura on the mesial of the first molar: whole model (A); up close (B)

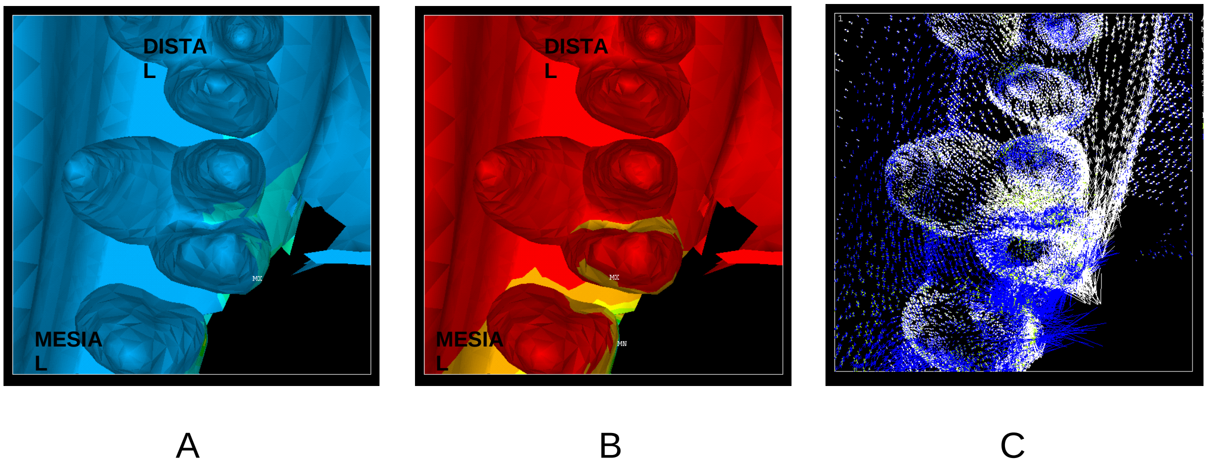
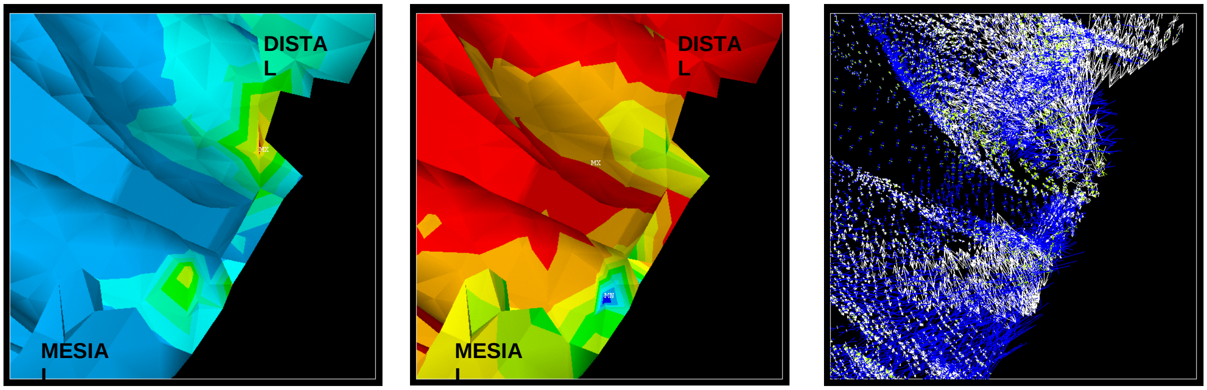
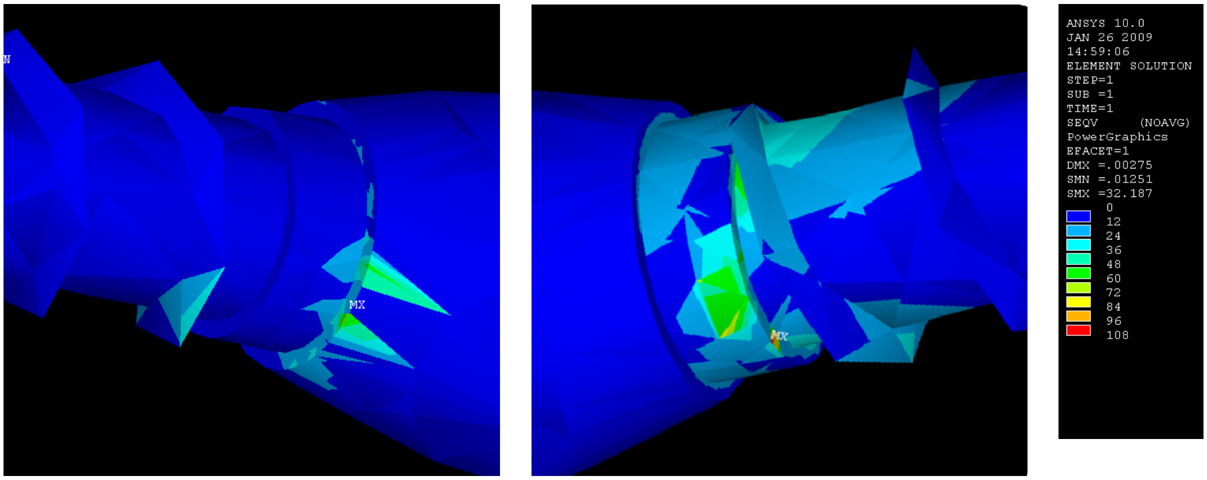


