# EVALUATION OF VARIABLES AFFECTING DISTRIBUTION OF STRESSES IN ORTHODONTIC MINISCREWS AND IN SURROUNDING CORTICAL/CANCELLOUS BONE-

## A THREE DIMENSIONAL FINITE ELEMENT ANALYSIS

*Dissertation Submitted to*

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*For partial fulfillment of the requirements for the degree of*

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**ORTHODONTICS AND DENTOFACIAL ORTHOPAEDICS**

**THE TAMILNADU DR. M.G.R MEDICAL UNIVERSITY**
**CHENNAI – 600 032**

**2012 – 2015**

# CERTIFICATE

This is to certify that **Dr. G.J. ARUNCHANDER**, Post graduate student **(2012 – 2015)** in the Department of Orthodontics and Dentofacial orthopaedics branch V, Tamil Nadu Government Dental College and Hospital, Chennai – 600 003 has done this dissertation titled ***"EVALUATION OF VARIABLES AFFECTING DISTRIBUTION OF STRESSES IN ORTHODONTIC MINISCREWS AND IN SURROUNDING CORTICAL/CANCELLOUS BONE – A THREE DIMENSIONAL FINITE ELEMENT ANALYSIS"*** under my direct guidance and supervision for partial fulfillment of the M.D.S degree examination in April 2015 as per the regulations laid down by The Tamil Nadu Dr. M.G.R. Medical University, Chennai -600 032 for **M.D.S., Orthodontics and Dentofacial orthopaedics (Branch – V)** degree examination.

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## DECLARATION

I, **Dr. G.J. ARUNCHANDER**, do hereby declare that the dissertation titled ***"EVALUATION OF VARIABLES AFFECTING DISTRIBUTION OF STRESSES IN ORTHODONTIC MINISCREWS AND IN SURROUNDING CORTICAL/CANCELLOUS BONE – A THREE DIMENSIONAL FINITE ELEMENT ANALYSIS"*** was done in the Department of Orthodontics, Tamil Nadu Government Dental College & Hospital, Chennai 600 003. I have utilized the facilities provided in the Government Dental College for the study in partial fulfillment of the requirements for the degree of Master of Dental Surgery in the speciality of Orthodontics and Dentofacial Orthopaedics (Branch V) during the course period **2012-2015** under the conceptualization and guidance of my dissertation guide, **Professor Dr. G. VIMALA MDS.,**

I declare that no part of the dissertation will be utilized for gaining financial assistance for research or other promotions without obtaining prior permission from The Tamil Nadu Government Dental College & Hospital.

I also declare that no part of this work will be published either in the print or electronic media except with those who have been actively involved in this dissertation work and I firmly affirm that the right to preserve or publish this work rests solely with the prior permission of the Principal, Tamil Nadu Government Dental College & Hospital, Chennai 600 003, but with the vested right that I shall be cited as the author(s).

**Signature of the PG student** **Signature of the HOD**

**Signature of the Head of the Institution**

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I offer my heartiest prayers and gratitude to my family members for their selfless blessings.

I seek the blessings of the Almighty without whose benevolence; the Study would not have been possible.

## TRIPARTITE AGREEMENT

This agreement herein after the “Agreement” is entered into on this............... day of December 2014 between the Tamil Nadu Government Dental College and Hospital represented by its **Principal** having address at Tamilnadu Government Dental college and Hospital, Chennai-03, (hereafter referred to as , ’the college’)

And

**Dr. G. VIMALA** aged 46 years working as professor at the college, having residence address at AP 115, 5th Street, AF Block, 11th main road, Anna nagar, Chennai 600040, Tamilnadu (Herein after referred to as the ‘Principal investigator’)

And

**Dr. G.J. ARUNCHANDER** aged 33 years currently studying as postgraduate student in department of Orthodontics in Tamilnadu Government Dental College and Hospital (Herein after referred to as the ‘PG/Research student and co- investigator’).

Whereas the ‘PG/Research student as part of his curriculum undertakes to research **“*EVALUATION OF VARIABLES AFFECTING DISTRIBUTION OF STRESSES IN ORTHODONTIC MINISCREWS AND IN SURROUNDING CORTICAL/CANCELLOUS BONE – A THREE DIMENSIONAL FINITE ELEMENT ANALYSIS*”** for which purpose the PG/Principal investigator shall act as principal investigator and the college shall provide the requisite infrastructure based on availability and also provide facility to the PG/Research student as to the extent possible as a Co-investigator.

Whereas the parties, by this agreement have mutually agreed to the various issues including in particular the copyright and confidentiality issues that arise in this regard.

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guidance from the principal investigator shall be subject to the prior approval, recommendations and comments of the Ethical Committee of the college constituted for this purpose.

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9. If any dispute arises as to the matters related or connected to this agreement herein, it shall be referred to arbitration in accordance with the provisions of the Arbitration and Conciliation Act, 1996.

In witness whereof the parties hereinabove mentioned have on this the day month and year herein above mentioned set their hands to this agreement in the presence of the following two witnesses.

College represented by its

**Principal** **PG Student**

**Witnesses** **Student Guide**

1.

2.

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# INTRODUCTION

Orthodontics is gradually evolving from an opinion based practice to evidence based practice. In contemporary period, it is necessary to have scientific rationale for any treatment modality and the evidence of tissue response to it. Anchorage control is one of the challenges for an orthodontist. Efficient attainment and control of anchorage is fundamental to successful orthodontic and dentofacial orthopedic treatment.[1]

Anchorage, defined as a resistance to unwanted tooth movement, is a prerequisite for the orthodontic treatment of dental and skeletal malocclusions. However, even a small reactive force can cause undesirable movements; it is important to have absolute anchorage to avoid them. Absolute or infinite anchorage is defined as no movement of the anchorage unit (zero anchorage loss) as a consequence to the reaction forces applied to move teeth. Such an anchorage can only be obtained by using ankylosed teeth or dental implants as anchors, both relying on bone to inhibit movement.

Anchorage provided by devices, such as implants or mini- implants fixed to bone, enhances the support to the reactive unit (indirect anchorage) or by fixing the anchor units (direct anchorage), thus facilitating skeletal anchorage.

Mini -implants are generally more widely used because of their lower cost structure, ease of insertion and removal, and versatility of placement. In clinical practice, mini-implants are loaded immediately after insertion and achieving maximum primary stability.[2]

Implants as defined by Boucher are “alloplastic devices which are surgically inserted into or onto the jaw bones”. From the time of 16th century till to date , various materials have been tried as implants which include gold, ivory, tantalum, stainless steel, cobalt chromium, vitreous carbon, vitallium, ceramics and titanium. Among these, titanium is the material of choice for implants today because of its excellent biocompatibility and ability to osseointegrate.

Stress analysis of the end osseous implants is necessary for bone turnover and hence maximum anchorage success. It is virtually impossible to measure stress accurately around mini-implants in vivo. Also, it is difficult to achieve an analytical solution for problems involving complicated geometries such as the maxilla and the mandible, which are exposed to various kinds of loads.

Finite element analysis provides an approximate solution for the response of the 3-dimensional (3D) structures to the applied external loads under certain boundary conditions.[3]

Incorrect loading or overloading may lead to disturbed bone turn-over and consequent implant loss. Since clinical determination of stress and strain distribution in the bone is not possible, an alternative technique should be used.

The finite element method (FEM), which has been successfully applied to the mechanical study of stresses and strains in the field of engineering, makes it practicable to elucidate stresses in the living structures caused by various internal and external forces. FEM offers a viable and non-invasive alternative for analysis of the stress and strain distribution, which is unique because of its ability to model geometrically complex structures. Both two and three dimensional stress analyses have been used to analyze the dental implants.

Many studies have made a comparison between the three dimensional and two dimensional finite element stress analysis.The three dimensional method has been shown to offer a more precise prediction of stress distribution than the two dimensional method. For over 200 years, people have tried to understand the mechanical influence on living bone. In the past two decades, Finite Element Analysis (FEA) has become an increasingly useful tool for the prediction of the effects of stress on the implant and its surrounding bone. The key factor for the success or failure of dental implant is the manner in which the stresses are transferred to the surrounding bone. FEA allows predicting stress distribution in the contact area of the implant with the cortical bone, and around the apex of the implant in the trabecular bone. Unlike prosthodontic implants Osseo integration is not necessary for orthodontic mini implants which allows immediate loading of the mini implants. [4]

The primary stability of the mini implants has been associated with many factors including cortical bone thickness, force applied and angle of the applied force, exposure length, thread pitch, insertion angle and diameter.[5] Hence there is a need to explore the variables related to the stress patterns in the bone adjacent to the different types of implant following orthodontic loading. Knowledge about these factors will enable the clinician to successfully manage anchorage problems in routine clinical practice. Therefore the present study has been undertaken with the following aim.

# AIMS AND OBJECTIVES

## AIM

The Aim of this study is to evaluate the variables affecting distribution of stresses in orthodontic miniscrews and in surrounding cortical/cancellous bone by three dimensional finite element analysis

## OBJECTIVES

1) To evaluate by applying FEM the stress distribution pattern in the bone surrounding a tapered mini implant in response to variables such as cortical bone thickness, force angulation, exposure length, thread pitch, insertion angle and diameter.

2) To evaluate by applying FEM the stress distribution pattern in the bone surrounding a cylindrical mini implant in response to variables such as cortical bone thickness, force angulation, exposure length, thread pitch, insertion angle and diameter.

# REVIEW OF LITERATURE

Voluminous literature has been published on implants. Writings relevant to the present study have been reviewed under the following categories:

1) Evolution of mini implants
2) Micro Computerized tomogram and radiographic study
3) Finite element analysis studies
4) Mechanical pull out strength of miniscrews.

***Evolution of mini-implants:***

***Gainesforth and Higley (1945)***[6] mentioned the use of implant supported anchorage. They used vitallium screws in six dogs. These implants were inserted in the ramal area, immediately loaded and were used to bring about retraction of upper canines. However, all screws were lost within a period of 16-31days. However, all screws were lost within a period of 16-31days. Following this failure to attain stable anchorage, there were no further reports of attempts to use endosseous implants to move teeth.

***Linkow(1970)***[7] about 25 years later used an implant as a replacement for a missing molar. This was then used as an anchor tooth, to which Class II elastics were used to retract the upper anteriors. The upper arch was consolidated using a fixed appliance, while in the lower arch, only the premolar and molar were banded and interconnected using a 0.040” rigid wire.

***Creekmore and Eklund (1983)***[8] published a case report of usage of a vitallium implant for anchorage, while intruding the upper anterior teeth. The vitallium screw was inserted just below the anterior nasal spine. After a healing period of 10 days, an elastic thread was tied from the head of the screw to the archwire. Within one year, 6mm intrusion was demonstrated along with 25° lingual torque.

**Eugene Roberts (1990, 1994)**[9] has done an extensive research relating to usage of retromolar implants for orthodontic anchorage. The first clinical trial was on an adult in whom an atrophic extraction site had to be closed. A special implant was developed that was 3.8mm wide and 6.9mm long, which was placed in the retromolar area. A 0.021” X 0.025” SS wire was used for anchorage from the screw around the premolar bracket. In the initial phases, this wire also aided in leveling. The extraction spaces were closed using forces from buccal as well as lingual arch. The premolar was prevented from moving distally with the help of 0.021” X 0.025” wire acting as an anchorage. The modification in this technique, as suggested by him in 1994, includes the usage of a 0.019” X 0.025” TMA wire. This wire was termed as the anchorage wire.

***Southard et al (1995)***[10] compared the intrusion potential of implants with that of teeth (dental anchors). Titanium implants were placed in extracted 4th premolar area in dogs, followed by a healing period of three months. Then, an intrusive force of 50-60gm via a ‘V’ bend was applied. This was compared with the intrusive potential of teeth on the other side using the same mechanics. No movement of implant was seen at the end of the experiment whereas, on the other side, the tooth acting as the anchor units tipped severely. Therefore, they concluded that implants were definitely superior to the teeth acting as anchor units.

***Block and Hoffman (1995)***[11] introduced Onplant, which is a classic example of a subperiosteal implant. Developed by, it consists of a circular disc, 8-10mm in diameter, with a provision for abutments in the center of the superficial surface. These abutments would enable the Orthodontist to carry out tooth movement against the Onplant. The undersurface of this titanium disc is textured and coated with hydroxyapatite. The hydroxyapatite, being bioactive, helps in stabilization of the implant by improving integration with the bone. The average thickness (height) of the implant is 3mm. Extensive animal studies have been carried out on Onplants. They point out to the fact that Onplants bio-integrate, and can tolerate a maximum force of 161 lbs. Block and Hoffman further suggest that these Onplants could be used not only for dental anchorage (e.g., for retraction of anteriors and distalization of posteriors), but also for orthopedic traction. Human trials are however limited.

Four new systems, which could be grouped under the category of osseous implants, were introduced. Osseous implants are those that are placed in dense bone such as the zygoma, the body and ramus area or the mid-palatal areas.

***Wehrbein (1996)***[12] developed the Orthosystem a titanium screw implant with a diameter of 3.3mm, inserted into the median palate or the retromolar regions of the mandible or the maxilla. The implants are surface treated with sandblasting and acid etching for making the surface rough in order to improve integration. They are available in two sizes of 4mm and 6mm. An 8 week waiting period has been suggested before applying forces onto this implant.

***Umemori and Sugawara (1999)***[13] developed the Skeletal Anchorage System. It essentially consists of titanium miniplates, which are stabilized in the maxilla or the mandible using screws. The recent versions of these miniplates have been modified for attaching orthodontic elastomeric threads or coil springs. Different designs of miniplates are available, and this fact offers some versatility in placing the implants in different sites. The ‘L’ shaped miniplates have been the most commonly used ones, while the ‘T’ shaped ones have been proposed for usage while intruding anterior teeth. The screws used for fixing the miniplate are usually 2-2.5mm in diameter.

***Karcher and Byloff (2000)***[14] introduced Graz implant supported system, consists of a modified titanium miniplate with provision for four miniscrews, and two oval shaped cylinders. This was used mainly as a support for the Nance button of a pendulum appliance in the palate.

***Hugo De Clerck and Geerinckx (2002)***[15] of Belgium, introduced the Zygoma anchor system. It is a curved titanium miniplate with provision for three screws of 2.3mm diameter to offer the necessary stability. The lower end of the miniplate projects outward. It contains a vertical slot for ligatures or other orthodontic attachments. The plate is designed for use in the zygomaticomaxillary buttress area. Placement is identical to that of the Skeletal Anchorage System. These osseous implants were effective in achieving complex tooth movements like molar intrusion. But, they had their own limitations. They needed a fairly complex surgery and, therefore, had to be placed by a surgeon. Secondly, the chances of infection were greater than the screw implants. Their removal was as difficult as their placement.

Interdental implants were developed in the late 1990's. They are endosseous implants, but of smaller diameter, which allows placement in interdental areas. These rely more on mechanical retention than complete osseointegration. The interdental implants are favoured over the retromolar implants due to the following reasons;

1) Placement is very simple and can be done under local anaesthesia.

2) They seem to be equally effective in resisting forces as the larger root form implants.

3) They can be used for bringing about all types of tooth movement.

4) Removal is an uneventful procedure.

***Bousquet et al (1996)***[16] published a case report, demonstrating the use of an impacted titanium post for orthodontic anchorage. The post, 0.7mm in diameter and 6mm long, was made of titanium alloy ($Ti_6Al_4V$). This post, which was impacted in the interdental septum between the upper right 1$^{st}$ molar and the extraction site of the second premolar, was used as anchorage for retraction of the 1$^{st}$ premolar on that side, while the left first premolar was retracted conventionally. The mesial movement of the two first molars was then compared. The right first molar tube was connected to the impacted post with a rigid 0.040” wire. At the end of retraction, it was seen that on the side of post, there was no mesial movement of the molar and complete retraction of the 1st premolar. On the left side, there was marked mesial movement of the 1st molar along with the distal movement of the 1st premolar. This case report showed the feasibility of using a titanium post for anchorage.

***Ryuzo Kanomi (1997)***[17] introduced the Mini-implant. This is a modified surgical miniscrew of 1.2mm diameter and 6-7 mm length, which can be placed interdentally. These implants have been used successfully for anterior intrusion and retraction, and molar intrusion.

***Costa et al (1998)***[18] published a preliminary report on their newly developed mini implant called as Aarhus Anchorage System. The initial design, with an internal Allen wrench-type hole in the head, fractured on removal of the implant. This design was later replaced by a miniscrew with a bracket-like head, which facilitated the insertion of a full-sized wire. Various lengths of transmucosal collar and threaded body are available for individual anatomies, in either 1.5mm or 2mm diameters.

***Park et al (2001)***[19] developed a customized implant system called Micro-implant Anchorage system. These are small diameter implants, which can be placed interdentally either in the buccal sulcus or palatal interdental areas. The screws are available in different lengths and diameters.

The maxillary implants are longer than the mandibular ones owing to the difference in the thickness of cortical bone. The micro-implants are made of titanium. In mobile mucosal areas, such as the buccal aspect in the maxillary arch, it has been suggested that the implants be placed directly without placing an incision. The pilot drill is usually 0.2-0.3 mm smaller than the desired implant size and is drilled at a slow speed. The implants are driven at an angle of approximately 30-40° to the long axis of the maxillary teeth, and 10-20° to the mandibular teeth. This ensures optimum retention by augmenting the area of contact between the implant and adjacent bone. Case reports on micro-implant usage have shown their efficacy in anterior retraction with/without intrusion and molar uprighting.

***Maino et al (2003)***[20] introduced Spider Screw system, and the OMAS (Orthodontic Mini Anchor System) introduced by ***Lin et al (2003)*** are identical to the micro-implants. They vary in their form and their head design. The principles, however, remains the same. The trend presently seen is towards immediate loading of these screws.

***Micro Computerized tomograph and Radiographic studies.***

***Gray et al (1983)***[21] conducted the first study, wherein they tested the abilities of two types (Bioglass-coated and Vitallium) of small cylindrical endosseous implants to resist movement, when loaded with constant forces of orthodontic magnitudes. After a 28-day healing period, these implants were loaded with forces of 60g, 120g, and 180g. Analysis of implant movement after 28 days revealed that no statistically significant movement occurred at any of the three force levels for either type of implant. Histologic evaluation revealed a connective tissue encapsulation with the Vitallium implant, and an implant- bone bond with the Bioglass implant. No histologic evidence of implant movement was observed for either implant type at any force level.

***Roberts et al (1984)***[22] investigated the osseous adaptation to continuous loading of rigid endosseous implants in the femurs of rabbits. After 6-8 weeks of healing, a 100g load was applied for 4-8weeks by stretching a stainless steel spring between the implant. All but one of twenty loaded implants remained rigid. Immediate loading of 4 pairs of implants resulted in spontaneous spiral- type (“torsional”) fractures of the femur within 1 week. Their results indicated that the relatively simple and inexpensive titanium implants developed a rigid osseous interface. Six weeks was an adequate healing period, prior to loading, to attain rigid stability and avoid spontaneous fracture. Continuously loaded implants remained stable within the bone. These endosseous implants had potential as a source of firm osseous anchorage for orthodontics and dentofacial orthopedics.

***W.E. Roberts (1991)***[23] subsequently conducted a study to compare the bone adaptation to loaded teeth and dental implants. He concluded that the rigid implants, the functional equivalent of an ankylosed tooth, appeared to maintain rigidity by continually remodeling fatigued bone at the osseous interface.

***Gotcher et al (1991 )***[24] evaluated the bone surrounding loaded endosseous implants. Eight Branemark implants were placed in the upper and lower jaws of 4 mongrel dogs. Three months later, the implants were loaded for

15, 21, 27, and 36 months. One animal was sacrificed at each time point, and histomorphometric analysis was done. They concluded that the loaded implant had enhanced bone turnover. This effect may be due to the loading alone or due to the implant itself.

***Wehrbein et al (1998)***[25] conducted a histomorphometric study to evaluate the bone-to-implant contact of orthodontic implants in humans subjected to horizontal loading. In this study, implants were temporarily inserted into the mid-palatal and the mandibular retromolar areas. These implants were subjected simultaneously to both axial and oblique forces. Following completion of the orthodontic therapy, the implants were explanted by means of a trephine. This yielded a bony cylinder of approximately 0.4mm in thickness around the 3.3mm implant. Histomorphometric evaluation indicated that the implants were well integrated into the bone despite the prolonged application of force in the magnitude of 2-6 N.

***Saito et al (2000)***[26] evaluated the anchorage potential of endosseous titanium implants as anchors for mesiodistal tooth movement in the beagle dog. Two implants were surgically placed in healed mandibular extraction sites of second and third premolars on each side. One side served as a control or unloaded side, and the other side implants were subjected to 200g of lateral force. Histomorphometric analysis indicated that there was no statistical difference in the percent of peri-implant bone volume between the loaded and the unloaded sides, and no statistical difference between the compression and tension sides in both loaded and unloaded implants, which suggests that the implants maintained rigid osseointegration.

The same Japanese team, in **2001**, conducted a clinical and histological evaluation of titanium mini-implants as anchors for orthodontic intrusion in beagle dogs. The methodology was similar to the previous study; an intrusive force of 150g was applied on the loaded side and compared with the unloaded implants. The morphometrical findings indicated that the calcification of the peri-implant bone on the loaded implants was equal to or slightly greater than those of the controls. In addition, 6 of the 36 mini-implants were removed after tooth movement, and all of them were easily removed with a screw driver. Their findings suggested that mini-implants were effective tools for the anchorage of orthodontic intrusion in beagle dogs.

***Melsen et al (2001)***[27] performed a histomorphometric analysis of tissue reactions around implants placed in 6 adult Maccaca fascicularis monkeys, subjected to a well-defined force system. The analysis was performed on undecalcified sections cut perpendicularly to the long axis of the implant. The degree of osseointegration, bone density at varying distances from the implant, as well as the relative extent of resorption and formation of alveolar bone adjacent to the implant-bone interface was evaluated. The results were correlated with the local strain of the tissue estimated by the means of finite element analysis (FEA). It was found that the loading significantly influenced both the turnover and the density of the alveolar bone in the proximity of the implants. However, even unloaded implants tended to maintain the bone characteristics of the alveolar process. But the degree of osseointegration appeared to be independent of the loading of the implant.

***Deguchi et al (2003)***[28] carried out an investigation in 8 dogs. The aim was to quantify the histomorphometric properties of the bone- implant interface to analyze the use of small titanium screws as an orthodontic anchorage and to establish an adequate healing period. Overall, successful rigid osseous fixation was achieved by 97% of the 96 implants placed. All the loaded implants, including the elastomeric chain-loaded implants showed rigid fixation. Mandibular implants had significantly higher bone-implant contact than maxillary implants. Within each arch, the significant histomorphometric indices noted for the “three-week unloaded” healing group were: increased labeling incidence, higher woven-to-lamellar-bone ratio, and increased osseous contact. Analysis of these data indicates that small titanium screws were able to function as rigid osseous anchorage against orthodontic load for 3 months with minimal (under 3 weeks) healing period.

***Motoyoshi et al (2007)***[29] investigated miniscrew stability with respect to cortical bone thickness, inter root distance, insertion torque, distance from alveolar crest to the bottom of the maxillary sinus. They conducted a computerised tomo graphic study. The success of the miniscrew was established as no mobility of the miniscrew or pain after six months of loading. The authors concluded that success of the miniscrew was not related to placement and width/height of peri-implant bone. However the cortical bone thickness (1mm) and insertion torque (10 Ncm) played a very important role in the success of miniscrews.

***Reint Reynders et al(2009)***[30] reviewed the literature to quantify success and complications encountered with the use of mini-implants for orthodontic anchorage, and to analyze factors associated with success or failure and concluded that mini-implants can be used as temporary anchorage devices, but research in this field is still in its infancy. Interpretation of findings was conditioned by lack of clarity and poor methodology of most studies. Questions concerning patient acceptability, rate and severity of adverse effects of miniscrews, and variables that influenced success remain unanswered.

***Mario Veltri et al(2009)***[31] evaluated the soft bone primary stability of 3 different orthodontic screws by using the resonance frequency analysis which included Aarhus mini-implant , Mini Spider Screws, and Micerium Anchorage System. Four screws per system were tested and each screw was placed in 5 excised rabbit femoral condyles, providing experimental models of soft bone. Placement was drill-free for the A screw, whereas the MAS and S screws required a pilot hole through the cortical layer. After each placement procedure, resonance frequency was assessed as a parameter of primary stability. Differences among the groups were not statistically significant and concluded that the resonance frequency analysis is applicable to comparatively assess the primary stability of orthodontic miniscrews and the 3 systems had similar outcomes in an experimental model of soft bone.

***Jung-Yul Cha et al(2010)***[32] aimed to determine the effect of bone mineral density (BMD), cortical bone thickness (CBT), screw position, and screw design on the stability of miniscrews using computerized tomography. They placed ninety-six miniscrews of both cylindrical and tapered types in 6 beagle dog. Results showed the placement torque showed a positive correlation in the order of removal torque, BMD of the cortical bone and

CBT. Placement and removal torque values were significantly higher in the mandible compared with the maxilla. Tapered miniscrews had higher placement torque than did the cylindrical type but, the removal torque was similar in both groups,and concluded that the BMD of cortical bone, screw type, and screw position significantly influence the primary stability of miniscrews.

***Jung-Yul Cha et al(2010)***[33] compared the insertion and removal torque of tapered and cylindric orthodontic miniscrews. Ninety-six miniscrews were placed into the buccal alveolar bone of the mandible in six male beagle dogs. Results showed that the tapered miniscrews showed a higher mean maximum insertion torque than the cylindric miniscrews. The mean maximum removal torque of the tapered miniscrews was significantly higher than that of the cylindric miniscrews at 3 weeks after placement, but there was no significant difference in the mean maximum removal torque value between the tapered and cylindric implants after 12 weeks of loading. The percentage of bone-implant contact was similar between the groups after 3 weeks of loading and increased later. The percentage of bone volume/total volume was higher in the tapered miniscrews than in the cylindric miniscrews after 3 weeks of loading, but there was no significant difference between the groups after 12 weeks of loading.

***Marco Migliorati et al (2012)***[34] evaluated the correlations between bone characteristics, orthodontic miniscrew designs, and primary stability. They placed four different miniscrews in pig ribs. The miniscrews were first scanned with a scanning electron microscope to obtain measurable images of their threads. Subsequently, the maximum insertion torque of the screws and the maximum load value in the pull out force tests were measured; furthermore, bone specimen characteristics were analyzed by using cone-beam computed tomography. For each bone sample, the insertion site cortical thickness as well as both cortical and marrow bone density were evaluated. They found a significant dependence between pitch and maximum insertion torque. Positive correlations were also found between pullout force and maximum insertion torque, cortical thickness, and marrow bone density and concluded a strong correlations were observed among miniscrew geometry, bone characteristics, and primary stability.

***Antonio Gracco et al (2012)*** [35] determined the effects of variations in thread shape on the axial pullout strength of orthodontic miniscrews. They used a total of 35 miniscrews of both self-tapping and self-drilling miniscrews of diameter 2 mm and a thread shaft length of 12 mm, 7 of each design being considered,and were tested by performing pullout tests on a synthetic bone support. Results showed the control group with a buttress reverse thread shape had consistently higher pullout strength values than did the experimental groups of buttress, 75 joint profile, rounded, and trapezoidal design. A statistically significant reduction in pullout force was found between the buttress reverse and the buttress thread miniscrews. They concluded that the thread design influenced the resistance to pullout of the orthodontic miniscrews. The buttress reverse thread shape provided the greatest pullout strength.

***Yi-Ra Jung et al(2013)***[36] evaluated the effect of placement angles on the success rate of orthodontic microimplants and other factors with cone-beam computed tomography images. The authors implanted 228 orthodontic microimplants into the maxillary buccal alveolar bone of 130 patients with malocclusion. Results showed the overall success rate was 87.7% (200 of 228) and the orthodontic microimplant success rate statistically significantly increased as root proximity increased, but there were no statistical significances between placement angles and success rates, and cortical bone thickness and success rate .They concluded that the success rate of orthodontic microimplants is not affected by placement angles and is more significantly affected by root proximity than by cortical bone thickness. Cortical bone thickness is affected by placement angles, but root proximity is not affected by placement angles.

***Karan Bhalla et al(2013)***[37] conducted a prospective clinical trial correlating miniscrew implant (MSI) Micro/macro architecture, the method of placement, and biologic markers in peri-MSI crevicular fluid (PMICF) as indicators of bone response. Two types of MSIs (hybrid and cylindric) were placed in ten patients using a split-mouth technique. The MSIs were immediately loaded, and PMICF was collected on days 0, 7, 14, 21, 28, and 42 and evaluated using a Standard laboratory protocol. Surface morphology before placement and after retrieval of the MSI was observed using scanning electron microscopy (SEM).They concluded that the levels of both the ALP and AST are significantly higher in cylindric MSIs compared with hybrid MSIs, indicating a correlation to the type and method of placement of the MSI. The inflammatory markers show a definitive trend, with an elevation until day 14 and a decline after that. Observations from the SEM show a greater oxide layer formation in the hybrid MSI, which could imply a better bone-MSI contact ratio.

### *3. FEM Studies:*

### *A brief note on Finite Element Method:*

Recent research in Biologic Sciences, including Medicine and Dentistry, has made increasing use of technological tools developed by Engineering and allied sciences. One such tool that has found widespread use in orthodontic research is the Finite Element analysis (FEA), also called the Finite Element Method (FEM). For problems involving complicated geometries, it is very difficult to achieve an analytical solution. Therefore, the use of numerical methods such as FEA came into existence. FEA was initially developed in the early 1940s to solve structural problems in the aerospace industry, but has since been extended to solve problems in heat transfer, fluid flow, mass transport, and electromagnetics.

The application of FEM most related to orthodontics is the structural stress analysis. There are a number of studies in orthodontic literature using the FEM. These include studies of wire configurations, stresses in the periodontal ligament, determination of centers of resistance and rotation of teeth with normal and reduced alveolar bone height, stresses in the temporo-mandibular joint, jaws and cranium, stresses in brackets and adhesives, design of ceramic brackets, and studies of craniofacial growth.

FEM applicability increased as the computers became readily available for making complex computations. Various software packages are commercially available, which have made this possible. Some examples of such software packages are MSC PATRAN - NASTRAN, ANSYS, NISA-DISPLAY, ALGOR etc.

Because the components in a dental implant-bone system are extremely complex geometrically, FEA has been viewed as the most suitable tool for analyzing them. Since the present study was carried out using the FEM, it is apt to review this technique, its theoretical bases, and the difference between linear and non – linear analyses.

***Methodology of the Finite Element Method and its theoretical basis;***

The term finite element method was coined by Clough in 1960. By definition, it is a technique of discretizing a continuum into simple geometric shapes called elements, enforcing material properties and governing relationships on these elements, giving due consideration to loading and boundary conditions which results in a set of equations, and obtaining their solution to arrive at the approximate behavior of the continuum.

***To understand this, the process can be divided into three parts –***

***1. Pre-Processing***

a) This is the first step in the FEM. The structure (continuum) under study is broken down into a number of sub-parts. These are known as elements. The elements can be of various types, including spring elements, line elements (representing rods, beams etc), surface elements (for plates or membranes) or solid elements (for any solid structure).

The points connecting two or more elements are known as nodes. This process of dividing the continuum into discreet number of elements and nodes is known as *discretization*. The collection of nodes and elements is known as the finite element mesh. Increasing the number of elements (mesh density) results in a more accurate solution. Although this concept is attractive, increasing the number of elements also increases the complexity of the problem, and increases system requirements (processor capabilities and speed, storage requirement etc). Practical knowledge and judgment are needed to limit the number of elements to the minimum amount conducive to acceptable results. Usually, areas where great variations of stress are anticipated are divided into more number of elements.

b) After discretization, various material properties have to be prescribed for the elements. These include Young's modulus or Modulus of elasticity, Poisson's ratio, density, yield strength etc. The most important properties are the Young's modulus and Poisson's ratio.

c) The model is now a rigid body, free to translate in all three planes of space, or to rotate around any of the three axes. To simulate the true situation, boundary conditions have to be enforced on the model. The term 'boundary conditions' implies the real life constraints that act on a body. For example, if one considers a cantilever beam, the fixed end of the beam will be a constraint, or a boundary condition. To study the bending or stress distribution of a structure (a building, bridge or a tooth), appropriate boundary conditions need to be applied to the finite element model.

d) Once this is done, a load is applied onto the body in order to simulate the loading condition under study. Loads are applied at specific nodes (according to the given situation), and can be applied as forces or forced displacements. Also, most FEM software's allow application of loads in either a single step, or multiple load steps.

### *2. Processing*

After the preprocessing is completed, the model is ready to be executed. Before the analysis proper, the software rechecks all the aspects of pre processing. Then the global stiffness matrix is calculated.

This is done by calculating the element stiffness matrix for each element (using the Young's modulus and the Poisson's ratio), and then assembling it into one large matrix, the global stiffness matrix. The next step is to solve a fundamental equation to obtain the displacements of each node in all three planes of space. The equation is as follows –

$$\{F\} = [K]\{u\}$$

***Where –***

$\{F\}$ = Nodal force matrix

$\{u\}$ = Nodal displacement matrix and

$[K]$ = Global stiffness matrix.

The basic unknown is the nodal displacements. By solving this equation, the nodal displacements are calculated, and then, element stresses are derived as a function of their displacements. The advantage of such a numerical method is that the calculations are done at the element level, in small sections of the structure. This is well suited for the study of composite materials.

### *3. Post-processing*

After the solution is obtained, the results are displayed and analyzed. This is known as post-processing. Three types of post processing are possible with most software –

***a) Numerical output:*** This is a large set of results, which gets displayed. The deformation of each and every element in all three planes of space can be studied in depth using this option.

***b) Graphical output:*** This is a pictorial representation of the results, which get displayed in a color-coded manner. The colors represent the magnitude of stress /deformation at that area. This enables a quick interpretation of the results.

***c) Animated output:*** As against the static result display in the graphic output, the animated output shows the effects of the resulting stresses on the entire body in the form of dynamic displacement.

### *Linear Vs Non – Linear analysis*

When the strain produced is strictly proportional to stress, the material is said to behave in a linear fashion (since the stress strain graph is a straight line). When this is not the case, the behavior becomes non-linear (as for example, when the material is deformed beyond its **Proportional or elastic limit** - i.e., the limiting level of stress within which, a material completely regains its original shape and size upon complete removal of the applied load - it suffers some amount of permanent set, and its behavior becomes non elastic). This must be taken into consideration when performing a FEM analysis. If the applied load results in the structure behaving in a non-linear fashion, a non –linear analysis must be done. A linear analysis will result in an erroneous result. The two principal causes for non-linearity, which are of interest to us, are geometrical and material.

1. Geometric Non-linearity. If a structure experiences large deformations, its changing geometric configuration can cause the structure to respond non-linearly. The study of orthodontic wires necessarily involves large deflections, and hence is non – linear in nature. As the deflection goes on increasing, the stiffness will also keep changing.

2. Material Non-linearity. Stress strain relationship of a material can be affected by such factors as deformation beyond elastic limit, environmental conditions like temperature and amount of time that the load is applied (creep response). Examples of such material are rubbers, concrete, soils etc.

## Finite element method studies

There have been many studies, which have examined the biologic interactions between dental implants and living tissue. Few studies have reported on the biomechanical aspects of dental implants. Most of these studies have been on prosthodontic implants and very few on the implants used for orthodontic anchorage.

***FEM studies on Orthodontic implants:***

***Chen et al (1999)***[38] conducted a coordinated histomorphometric and 3D FEA to investigate the mechanical environment of cortical bone adjacent to the threads of a retromolar endosseous implant, used for orthodontic anchorage to mesially translate mandibular molars in response to normal functional loading. A 3D model of the mandible and the retromolar implant with the surrounding cortical bone were modeled. A strong stress pattern change was found immediately around the implant, which was reflected by a moderate change of stresses between the threads and a significant increase in stress at the tips of the threads.

***Va´squez M et al (2001)***[39] conducted a 3-D FEA to evaluate the initial stress differences between sliding and sectional mechanics with an endosseous implant as anchorage. A mathematical model was constructed that used the finite element method, which simulated an endosseous implant and an upper canine with its periodontal ligament and cortical and cancellous bone. Levels of initial stress were measured during 2 types of canine retraction mechanics (friction and frictionless). The lower magnitude and more uniform stresses in the implant and its cortical bone were found to have a moment-force ratio (M/F) of 6.1:1, whereas the canine and its supporting structures exerted a M/F ratio of 10.3:1. On the basis of these results, they concluded that when the anchor unit is an endosseous implant, it was better to use a precalibrated retraction system without friction (T-loop) where a low load-deflection curve would be generated. Overall, the area with the highest stress was the cervical margin of the osseointegrated implant and its cortical bone. These stresses are of such low magnitude that they are unable to produce a permanent failure of the implant.

***Gedrange et al (2003)***[40] conducted a 3D FEA of endosseous palatal implants and bone after vertical, horizontal and diagonal force application. Three types of endosseous implants were: type 1 was a simple, cylinder-shaped implant; type 2 a cylinder-shaped implant with a superperiosteal step; and type 3 a cylinder-shaped implant, subperiosteally threaded, and with a superperiosteal step. The load on the implant was investigated under three conditions of bite, and orthodontic forces ranged from 0.01 to 100N (vertically, horizontally and diagonally). Vertical loading caused bone deformation of more than 600μeps (micro-strains) at the simple implant. The largest deformations at this load were found in the trabecular bone with all three implant geometries. However, trabecular bone deformation was reduced by a superperiosteal step. Horizontal loading of the implants shifted the deformation from the trabecular to the cortical bone. Furthermore, a large deformation was measured at the transition from cortical to trabecular bone. The smallest deformations (less than 300μeps) were found for implants with superperiosteal step and diagonal loading (type 2). The use of threads provided no improvement in loading capacity. All implant types investigated showed good biomechanical properties. However, endosseous implants with a superperiosteal step had the best biomechanical properties under low loads.

***Melsen and Verna (2005)***[41] evaluated the load transfer from the Aarhus miniscrew to the surrounding bone and the effect of varying thickness of the underlying cortical and trabecular bone. They developed two different 3D finite element models: a geometrically accurate model and a parametric model. A mesially directed force of 50g was applied at the head of both the models. The results indicated that the primary component of the load transfer occurred at a single revolution of the miniscrew thread within the cortex. The implant tipped, causing tensile stress in the direction of the force. Stress levels were higher in the cortical bone compared to the trabecular bone and vice versa for the strain. When the peak strains in bone were co-related to Frost's mechanostat theory, it was observed that these strains reached the pathological overload window only when the cortical bone thickness was reduced to <0.5mm.

***Gallas et al (2005)***[42] carried a 3D FEA to determine the pattern and distribution of stresses within the ITI-Bonefit endosseous implant and it's supporting tissues, when used as orthodontic anchorage unit. They simulated 2 conditions, one with no osseointegration and the second model with full osseointegration. The threaded implant was simulated in an edentulous segment of a human mandible with cortical and cancellous bone. The results, using both models, indicated that the maximum stresses were always located around the neck of the implant, in the marginal bone. They concluded that this area should be preserved clinically in order to maintain the bone-implant interface structurally and functionally.

***Eva Stahl et al (2009)***[43] compared the stresses produced in cortical bone between sixteen different implants from six manufacturers .The cortical bone thickness was maintained at 1mm and 2mm and the Young's modulus of the cancellous bone was fixed at 100 MPa and 1GPa . The load direction was at 0 and 45 degrees and the buccal load direction was at 5 N. They concluded that stress values are higher for a cortical bone thickness of 1mm and Young's modulus of cancellous bone at 100 MPa. The deflection of most implants were of 4-10 µm in magnitude .The deflection of the implants increased as the Young's modulus of the cancellous bone decreased when the cortical bone thickness was at 1mm.This association was not seen when the bone thickness was at 2mm.The load direction when placed in a buccal direction decreased the stresses by 35%.

***Ashish Handa et al (2011)***[44] evaluated the influence of thread pitch on maximum effective stress in cortical bone. They selected implants of thread pitch of 0.5, 1.0, 1.5mm placed 1mm mesial to lower mandibular molar with a force magnitude of 2 N at a 45 degree angulation. The Von-mises stresses were maximum at the neck of the implant, bone interface and increased with increasing thread pitch. However the pattern of the stress was not influenced by the thread pitch.

***Akihiro et al (2011)***[45] measured the stress distribution in cortical as well as cancellous bone using 3 dimensional FEM .The authors used 3 types of implant design cylindrical pin, helical thread,non helical thread designs. A force of 2N were applied in all 3 directions and force angulations of 30,40,45,50,60,70,80,90 were used. Peak stresses of 9.16Mpa-14.8Mpa were obtained in the case of pin type and 17.8Mpa-75.2Mpa in the case of helical thread implant. They concluded that placement angle, type of miniscrew and direction of the force played a major role in determining the magnitude of stresses in bone.