Figure 22: Apical view to visualize stress magnitudes and vectors on lamina dura and bone: Max PS (A); MinPS (B); vectors depicting tensile (white) and compressive (blue) stresses in bone (C)



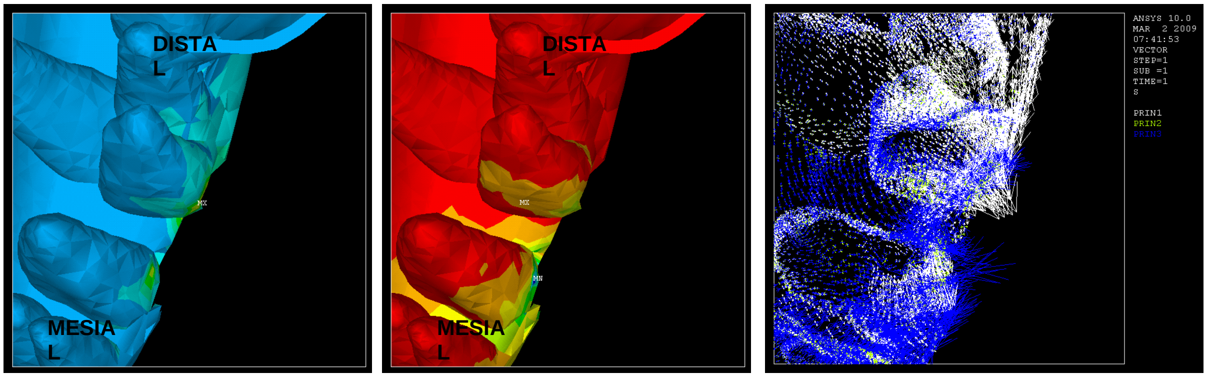
B C

Figure 23: Oblique view to visualize stress magnitudes and vectors on lamina dura and bone: Max PS (A); MinPS (B); vectors depicting tensile (white) and compressive (blue) stresses in bone (C)



Viewed from mesial (90°) Viewed from mesial (45°)

Figure 24: von Mises (equivalent) Stress patterns generated on TOMAS miniscrew viewed up close



A B C

Figure 25: Alternative oblique view to visualize stress magnitudes and vectors on lamina dura and bone: Max PS (A); MinPS (B); vectors depicting tensile (white) and compressive (blue) stresses in bone (C)

Table of Material Properties							
	Cortical	Trabecular	PDL	Dentin	Enamel	Pulp	Miniscrew
Young's Modulus (MPa)	13,700	1370	175	18,000	77,900	175	113,000 (Ti)
Poisson's coefficient	0.3	0.3	0.4	0.3	0.3	0.4	0.3