***Athina Chatzigianni et al (2011)***[46] compared the numerical and experimental data obtained from FEM and mechanical testing of diameter, angle of insertion, implant type and length of the miniscrews .Various diameter (1.5-2mm) , lengths (7,9mm) were tested at an angulation of 45,90 degrees of the miniscrews at a force level of 0.5,5N. The authors simulated the above parameters using a FEM and 3 D laser optic reflection was used to study the mechanically tested specimens. The authors concluded that the results obtained from the FEM was comparable to that of mechanical testing of the miniscrews and also at low force levels 0.5N none of the above said parameters had an influence on the miniscrew deformation.

***Choi et al (2011)***[47] investigated the stress pattern in a two-piece miniscrew. The authors used a 1.8×8.5 mm miniscrew. Three different head sizes of 1mm, 2mm, 3mm were available .The cortical bone thickness was maintained at 1mm and a 2N lateral force was used. The head part of the screw was friction fitted and an FEM study was done. They concluded that the minimum principal level forces created were 10.84-15.33 M pa which was within the tolerance level of bone.

***Shivani Singh et al (2012)***[48] calculated the stress distribution in two types of implant materials titanium **ASTM 136** and medical grade stainless steel **ASTM 316 L**. Both horizontal and torsional loading was used. Peak stresses was produced in the neck of the implants in both materials at the cortical bone implant interface .Slight bending of the neck of the titanium implant was also seen.

***Jenny Zwei-Chieng Chan et al (2012)***[49] evaluated the influence of thread angulation, degree of taper, taper length, stiffness and screw displacement on stresses created in bone. Four types of implants were custom manufactured and tested mechanically in artificial, homogenous bone. FEA was done and compared to the results obtained from mechanical testing. With a fixed diameter of 2mm, thread length of 9.82 mm, pitch of 0.75 mm greater thread depth, smaller taper angle, shorter taper length produced maximum stresses in bone. They also concluded that maximum pull out strength was found in implants with a core/external diameter of 0.68.

***Te-chun Liu (2012)***[50] et al compared the bone quality, loading conditions, screweffects and implanted depth. They utilized a 3 D bone block and used a FEM to analyze the stress patterns produced in bone. The bone blocks were of varying cortex thickness, cancellous bone density and the implants simulated were of varying diameter, length with differing depth. The results were as follows, the stress/displacement decreased with decreasing cortical bone thickness, but cancellous bone played a minor role in influencing the stress patterns. The two indices were linearly proportional to the force magnitude and produced the highest values perpendicular to the long axis of the screw. They also concluded that a wider screw produced a decreased force and exposure length has the most influence on stress patterns in cortical bone.

***Abhishek H Meher et al (2012)***[51] compared varying cortical bone thickness, applied forces and angulations. They compared cortical bone thickness of 1.5mm and 2mm, force or two different magnitudes 200 and 300 g respectively. The force angulation was maintained at 70, 90,110 and 130 degrees. They concluded that the maximum equivalent stress **MES** and maximum deformation **MD** had an inverse relation with cortical bone thickness. The maximum equivalent stress **MES** decreased with increased force angulations whereas the maximum deformation **MD** increased with increasing force angulation .On increase in load the maximum equivalent stress **MES** showed no change but the maximum deformation **MD** increased.

***Sun-Hye Baek et al (2012)***[52] compared the stress in the cortical bone due to mastication and orthodontic miniscrews in three types of morphogenetic features (high, average, low angle) cases. The three morphological characteristics were modeled three dimensionally by hyper mesh software. A miniscrew was paced between lower mandibular molar and second premolar. The stress produced in the cortical bone during mastication and orthodontic loading of the miniscrew (150 g) was simulated using the FEM and the peri orthodontic miniscrew compressive stress (POMI-CSTN) was compared. The results was that (POMI-CSTN) produced due to mastication was in the order of 1401.75µE and due to orthodontic loading was 1415µE.The authors concluded that the strain due to masticatory stress do not exceed the normally allowed compressive bone strains but additional orthodontic traction force to the miniscrew may exceed the threshold value result in the failure of the miniscrew.

***Issa et al (2012)***[53] evaluated the effect of different angulations in the cortical bone stresses produced in the bone. The authors conducted an FEM study where miniscrews of 1.3×7, 1.3×8 mm implants were generated in D3 and D2 bone .Simulations were done with a force load of 200g and different angulations of 30, 45, 60 and 90 degree angulations. The authors concluded that the maximum Von-mises stresses produced with increasing degrees of angulation of the miniscrew. Greater stresses were produced in cortical than cancellous bone and D3 bone showed higher stresses than D2 bone.

***Ramzi et al (2012)*** [54] investigated the Von mises , principal stresses in cortical bone influenced by design factors of the implant as well as bone factors such as Young's modulus of the cancellous bone. They used 26 bone blocks modeled 3-dimensionally and a force of 2 N in a linear direction was used. They concluded that certain implant design factors such as implant diameter, head length, thread size and bone factors such as Young's modulus of elasticity influenced the stresses produced in cortical bone. Other factors such as thread shape, thread size, cortical bone thickness did not have an impact.

***Ting-Sheng Lin et al (2013)***[55] compared exposure length, insertion angle and direction of force on cortical bone stresses. The authors used 27 models to simulate the effects of these variables on cortical bone stress. The results showed that exposure length has the highest influence (82.38%) followed by insertion angle (6.03%).Direction of forces had no influence on stress values. All the three parameters had an influence on cancellous bone values.

***Jordi Marce-Norgue et al (2013)***[56] evaluated miniscrew design parameters such as screw length, head diameter, shank diameter, shank length, shank shape, thread shape and pitch of ten different commercially available miniscrews. Bio mechanical parameters such as cortical bone thickness, insertion angle, drilling depth was also considered in evaluating the Von-mises stresses and the deformation of the screws. A force value of 1 N was applied perpendicular to the implant was applied to the miniscrews. They concluded that the direction of the applied forces had a limited influence on the stresses produced and geometric parameters such as head diameter, diameter of the shaft, thread pitch had an important relation to the Von-mises stresses produced. According to the authors an ideal implant should have a diameter of 2mm, cylindric shape, short and wide head, short and wide shank and appropriately sized threads.

***4. Mechanical pull out strength of miniscrews:***

***Sean Shiu-Yao Liu et al (2012)***[57] compared the effect of diameter of mini screws in propagation of cracks in bone. The authors compared micro cracks in bone by comparing a control, pilot drill only, pilot drill with miniscrews of various diameters (1-3mm) in maxilla and mandible randomly assigned in beagle dogs. The authors used basic fuschin dye followed by fluorescence microscopy. The authors concluded that pilot drilling with miniscrews produced the highest amount of micro cracks. The diameter of the miniscrews did not influence the amount of crack production in bone.

***Nam-Ki Lee et al (2010)***[58] compared the effect of diameter as well as the shape of miniscrews in propagation of cracks in bone. The authors compared number micro cracks (NC), accumulated crack length (ACL), maximum radius of cracks (MRC), longest crack (LC), maximum insertion torque (MIT) and cortical bone thickness. The study was done in rabbit bone with 28 miniscrews of 6mm length, 1.5-2mm diameter. Both conical and cylindrical miniscrews were used. The authors concluded that increased diameter and taper resulted in an increase in maximum insertion torque (MIT), number of cracks (NC) and longest cracks. (LC) and an increased diameter resulted in an increase in accumulated crack length (ACL) and maximum radius of cracks (MRC).

***Benedict Wilmes et al (2007)***[59] compared the insertion torque and subsequently the primary stability between conical and cylindrically designed six commercially available miniscrews in pig tibia. The authors used 42 sections of pig tibia and each one of the sections had 5 Dual Top (Jeil Medical) as a control. The authors concluded that the tapered miniscrews of similar dimensions had significantly increased insertion torque values compared to the cylindrical miniscrews and increased diameter also increased the insertion torque values. However the authors also concluded that increasing insertion torque values beyond a threshold value also may result in excessive bone compression and subsequent reduction in primary implant stability.

***Athena Chiatzigianni et al (2010)***[60] evaluated the effect of diameter and length on primary stability of the implants. The authors used 90 self drilling implants of two different diameters (1.5-2mm) and two different lengths (7,9mm) in bovine bone with two different force levels applied (0.5,5N) using niti coil springs. A 3D laser optic system was used to measure the miniscrew stability non- invasively. The authors concluded that both increased diameter and length resulted in decreased miniscrew mobility.

***Britta Florvag et al (2010)***[61] compared the pull out strengths to determine the initial primary stability of six different commercially available miniscrews. The authors compared the pull out strengths between conical and cylindrical implants using pig femoral bone as the implant site and an instron testing machine to measure the pull out strengths at various angulations. The authors concluded that conical implants had a better insertion torque values than cylindrical screws and larger diameter screws had better pull out strength than smaller diameter miniscrews. The conical screws did not show a reduced pull out strengths at lesser angulations whereas the cylindrical screws showed a better pull out strengths at lesser angulations but were not significant to conical screws at 40 degrees.

# MATERIALS & METHODOLOGY

The following materials was used for this study,

***A) Miniscrews (*Absoanchor, Dentos, *Daegu, Korea):***

1) Short head 1.3mm diameter, 1.2 degree taper, 8mm length, pitch 0.5mm (SH 13-1208)

2) Short head 1.3mm diameter, 8mm length, 0.5mm (SH 13-08)

3) Short head 1.5mm diameter, 1.4 degree taper, 8mm length, pitch 0.7mm (SH 15-1408)

4) Short head 1.5mm diameter, 8mm length, pitch 0.7mm (SH 15-08)

***B) Software:***

1) *Solid works modelling* (Solid works Corp; Concorde, MA, USA) version 13.0

2) *Hyper mesh CAE* (Altair engineering, Troy, Michigan, USA) version 11.0

3) *Ansys* (Swanson Analysis Inc**;** Huston, PA, USA) version 12.1.

***C) Workstation:***

1) Intel core 2 duo with 2.1 GHz

2) 2 GB of RAM

3) 2GB Graphics card

4) 320GB hard Disc

5) 17” Monitor

# METHODOLOGY

The study was conducted at the Department of Orthodontics and Dentofacial Orthopedics, Tamilnadu Govt. Dental College, Chennai, with technical assistance from CAD Designing & Drafting solutions, Chennai

The study was done in the following way;

1) Measurement of the dimensions of the miniscrews obtained from the manufacturer.

2) Modelling of the miniscrews using solid works imaging software

3) Geometric model of the miniscrew discretized using the hypermesh software

4) Bone block of dimensions (20×20×10 mm) was constructed and meshed

5) Boundary conditions and assumptions

6) Assigning of material properties

7) Finite element analysis:

a) Meshed model of both the miniscrews and the bone block imported into the Ansys software for testing the various parameters such as diameter, insertion depth, pitch of the miniscrew, cortical bone thickness and angulation of the applied force.

b) Analysis of stresses

***1) Measurement of the miniscrews:***

The miniscrew micro measurements was obtained from the manufacturer *Absoanchor Dentos* (Daegu, Korea) for the miniscrew types , small head type 1.3 mm diameter, 1.2 degree taper, 7mm length (SH 1312-07) , small head type 1.3 mm diameter, 7mm length (SH 13-07) , small head type 1.5 mm diameter, 1.4 degree taper, 7mm length (SH 1514-07).

***2) Modelling of the miniscrews:***

The miniscrews of various dimensions 1.3mm diameter, 1.2 degree taper, 8mm length (model A), 1.3mm diameter, 8mm length (model B),1.5mm diameter, 1.4 degree taper, 8mm length (model C), 1.5 mm diameter, 8mm length (model D), 1.3 mm diameter 1.2 degree taper, 8mm length, pitch 1mm (model E), 1.3 mm diameter, 8mm length, 1mm pitch (model F) were modelled using the micro-measurements obtained previously from the manufacturer *Absoanchor Dentos* ( Daegu, Korea) using *solid works modelling software* (Solid works Corp;Concorde, MA, USA) version 13.0. The geometric models of the implant was thus obtained using the reverse engineering technique by measuring the miniscrews using tool makers microscope or, in this case obtained from the manufacturer and creating a 3D CAD models. The resulting 3D CAD models were obtained in .IGES (initial graphics exchange specification) format which is subsequently uploaded to the *hypermesh software* for discretization and meshed.

***3) Meshing of the miniscrews:***

The 3D CAD models in .IGES format were subsequently uploaded to the meshing software *Hypermesh CAE version 11.0* (Altair engineering, Troy, Michigan, USA). The models were meshed and a 10 node tetrahedral element was selected resulting in approximately 69,000 elements for each model which was sufficient to obtain solution convergence.

### *4) Construction of bone block and meshing:*

A solid bone block of 20mm×20mm×10mm was constructed with two different cortical bone thicknesses of 1mm and 2mm respectively. A 2mm cortical bone thickness is a pre-requisite for miniscrew insertion and hence was selected to compare stresses created against a 1mm thickness of cortical bone. The bone block was also subsequently meshed with hypermesh software and the whole assembly was imported into the finite element software.

### *5) Boundary conditions and assumptions:*

The boundary conditions were fixed at superior, inferior, mesial and distal boundaries. All materials were considered to be linear and isotropic and the bone/implant interface was assumed to be rigidly bonded.

### *6) Assigning of material properties:*

Young's modulus and Poisson's ratio was assigned for cortical, cancellous and miniscrews according to those reported in the literature[45].

### *7) Finite element analysis:*

The whole assembly of miniscrews and the bone block was imported into the finite element workbench *Ansys* (Swanson Analysis Inc; Houston, PA, USA) version 12.1. A 2N force was applied to the miniscrew (co-ordinates X=0.7N, Y=-1.278N, Z=-0.2N) at two different angulations of 30 and 60 degrees.

## ANALYSIS OF STRESSES:

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

- Maximum or1st principle stress - MaxPS :[tensile stress,; $\sigma max=(\sigma x+\sigma y)/2+\tau max$];
- Minimum or 3rd principle stress- MinPS: [compressive, $\sigma min=(\sigma x+\sigma y)/2-\tau max$], and
- Von Mises stress- vonMS [equivalent,; $\sigma v= \sqrt{([(\sigma 1-\sigma 2)2+(\sigma 2-\sigma 3)2+(\sigma 3-\sigma 1)2]/2)}$ ].

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. Stress analysis was performed both visually and numerically. 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 blue to red. 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.

The yield strength of the materials in the model (bone and titanium) was equated with those extracted from existing studies[45]. 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.

***Table-1: Table showing control and variable simulations between tapered (Group-I) and cylindrical (Group-II)***

| **Tapered Implant** | **Cylindrical implant** |
|---|---|
| Control Simulation | |
| Dia-1.3mm,taper-1.2mm,length-8mm<br><br>Insertion depth-till neck, cortical bone thickness-2mm,force value-2N,force angulation-45 degrees | Dia-1.3mm,taper-0mm, length-8mm<br><br>Insertion depth-till neck, cortical bone thickness-2mm,force value-2N,force angulation-45 degrees |
| Variable Simulation | |
| Diameter-1.5 mm | Diameter-1.5 mm |
| Cortical bone thickness 1mm | Cortical bone thickness 1mm |
| Angulation of force -60 degrees | Angulation of force -60 degrees |
| Pitch of implant-1mm | Pitch of implant-1mm |

| | |
|---|---|
| Insertion depth-5mm | Insertion depth-5mm |

***Table-2: Table showing the number of nodes and elements employed in the control and variable simulations between the two groups.***

**Total number of nodes and elements in each case**

| | Group-I | | Group-II | |
|---|---|---|---|---|
| | **Nodes** | **Elements** | **Nodes** | **Elements** |
| **Control Simulation** | | | | |
| Control | 14003 | 69655 | 13565 | 67195 |
| **Variable Simulation** | | | | |
| Diameter | 13723 | 68619 | 13765 | 68865 |
| Cortical Bone Thickness | 13150 | 64298 | 12718 | 61672 |
| Pitch | 14003 | 69655 | 13565 | 67195 |
| Angulation of force | 13302 | 65615 | 13212 | 65021 |
| Exposure length | 13408 | 63389 | 13016 | 61345 |

***Table-3: Properties of Materials used for finite element analysis.***[45]

| Material | Young's Modulus (GPa) | Poisson's ratio |
|---|---|---|
| Pure Titanium | 110.00 | 0.33 |
| Cortical Bone | 14.70 | 0.30 |
| Cancellous Bone | 0.49 | 0.30 |

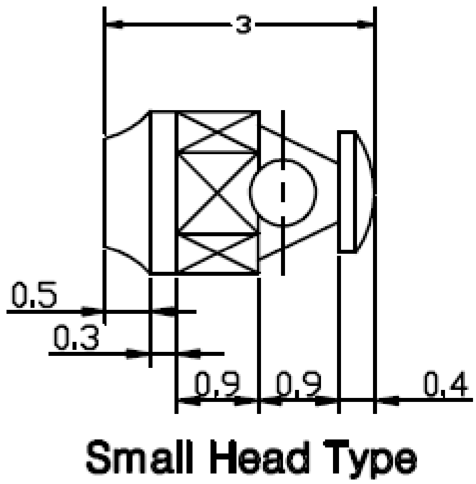



**Fig 1 small head type**

0.4mm-attachement head, 0.9mm-recess, 0.9mm-hex head, 0.3mm-tissue stop, 0.5 mm-transmucosal collar, 3mm-head length

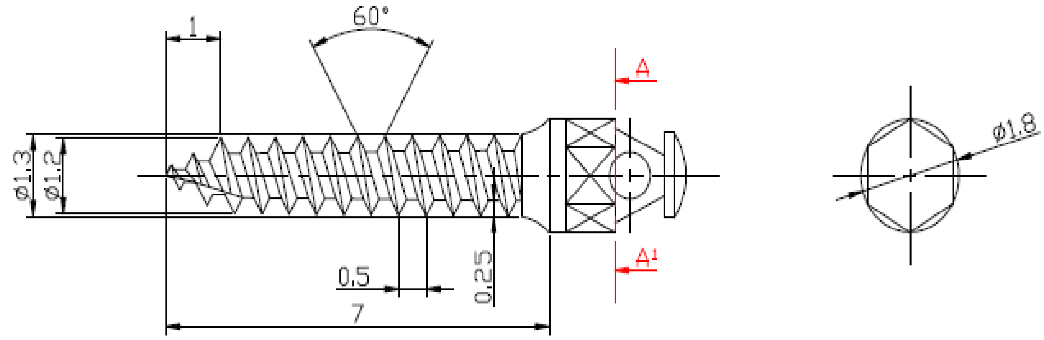



**Fig 2a Measurement of short head,1.3 mm diameter,1.2 mm taper,7mm in length**

1.3mm-major diameter, 1.2mm-taper, 1mm-screw tip length, 60 degrees-thread angle, 0.5mm-pitch, 0.25mm-thread depth, 7mm-screw length, 1.8mm (separate diagram)-width across the flats

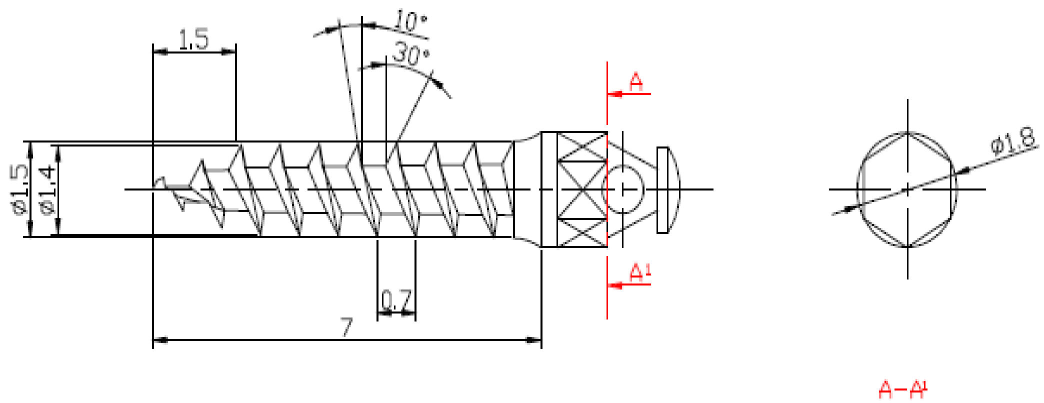



**Fig 2b Measurement of short head,1.5mm diameter,1.4mm tper,7mm in length**

1.5mm-major diameter, 1.4mm-taper, 30 degrees-half angle, 10 degrees-helix angle, 0.7mm-pitch, 1.8mm 1.5mm-screw tip length, 7mm-screw length, (separate diagram)-width across the flats,

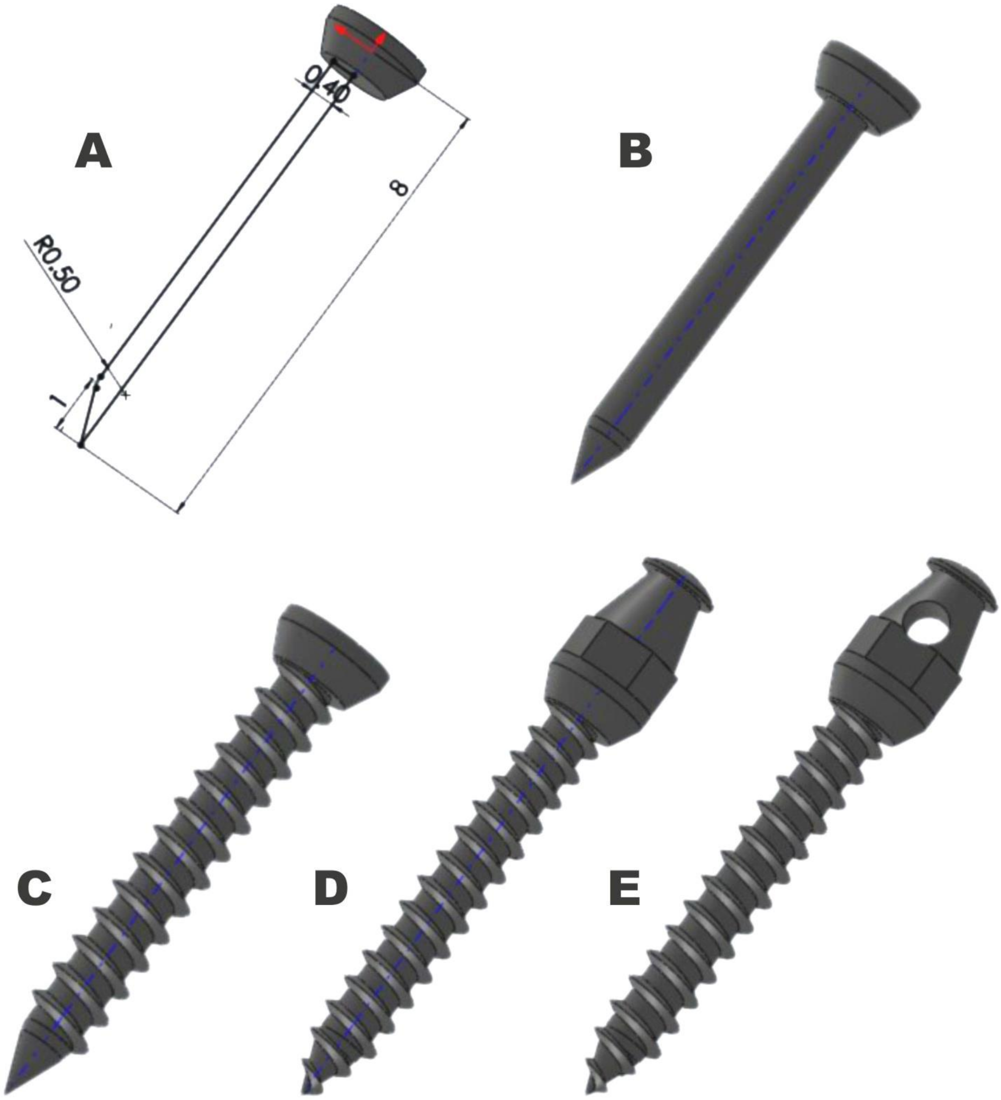
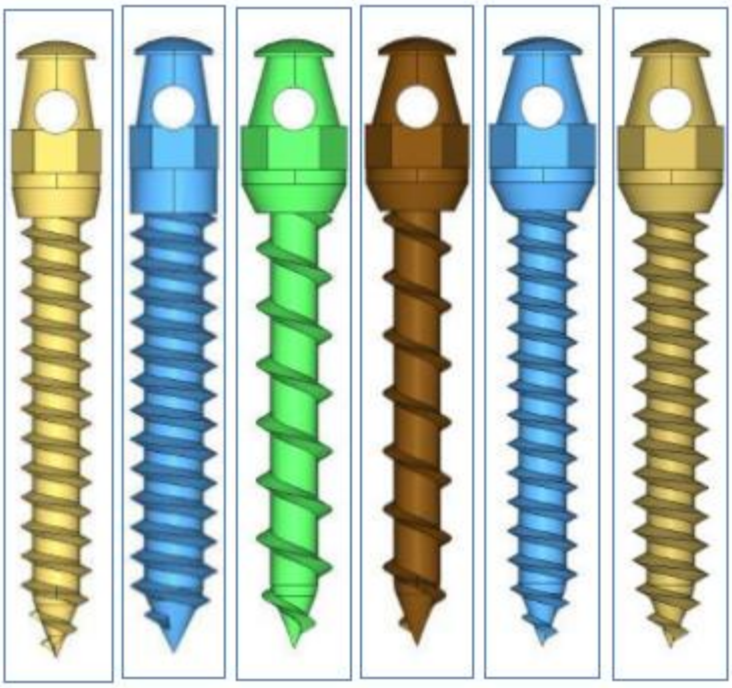
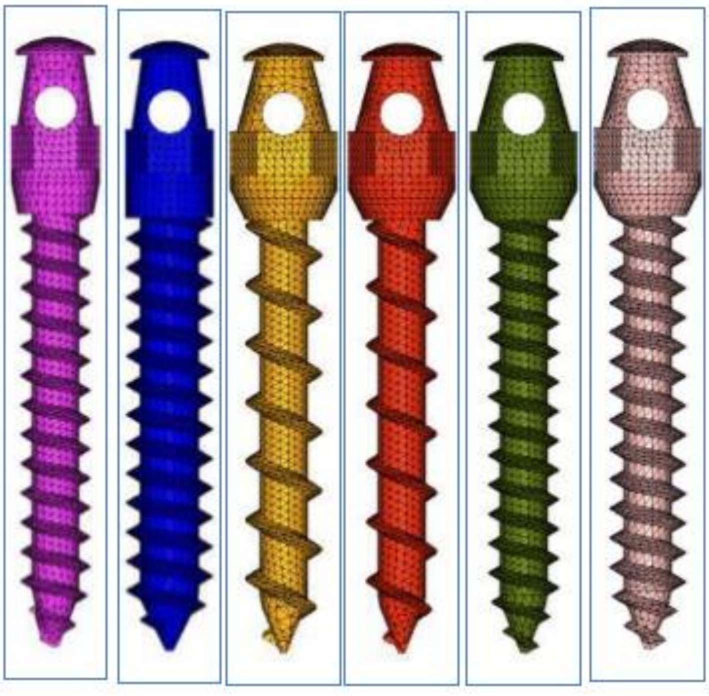



**Fig 3 Solid works construction of *Absoanchor Dentos* miniscrew**
(a) Initial hemi sectioned outline of screw (b) 3D revolution of screw outline (c) Addition of hexagonal element (d) Thread addition via helical sweep function (e) Removal of head segment material and excess threads by substraction cuts



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

Fig-4 (a): Geometric models of the mini screws after solid works imaging (a) 1.5mm diameter tapered (b) 1.5mm cylindrical (c) 1.3mm diameter, pitch 1mm (d) 1.3mm diameter, Pitch 1mm, (e) 1.3mm taper (f) 1.3mm cylindrical



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

Fig-4 (b): Finite element models of the mini screws after meshing

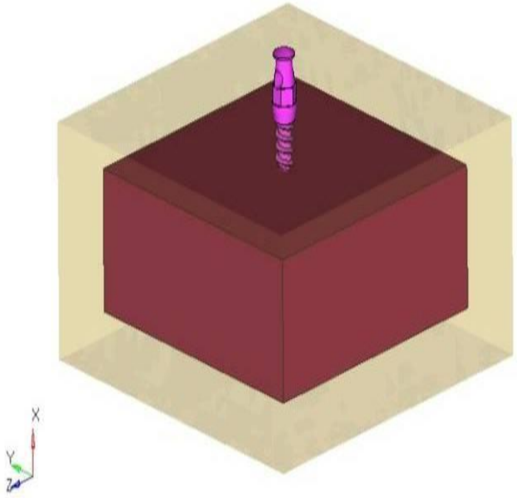


Fig-5 (a): Construction of solid bone block with mini screw at the centre (oblique view).

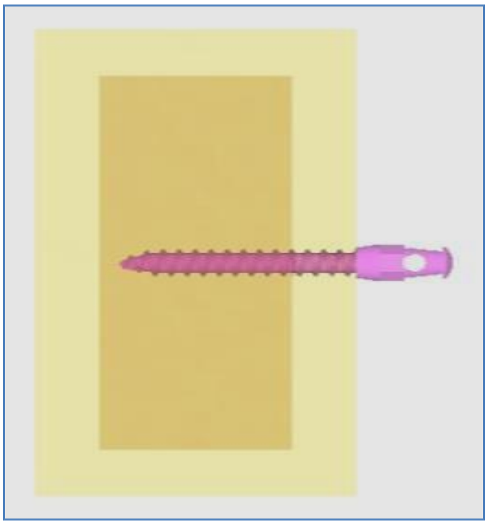


Fig-5 (b): Construction of solid bone block with mini screw at the centre (Lateral view)

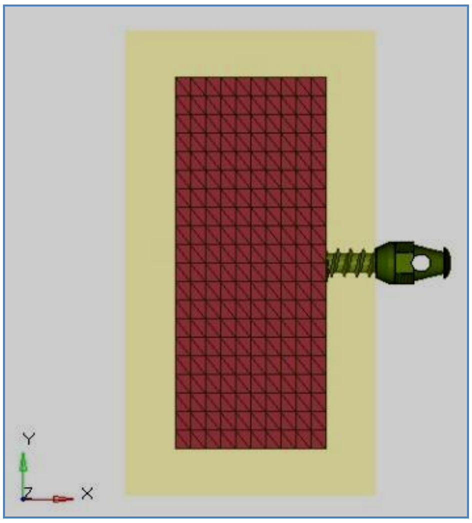



Fig-6 (a): Finished bone block model with implant meshed and boundary conditions applied.

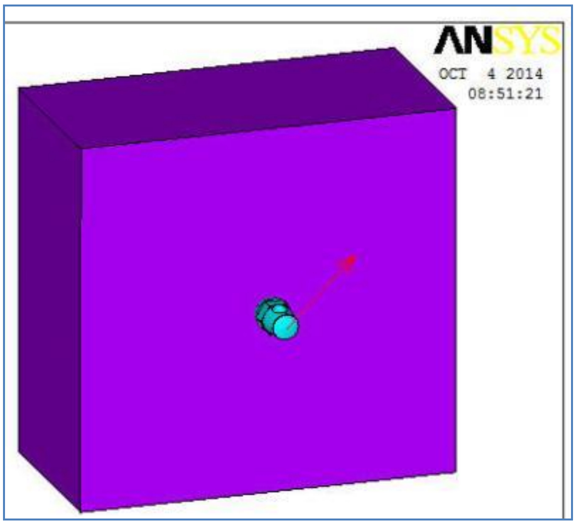



Fig- 6 (b): Application of 2N force to the finished bone block model with implant.

# RESULTS

***Table-4: Comparison of stress between Group-I/ Group-II with control simulation.***

| | | Group I | Group II |
|---|---|---|---|
| Maximum Principal stress (MPa) | Implant | 30.975 | 32.23 |
| | Cortical | 11.478 | 16.434 |
| | Cancellous | 0.080562 | 0.055537 |
| Minimum Principal stress (MPa) | Implant | -50.059 | -41.582 |
| | Cortical | -17.596 | -17.674 |
| | Cancellous | -0.027194 | -0.035422 |
| Von Mises (MPa) | Implant | 28.598 | 30.356 |
| | Cortical | 9.456 | 9.872 |
| | Cancellous | 0.057456 | 0.047493 |

***Table-5: Comparison of stress between Group-I/ Group-II with diameter as variable.***

| | | Group I | Group II |
|---|---|---|---|
| Maximum Principal stress(MPa) | Implant | 61.4 | 51.105 |
| | Cortical | 17.176 | 12.381 |
| | Cancellous | 0.045327 | 0.071025 |
| Minimum Principal stress(MPa) | Implant | -60.518 | -54.731 |
| | Cortical | -16.775 | -9.7 |
| | Cancellous | -0.03376 | -0.02634 |
| Von Mises (MPa) | Implant | 56.222 | 51.57 |
| | Cortical | 9.994 | 9.152 |
| | Cancellous | 0.038496 | 0.052328 |

***Table-6: Comparison of stress between Group-I/ Group-II with cortical bone thickness as variable.***

| | | Group I | Group II |
|---|---|---|---|
| Maximum Principal stress(MPa) | Implant | 30.987 | 32.229 |
| | Cortical | 14.913 | 20.219 |
| | Cancellous | 0.248057 | 0.135663 |
| Minimum Principal stress (MPa) | Implant | -50.059 | -41.582 |
| | Cortical | -22.704 | -21.862 |
| | Cancellous | -0.128302 | -0.131533 |
| Von Mises (MPa) | Implant | 28.584 | 30.372 |
| | Cortical | 15.837 | 17.534 |
| | Cancellous | 0.223655 | 0.153434 |

***Table-7: Comparison of stress between Group-I/ Group-II with pitch of the mini screw as variable.***

| | | Group I | Group II |
|---|---|---|---|
| Maximum Principal stress(MPa) | Implant | 28.568 | 27.582 |
| | Cortical | 11.487 | 16.44 |
| | Cancellous | 0.074698 | 0.042284 |
| Minimum Principal stress(MPa) | Implant | -49.583 | -27.282 |
| | Cortical | -17.45 | -16.491 |
| | Cancellous | -0.032838 | -0.034559 |
| Von Mises (MPa) | Implant | 24.431 | 25.968 |
| | Cortical | 9.366 | 10.371 |
| | Cancellous | 0.053847 | 0.041637 |

***Table-8: Comparison of stress between Group-I/ Group-II with angulation of bone as variable.***

| | | Group I | Group II |
|---|---|---|---|
| Maximum Principal stress (MPa) | Implant | 31.568 | 31.562 |
| | Cortical | 12.71 | 14.37 |
| | Cancellous | 0.059265 | 0.073892 |
| Minimum Principal stress(MPa) | Implant | -31.989 | -29.215 |
| | Cortical | -15.468 | -13.298 |
| | Cancellous | -0.032291 | -0.028465 |
| Von Mises (MPa) | Implant | 30.402 | 27.639 |
| | Cortical | 9.407 | 10.923 |
| | Cancellous | 0.04124 | 0.050909 |

***Table-9: Comparison of stress between Group-I/ Group-II with exposure length of the mini screw as variable***

| | | Group I | Group II |
|---|---|---|---|
| Maximum Principal stress(MPa) | Implant | 197.142 | 126.045 |
| | Cortical | 59.628 | 46.752 |
| | Cancellous | 0.063419 | 0.067669 |
| Minimum Principal stress(MPa) | Implant | -207.281 | -140.561 |
| | Cortical | -53.748 | -19.268 |
| | Cancellous | -0.142616 | -0.114158 |
| Von Mises (MPa) | Implant | 174.006 | 30.356 |
| | Cortical | 41.922 | 9.872 |
| | Cancellous | 0.098764 | 0.047493 |

# RESULTS

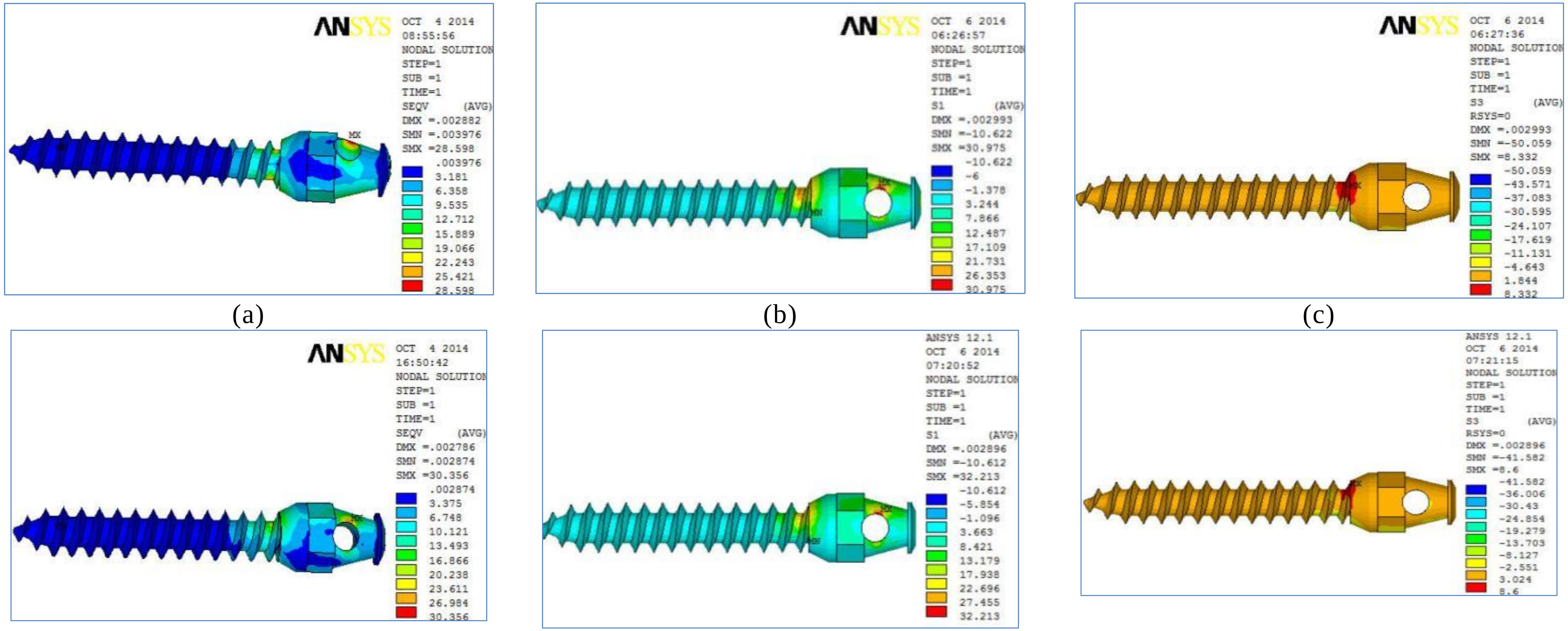



Fig-7: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in mini screws for control simulation

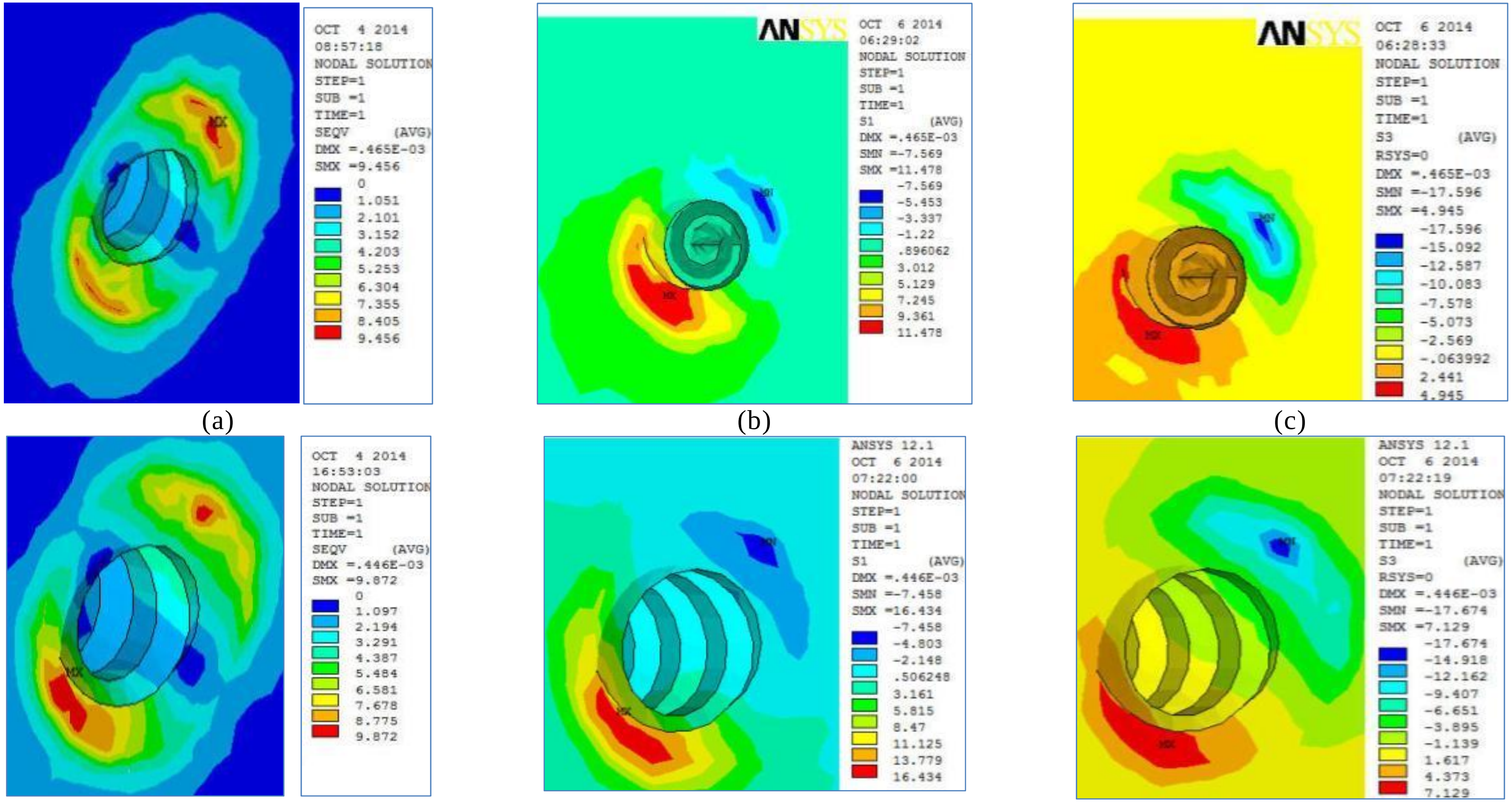



Fig-8: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in cortical bone for control simulation