Table 1: Computer model component material properties

Stress Type	Abbreviation	Variable Measured
Maximum (1st) Principle Stress	MaxPS	tension
Minimum (3rd) Principle Stress	MinPS	compression
von Mises (equivalent) Stress	vonMS	equivalence

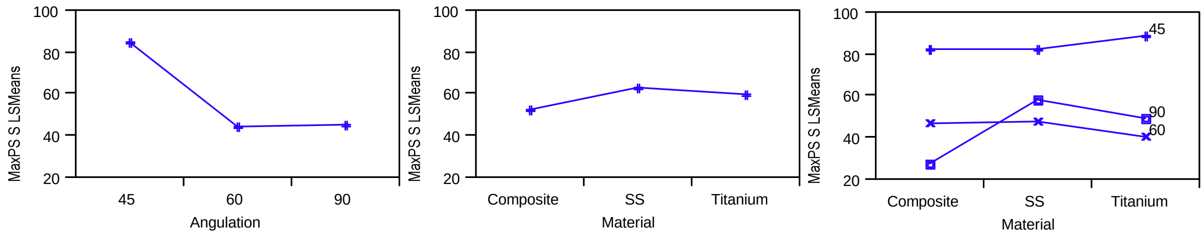
Table 2: Stress types measured

Miniscrew	MaxPS		MinPS		vonMS	
Angulation	**Miniscrew**	**Bone**	**Miniscrew**	**Bone**	**Miniscrew**	**Bone**
45°	**89.3**	**17.93**	**-117.63**	**-11.68**	**107.54**	**12.89**
60°	**40.31**	**16.55**	**-32.18**	**-16.23**	**31.56**	**15.13**
90°	**49.73**	**16.01**	**-75.82**	**-10.29**	**67.24**	**9.43**

Table 3: Stress values (MPa) of miniscrews at varied angulations and with differing materials

MaxPS S.rtf

Source	SS	MS Num	DF Num	F Ratio	Prob>F	
Angulation		3185.14	1592.57	2	16.5909	0.0116
Material	168.303	84.1514	2	0.8767	0.4834	



MaxPS B.rtf

Source	SS	MS Num	DF Num	F Ratio	Prob>F	
Angulation		69.0569	34.5284	2	0.9902	0.4474
Material	470.855	235.427	2	6.7512	0.0522	

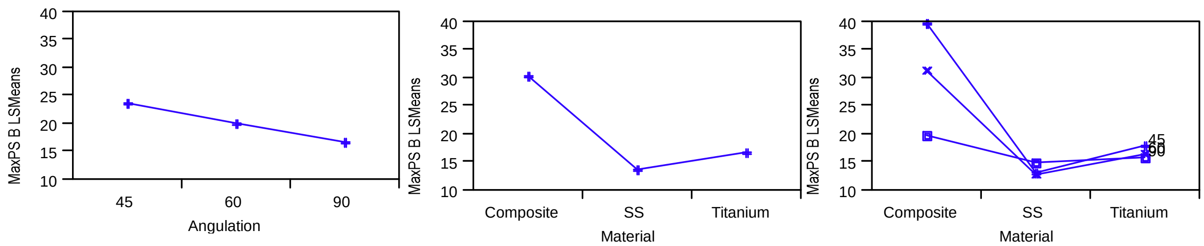
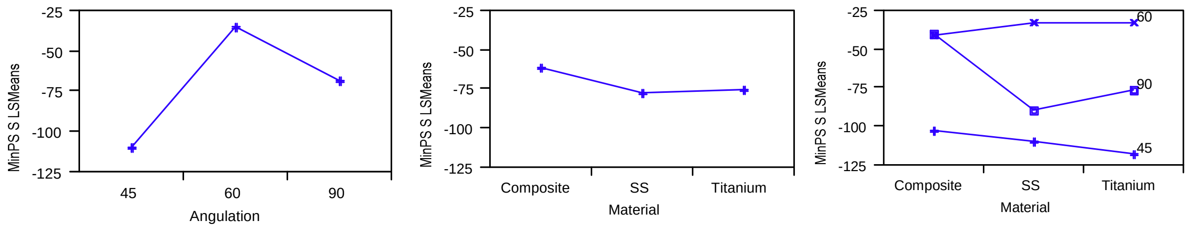