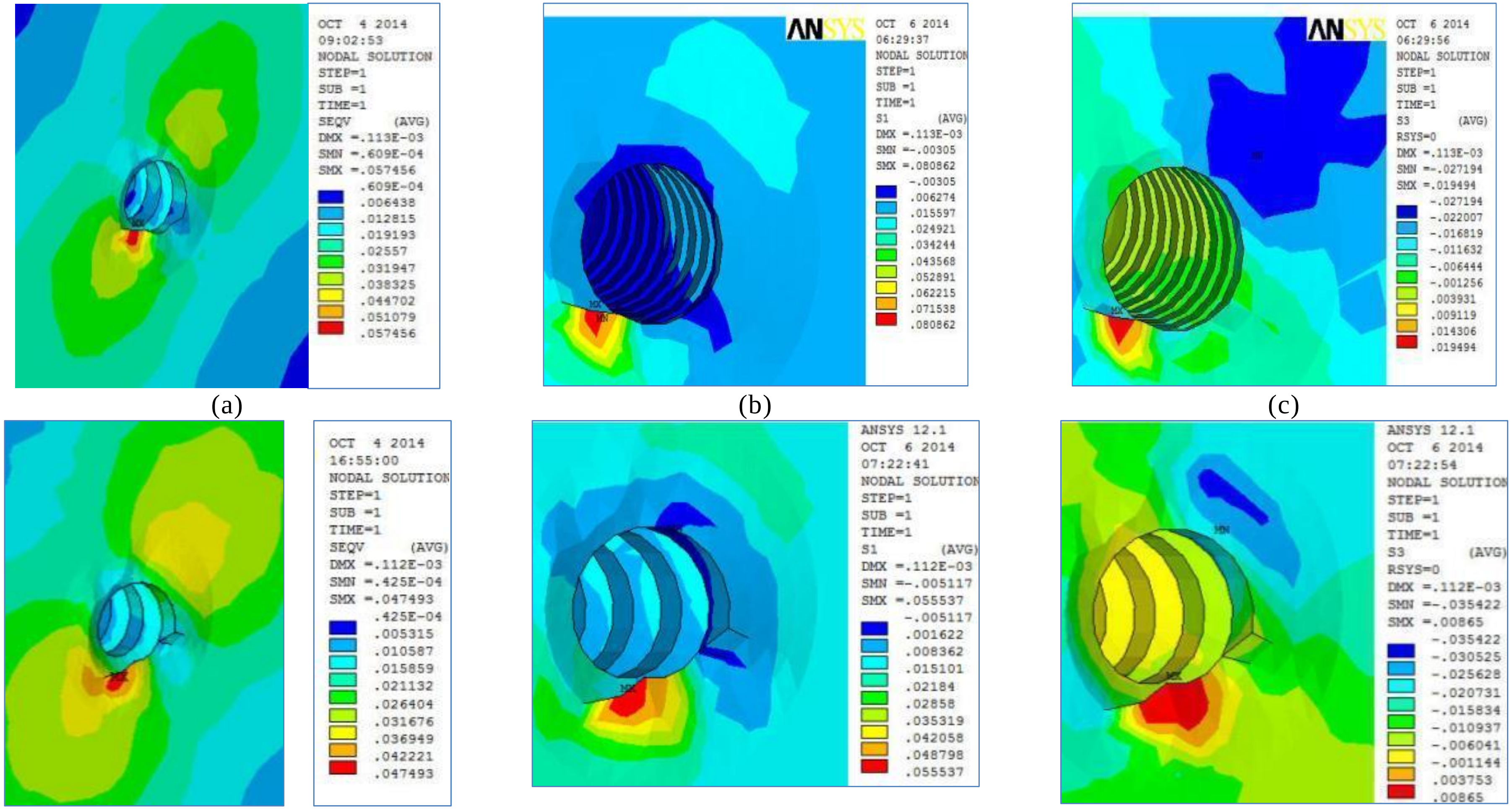



Fig-9: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in cancellous bone for control simulation

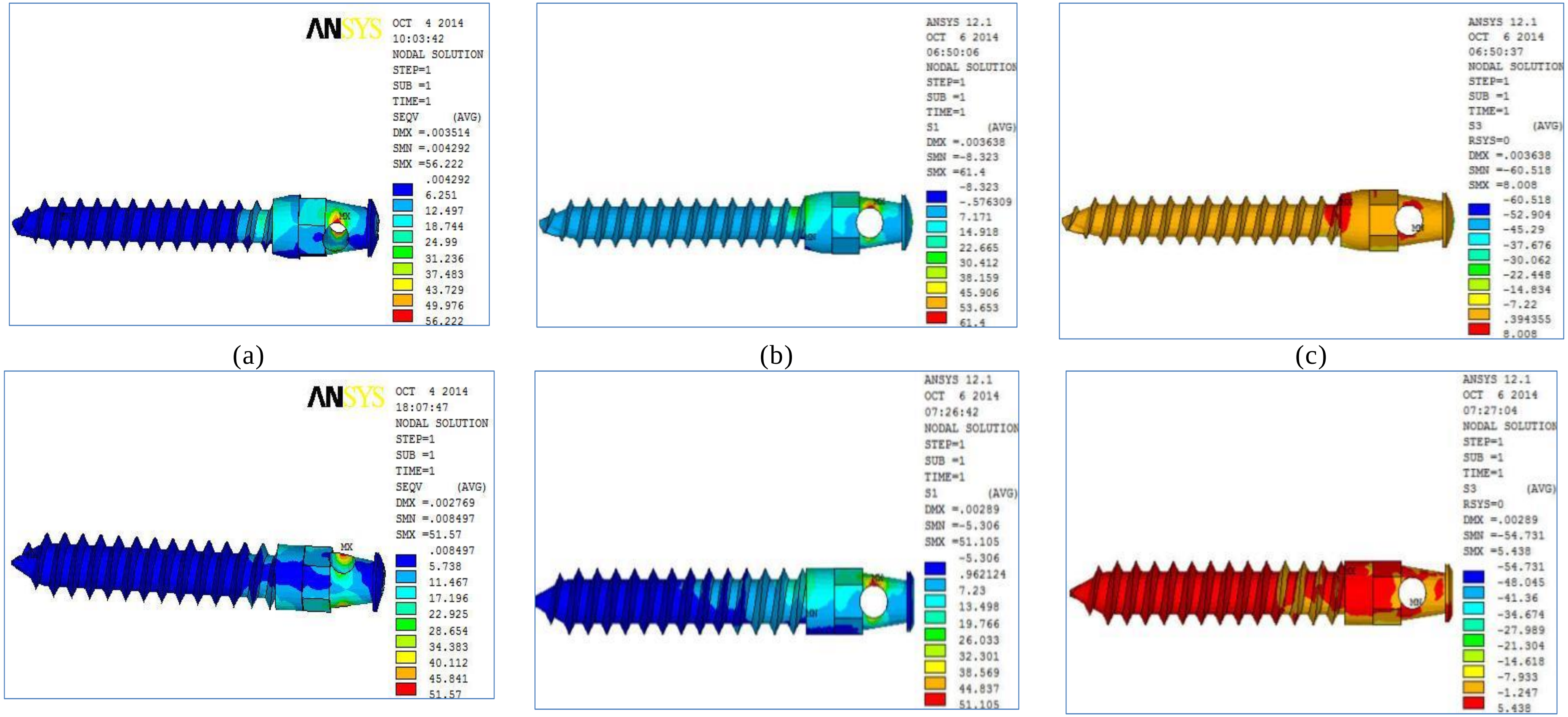



Fig-10: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in mini screw with diameter as variable

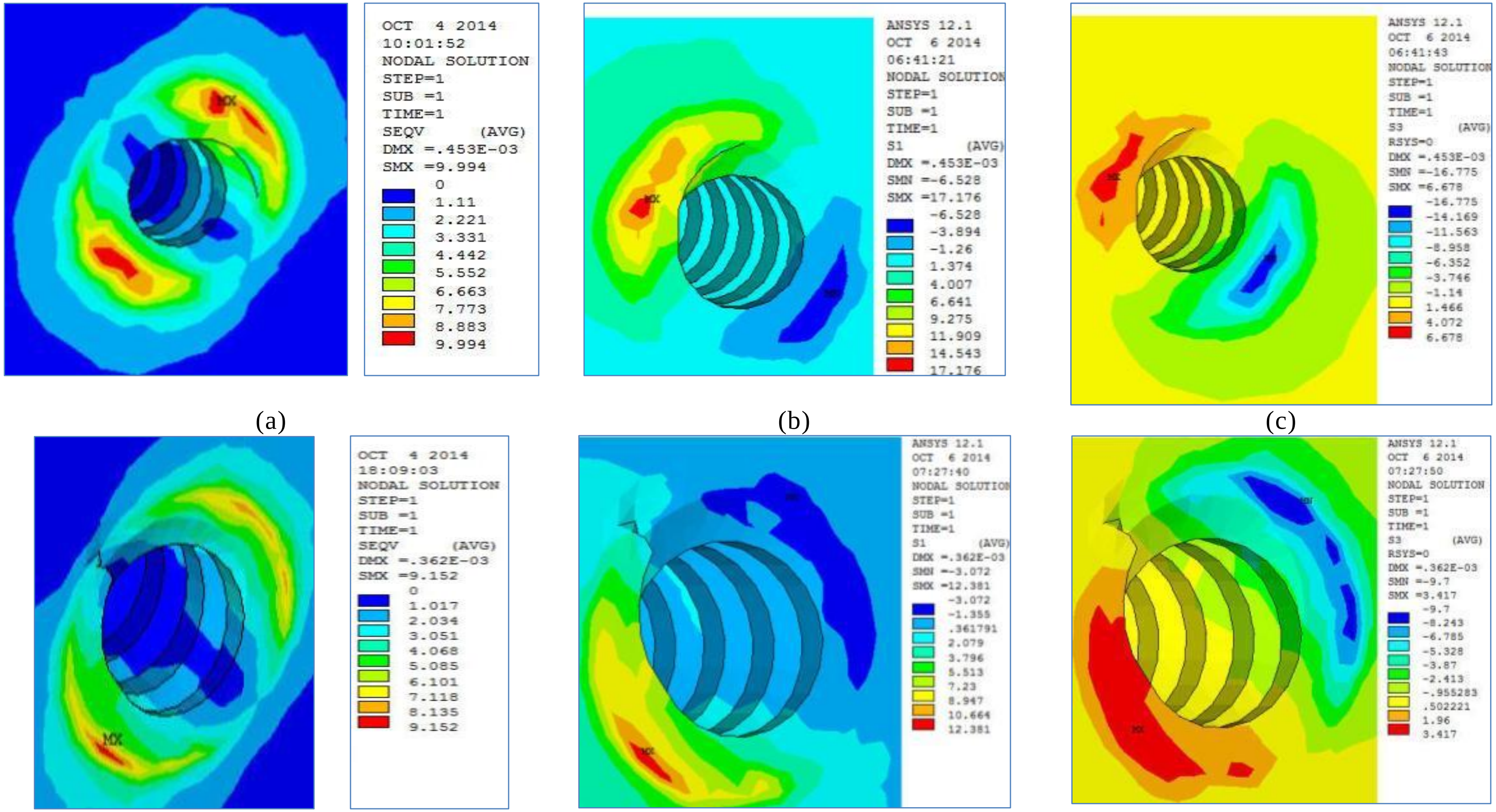



Fig-11: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in Cortical Bone with diameter as variable

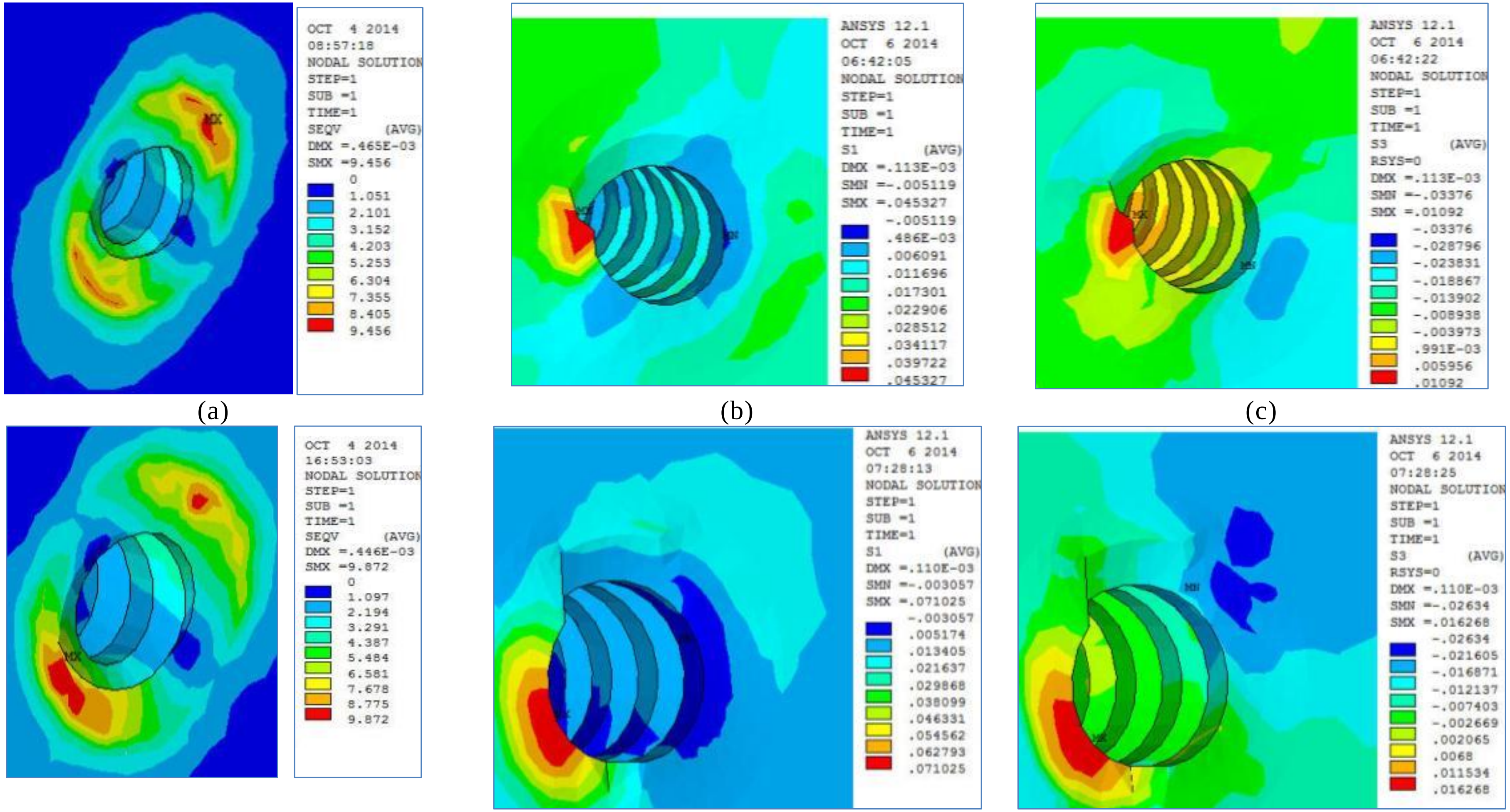



Fig-12: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in Cancellous Bone with diameter as variable

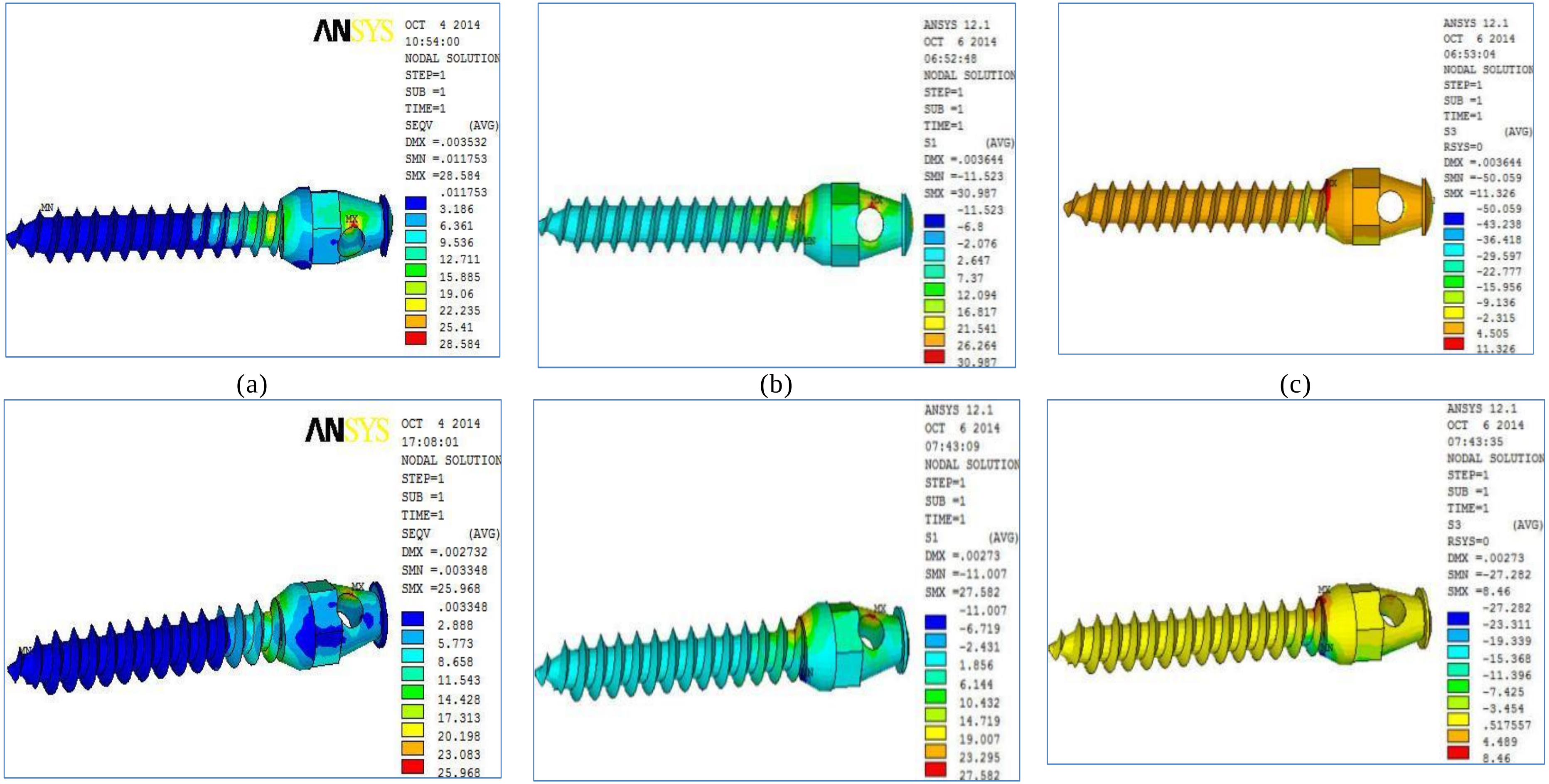



Fig-13: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in mini screw with Cortical bone thickness as variable

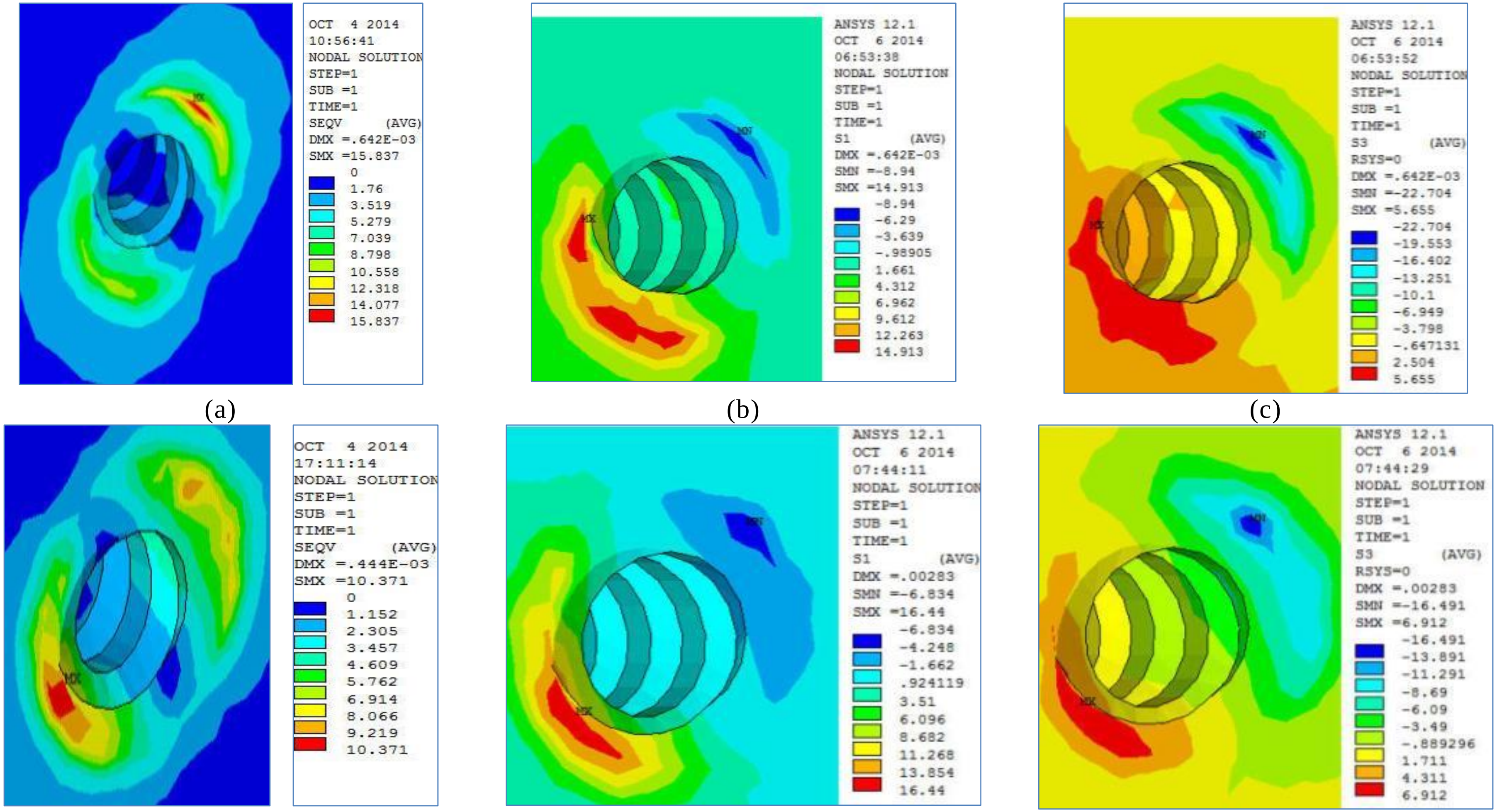



Fig-14: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in Cortical Bone with Cortical bone thickness as variable

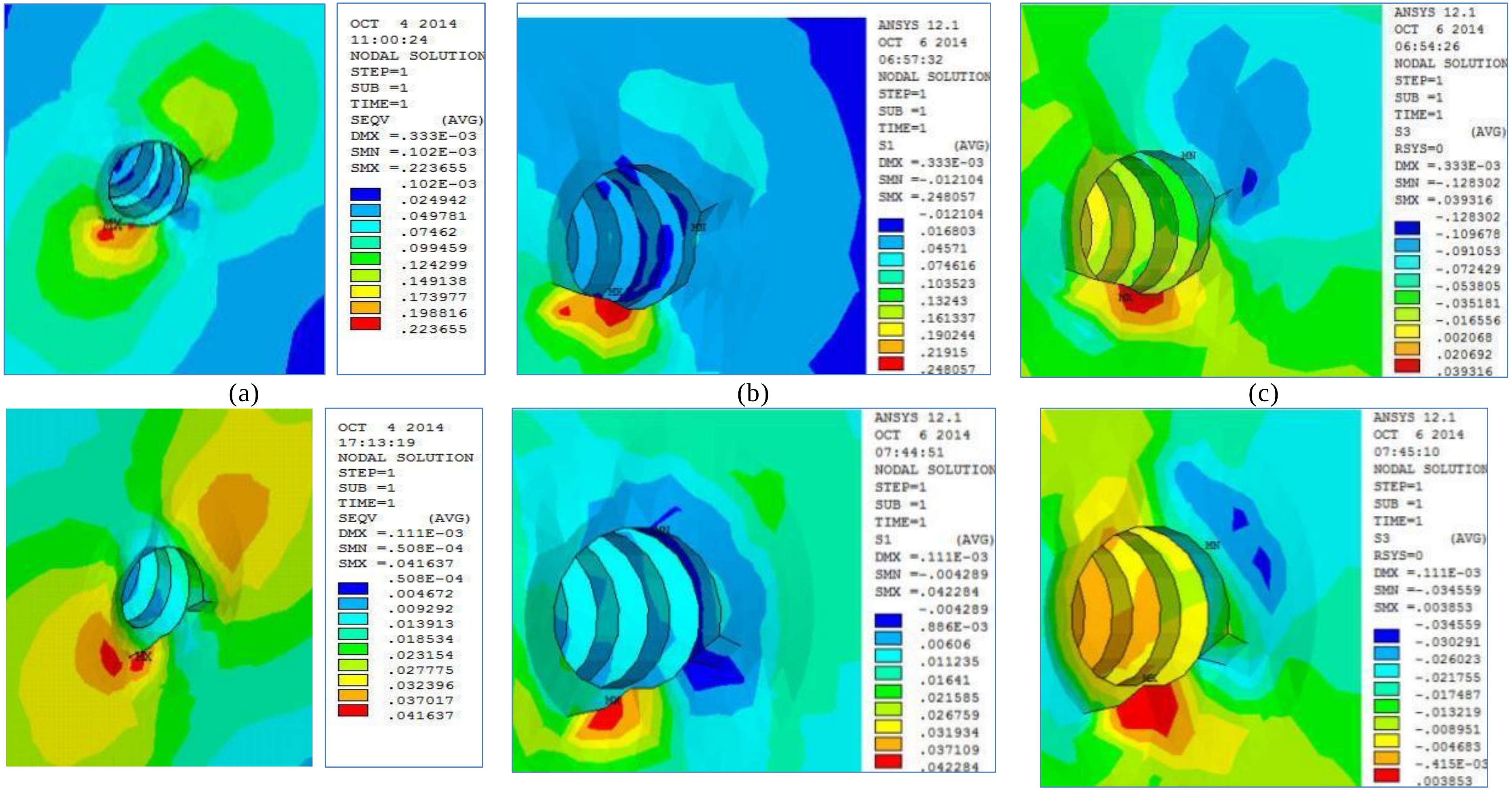



Fig-15: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in Cancellous Bone with Cortical bone thickness as variable

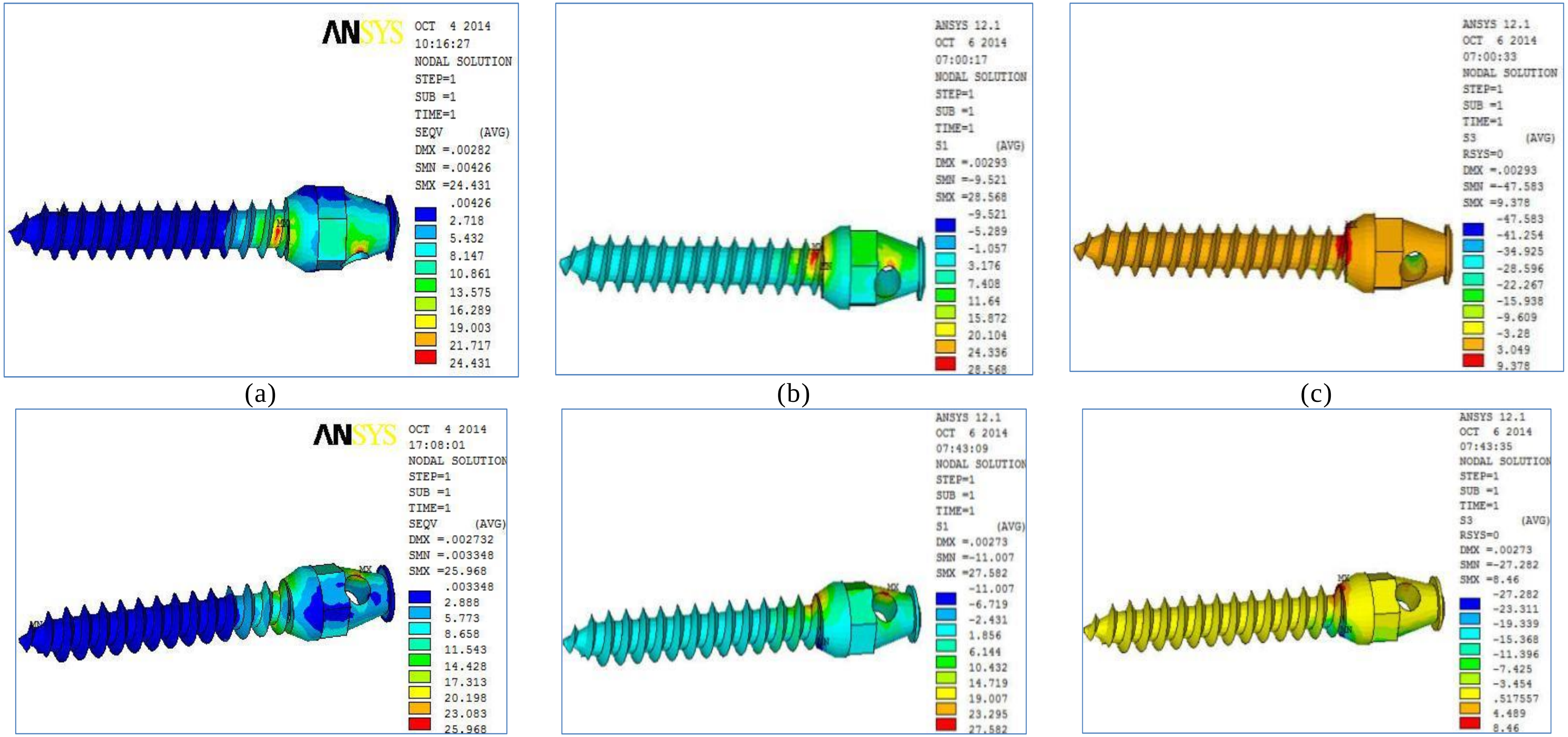



Fig-16: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in mini screw with Pitch of the mini screw as variable

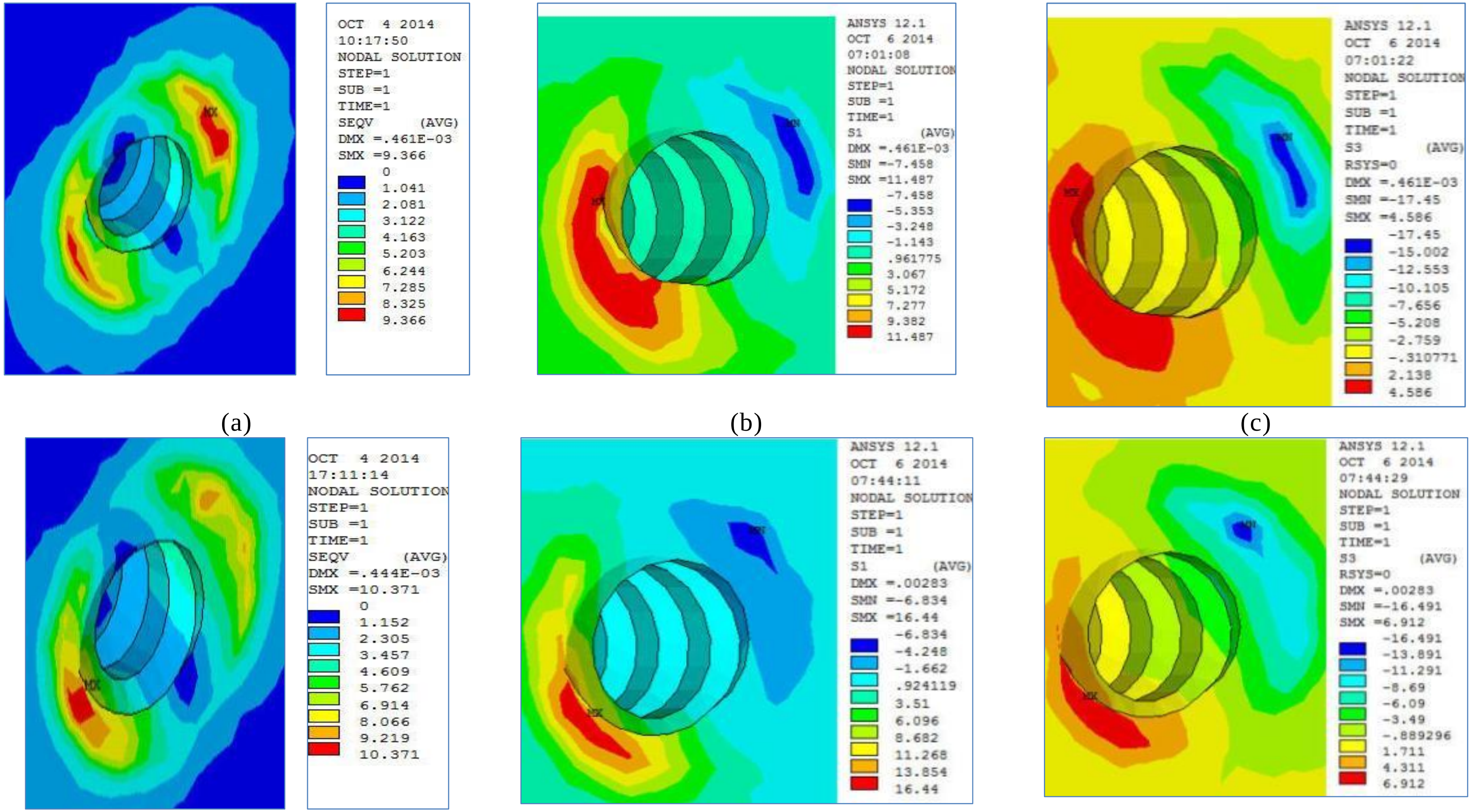



Fig-17: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in Cortical Bone with Pitch of the mini screw as variable

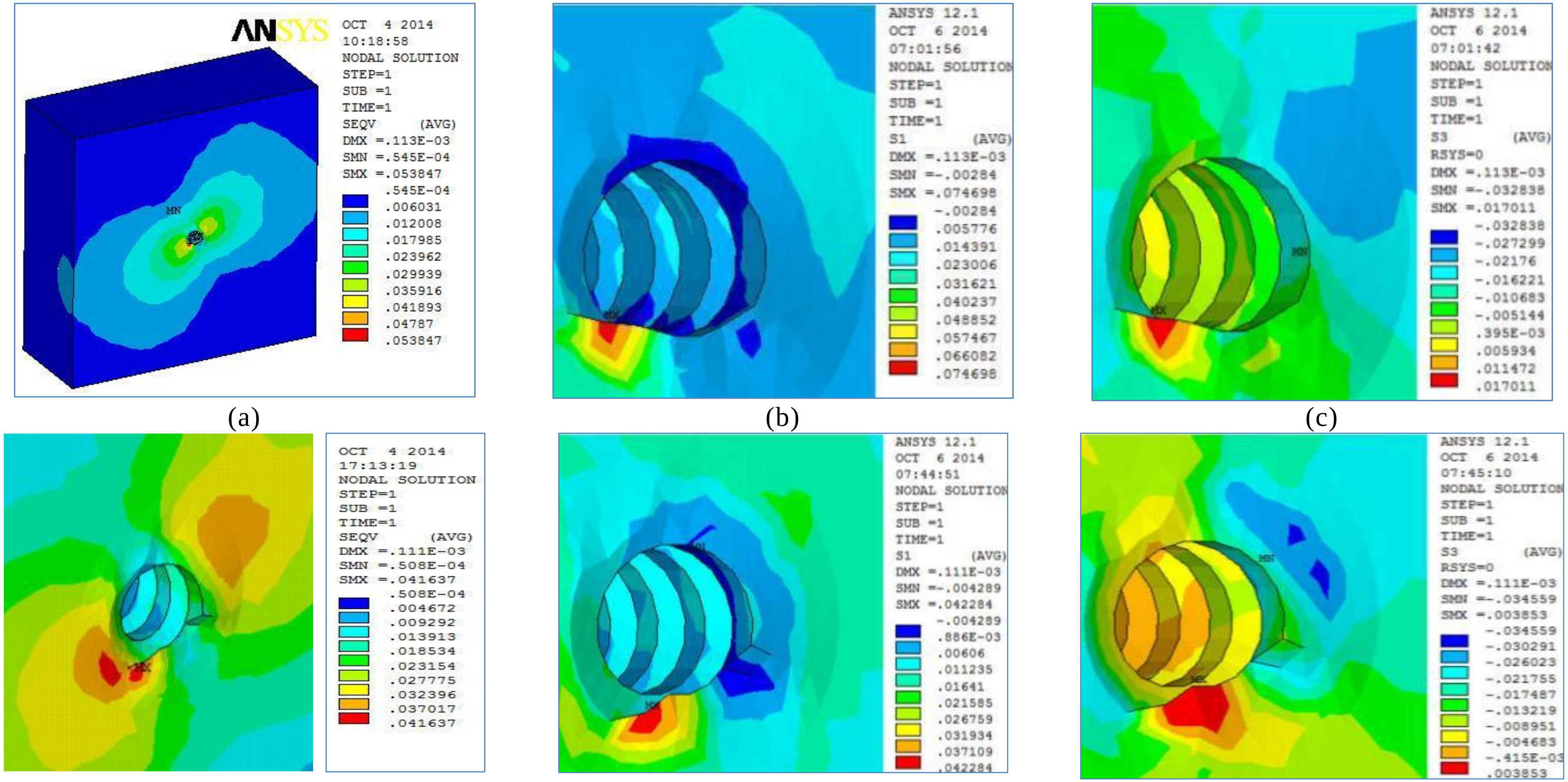



Fig-18: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in Cancellous Bone with Pitch of the mini screw as variable