Table 4: MaxPS statistical values

MinPS S.rtf

Source	SS	MS Num	DF Num	F Ratio	Prob>F
Angulation	8435.15	4217.58	2	16.9838	0.0111
Material	467.971	233.985	2	0.9422	0.4621



MinPS B.rtf

Source	SS	MS Num	DF Num	F Ratio	Prob>F
Angulation	123.586	61.7928	2	1.3752	0.3511
Material	322.036	161.018	2	3.5835	0.1283

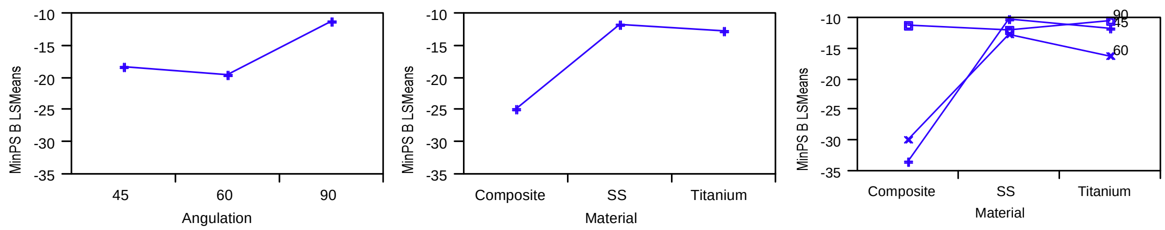
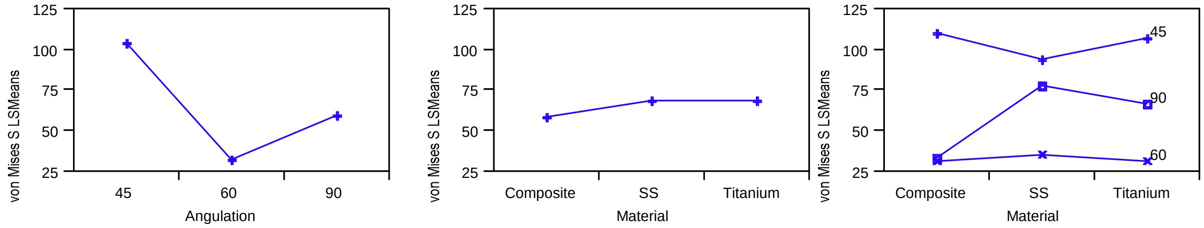


Table 5: MinPS statistical values

von Mises S.rtf

Source	SS	MS Num	DF Num	F Ratio	Prob>F	
Angulation		7728.53	3864.26	2	15.1908	0.0135
Material	223.309	111.655	2	0.4389	0.6725	



von Mises B.rtf

Source	SS	MS Num	DF Num	F Ratio	Prob>F	
Angulation		135.532	67.7661	2	2.0282	0.2465
Material	321.007	160.504	2	4.8038	0.0864	

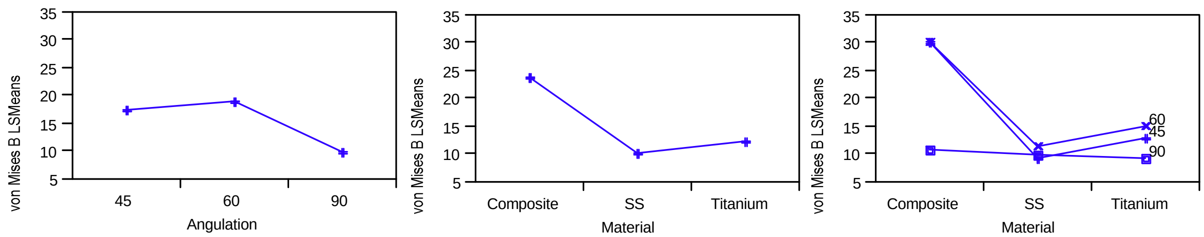


Table 6: vonMS statistical values