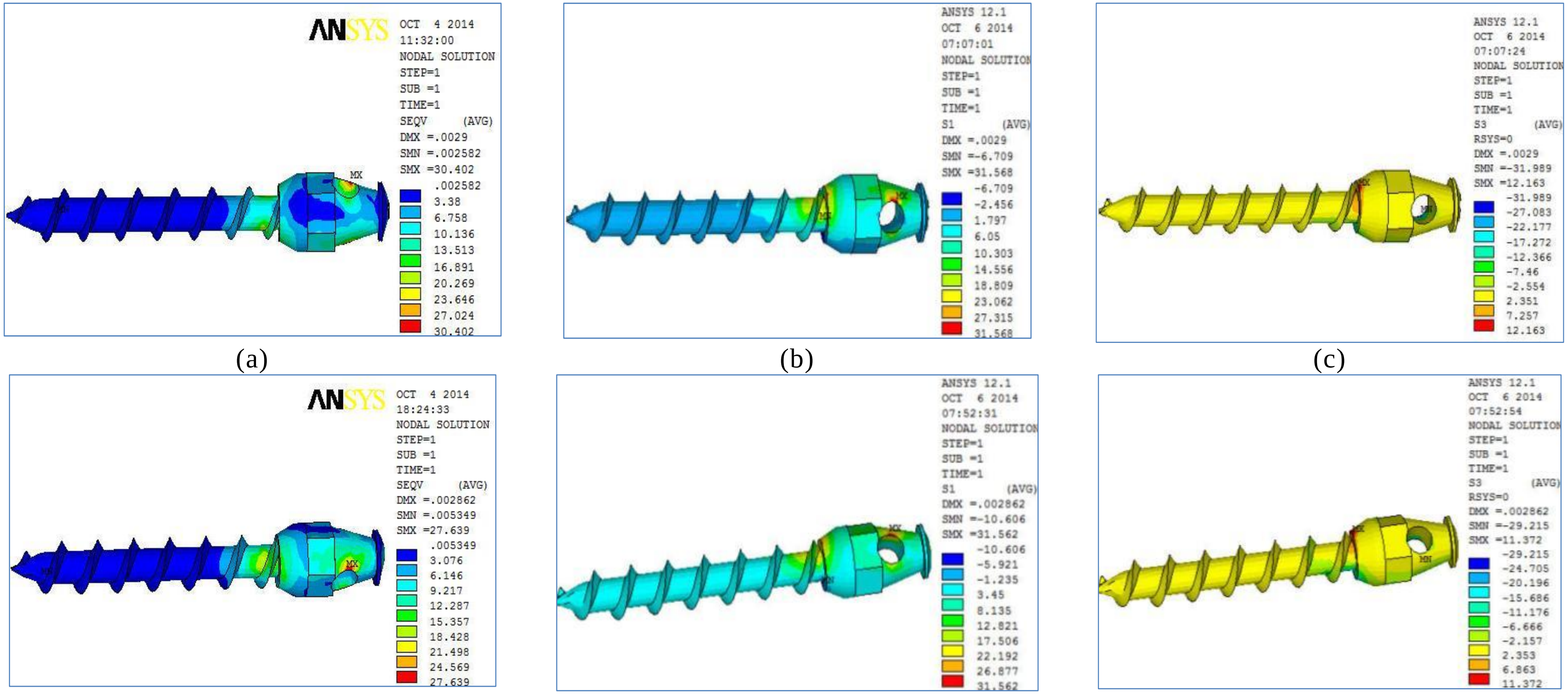



Fig-19: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in mini screw with Angulation of Force as variable

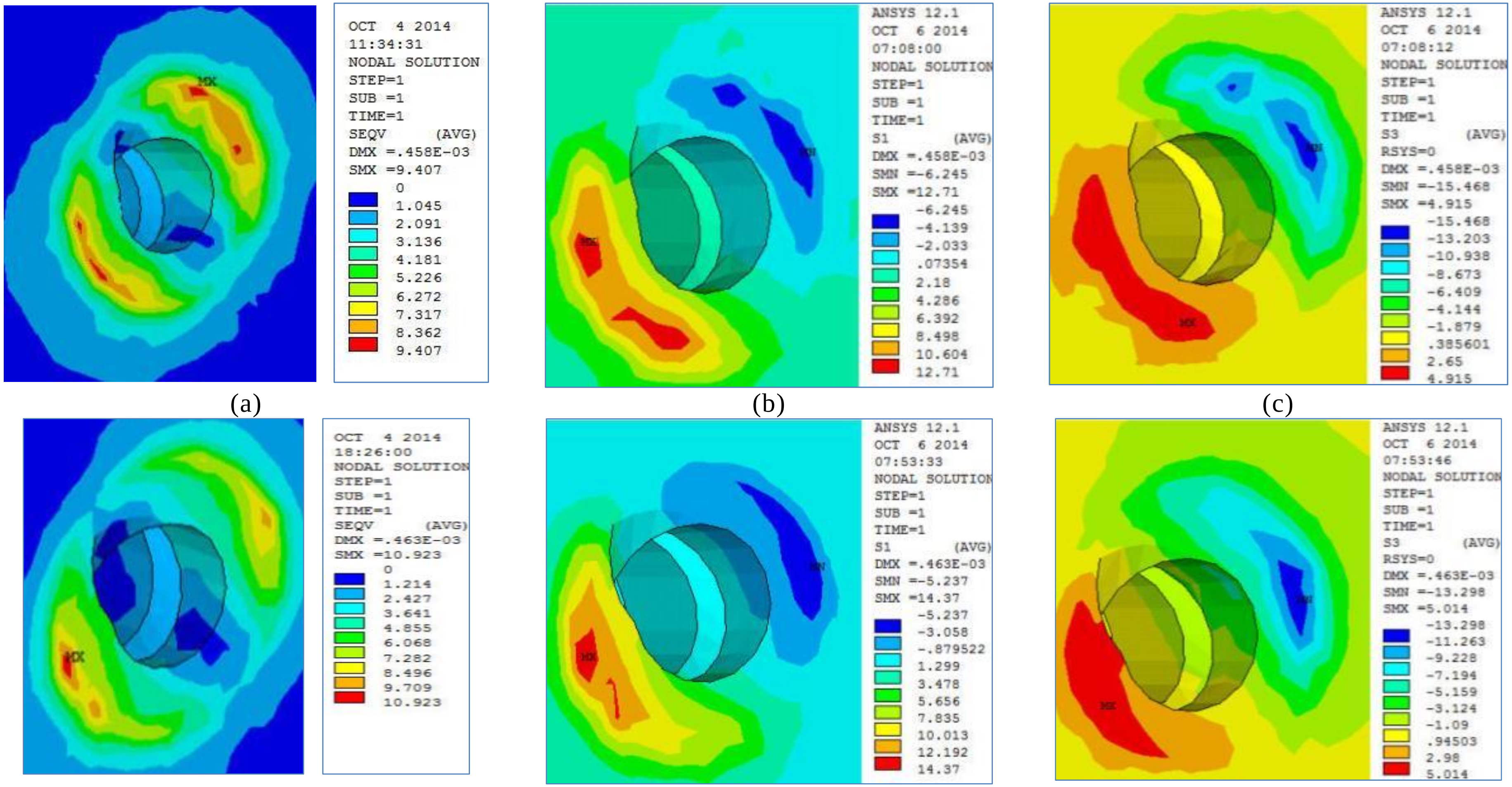



Fig-20: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in Cortical Bone with Angulation of Force as variable

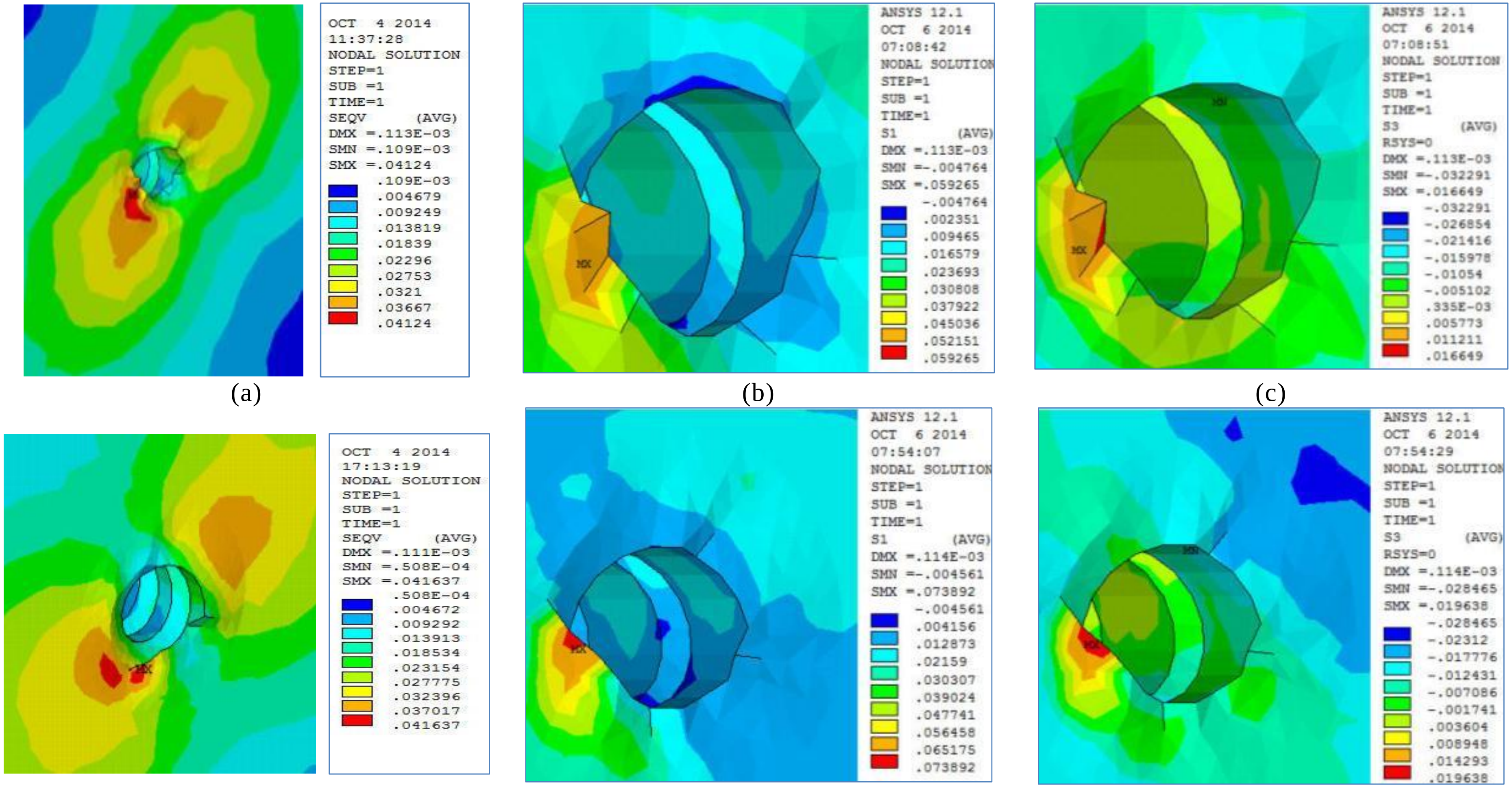



Fig-21: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in Cancellous Bone with Angulation of Force as variable

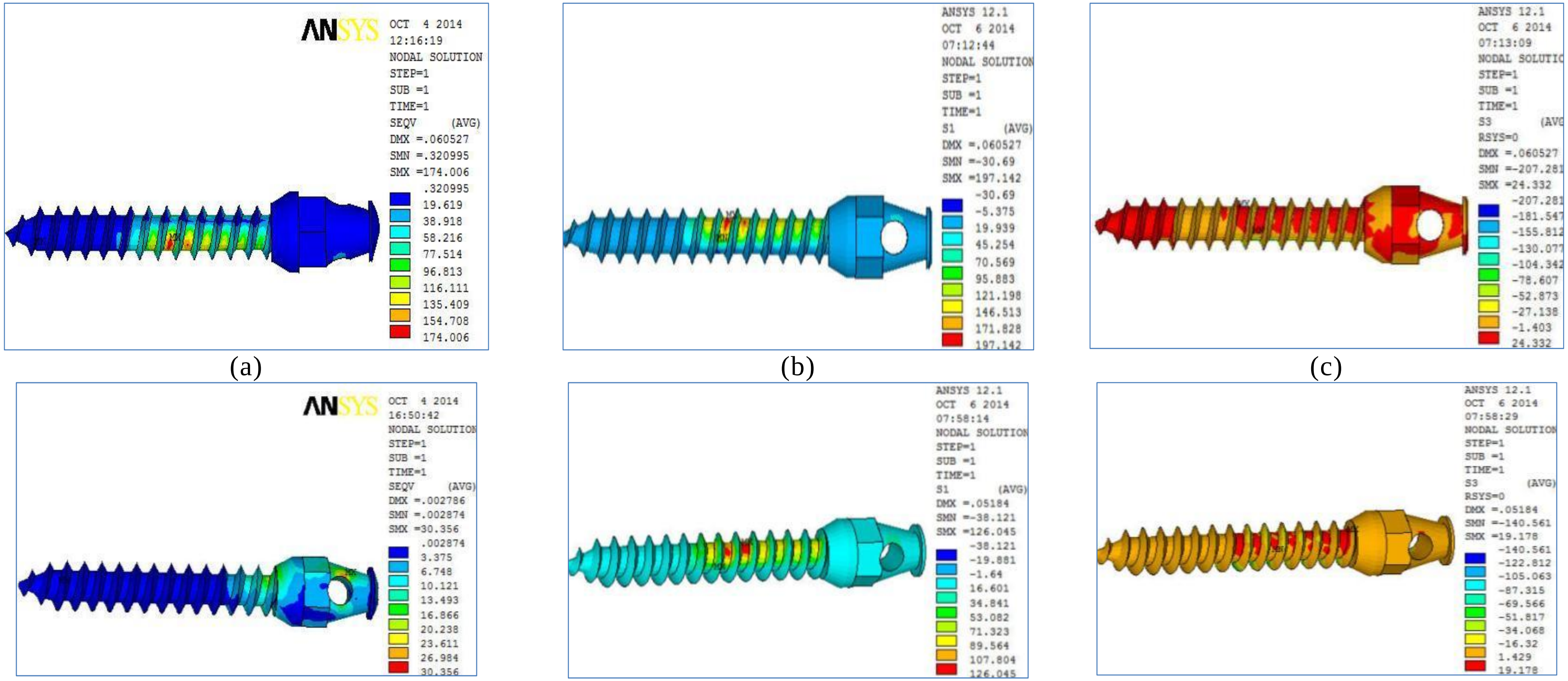



Fig-22: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in mini screw with Exposure Length of the mini screw as variable

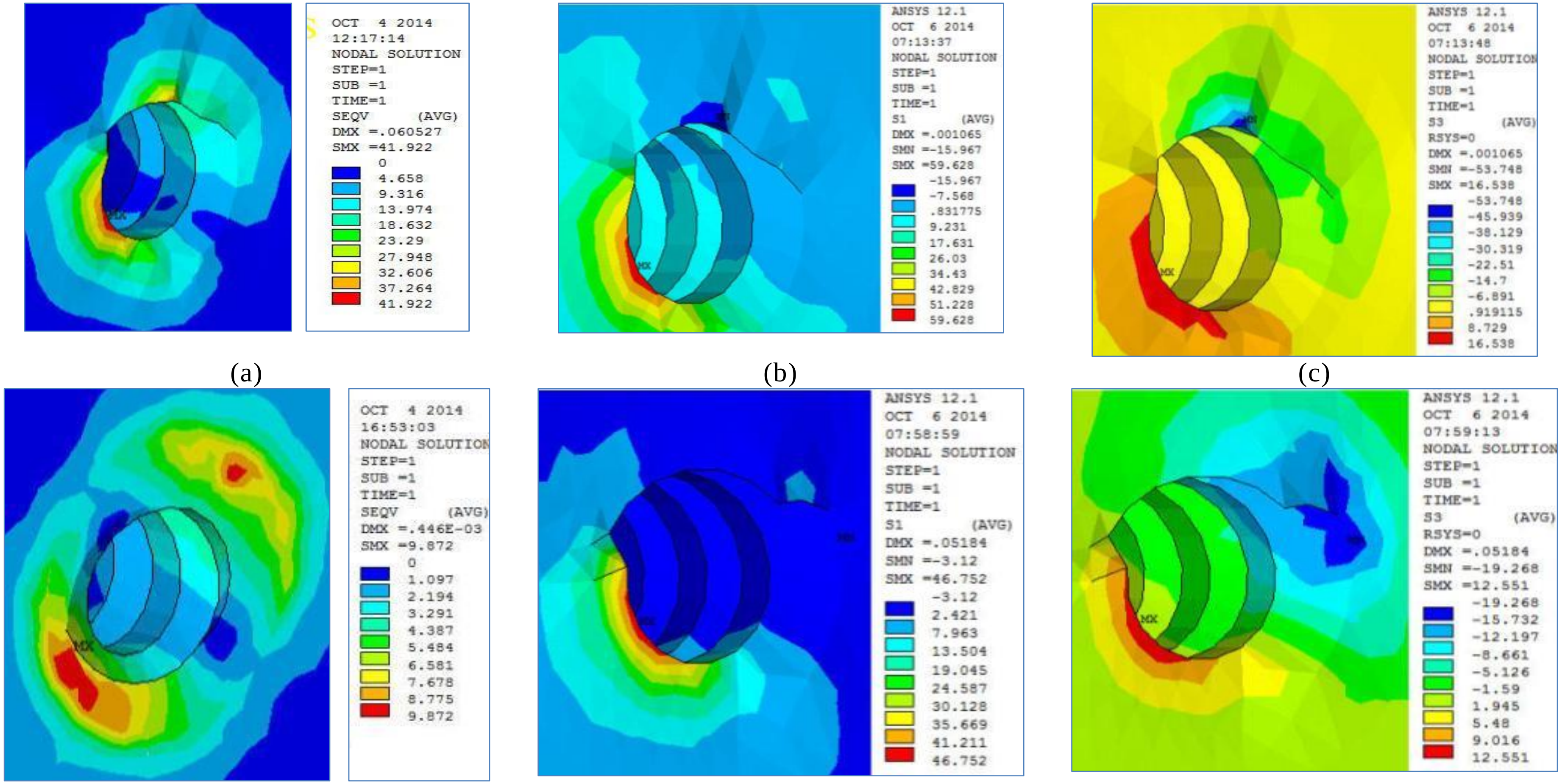



Fig-23: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in Cortical Bone with Exposure Length of the mini screw as variable

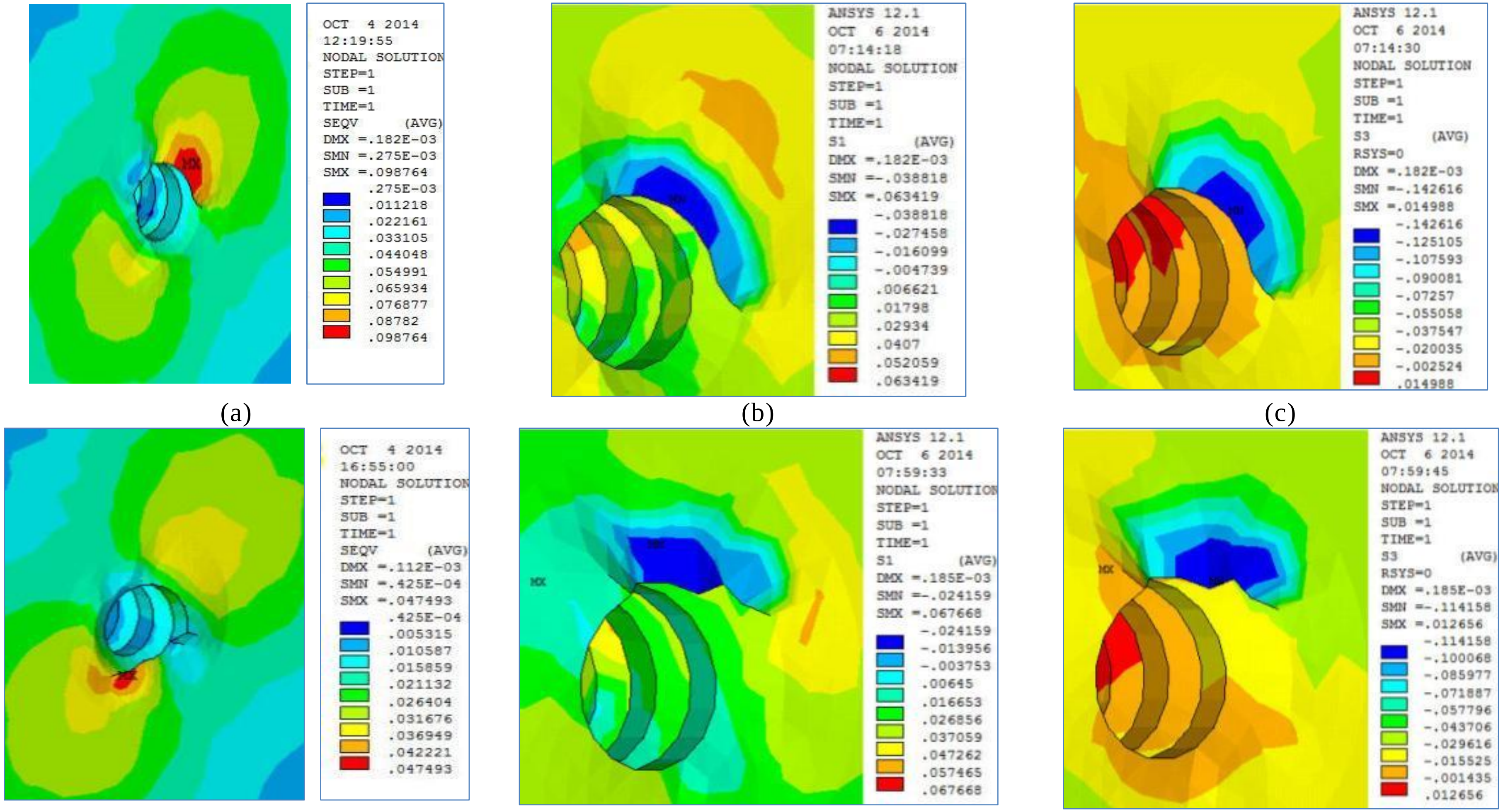



Fig-24: Stress plot showing (a) Vonmises (b) Maximum Principal (c) Minimum Principal stress between Group-I (Top Panel) and Group-II (Lower Panel) in Cancellous Bone with Exposure Length of the mini screw as variable

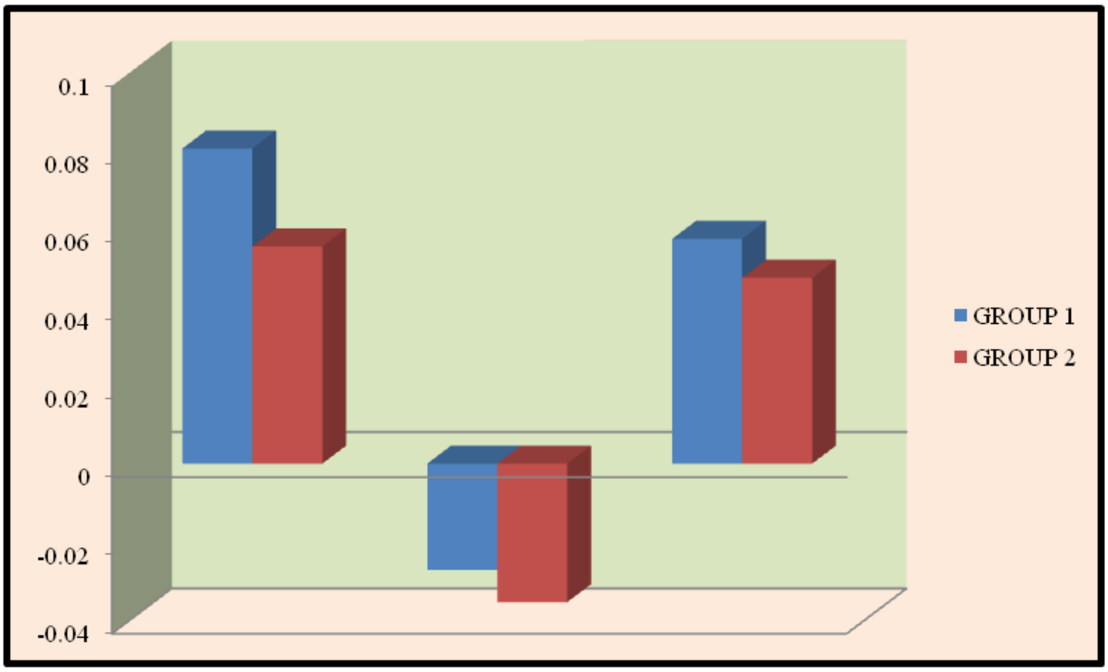



Fig-25: Comparison of Stress in Group-I and Group-II in Cancellous Bone for Control Simulations

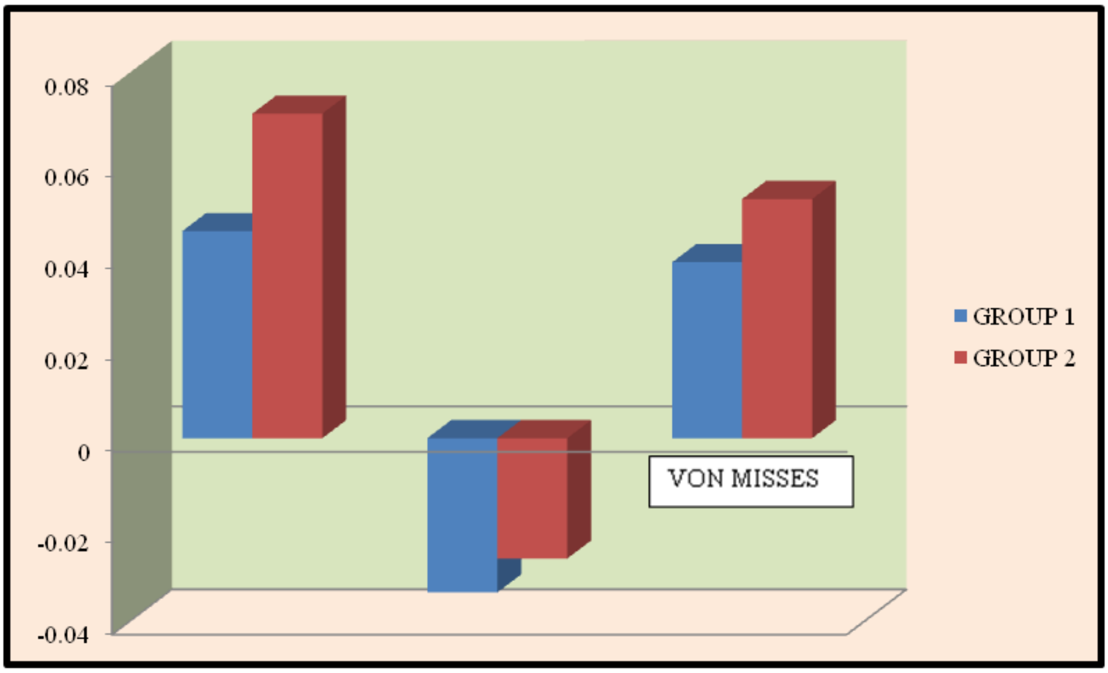



Fig-26: Comparison of Stress in Group-I and Group-II in Cancellous Bone with diameter as variable

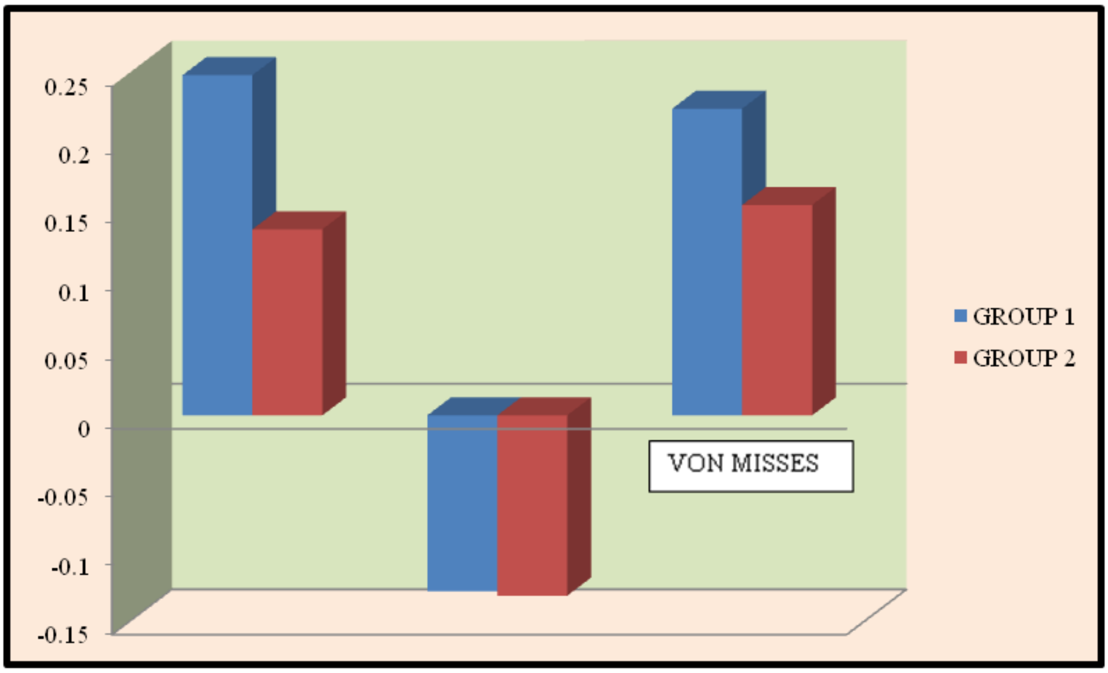



Fig-27: Comparison of Stress in Group-I and Group-II in Cancellous Bone with Cortical bone thickness as variable

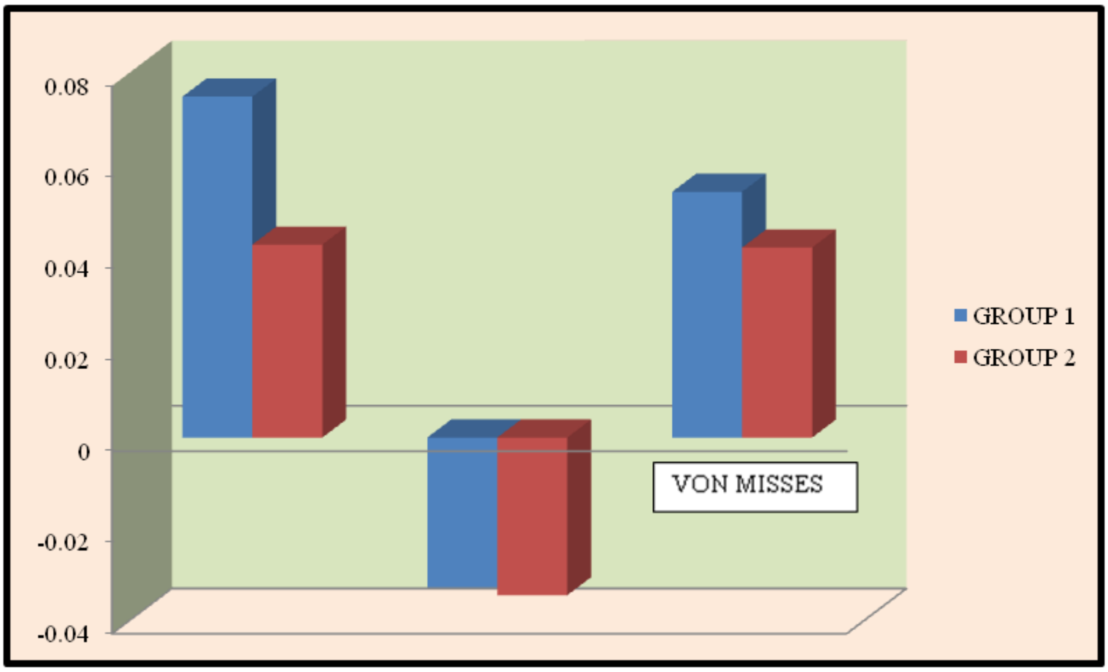



Fig-28: Comparison of Stress in Group-I and Group-II in Cancellous Bone with Pitch of the mini screw as variable

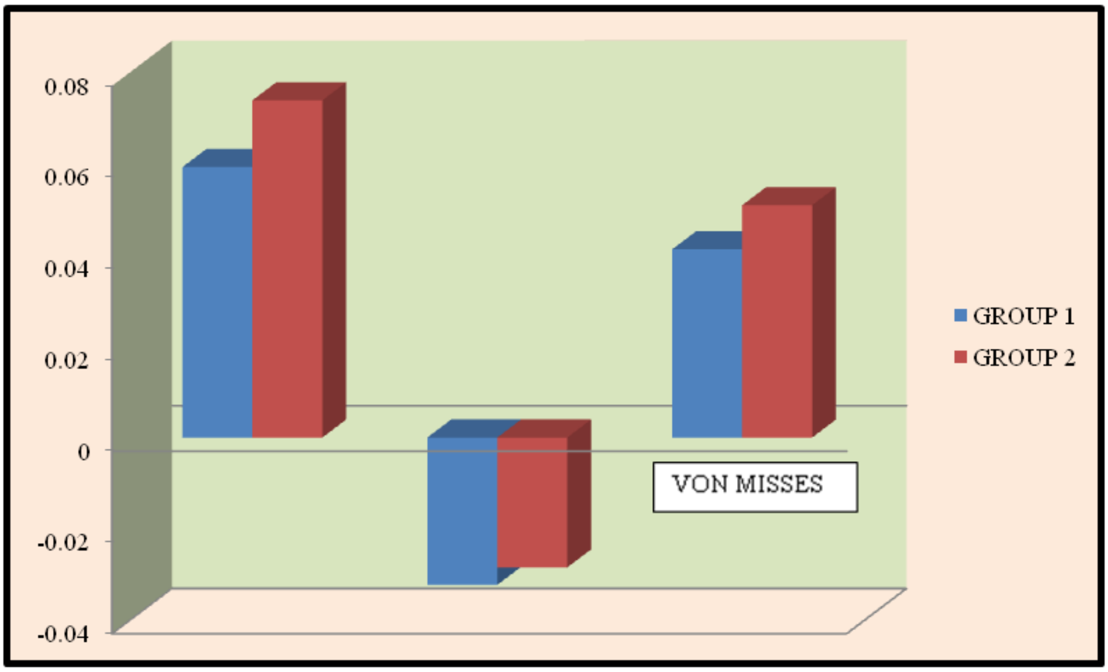



Fig-29: Comparison of Stress in Group-I and Group-II in Cancellous Bone with Angulation of Force as variable

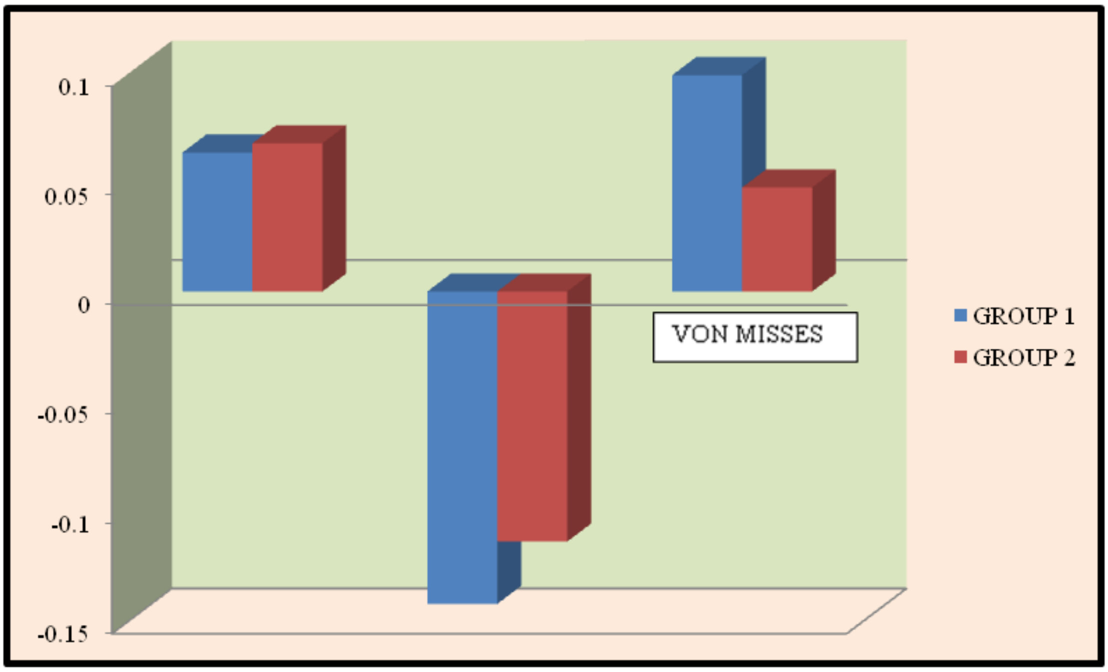



Fig-30: Comparison of Stress in Group-I and Group-II in Cancellous Bone with Exposure length of the mini screw as variable

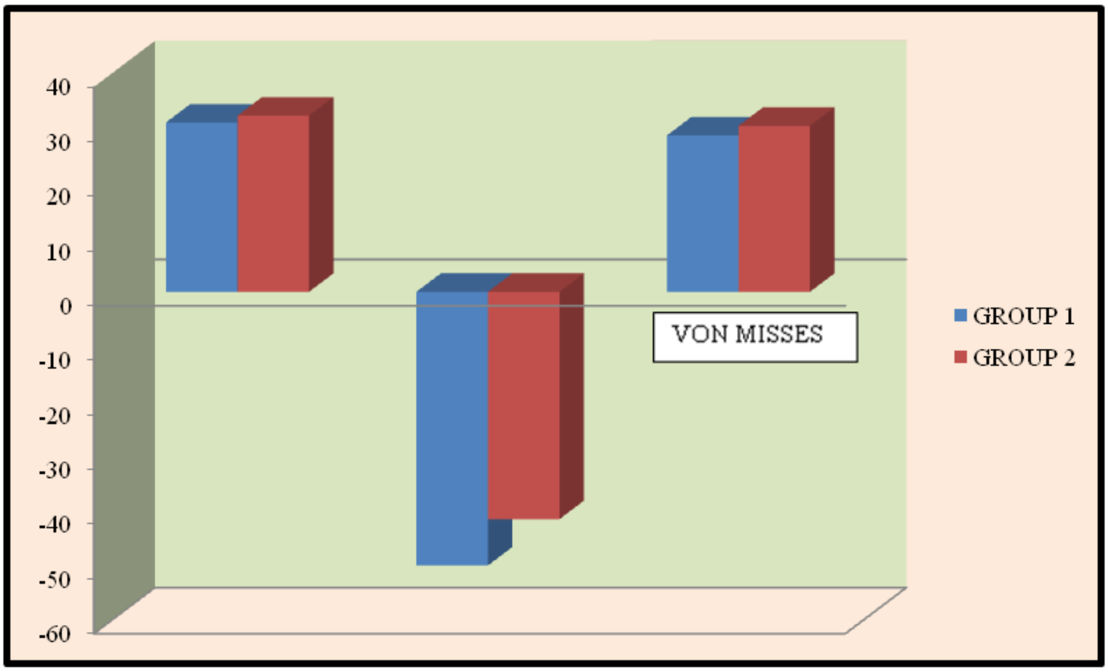



Fig-31: Comparison of Stress in Group-I and Group-II in Mini Screw for Control Simulations

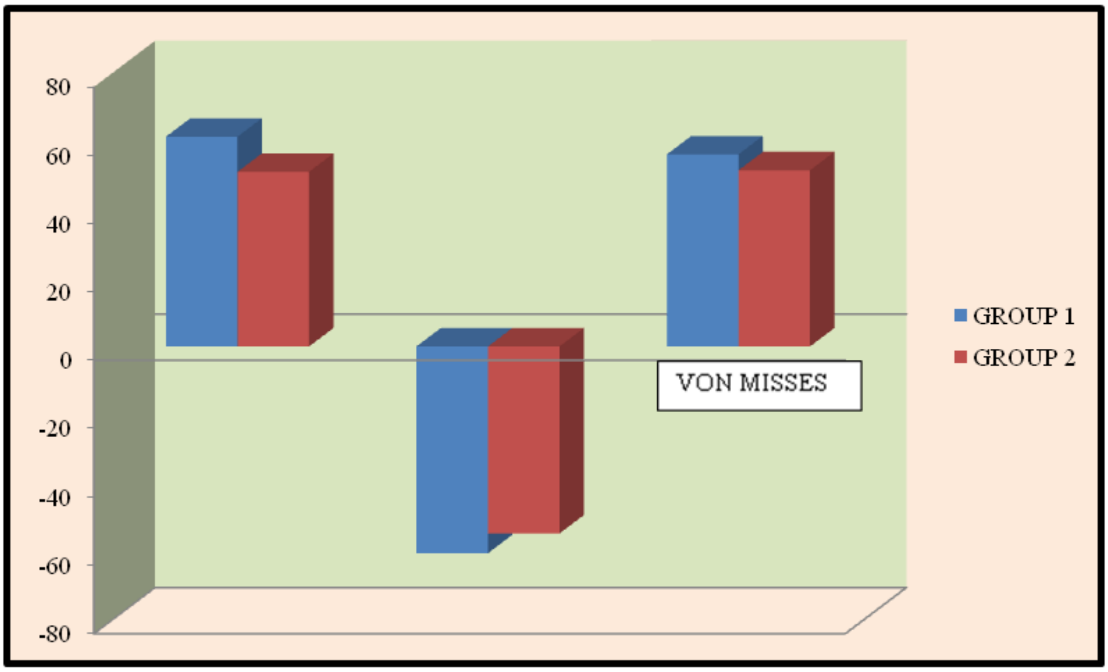



Fig-32: Comparison of Stress in Group-I and Group-II in Mini Screw with diameter as variable

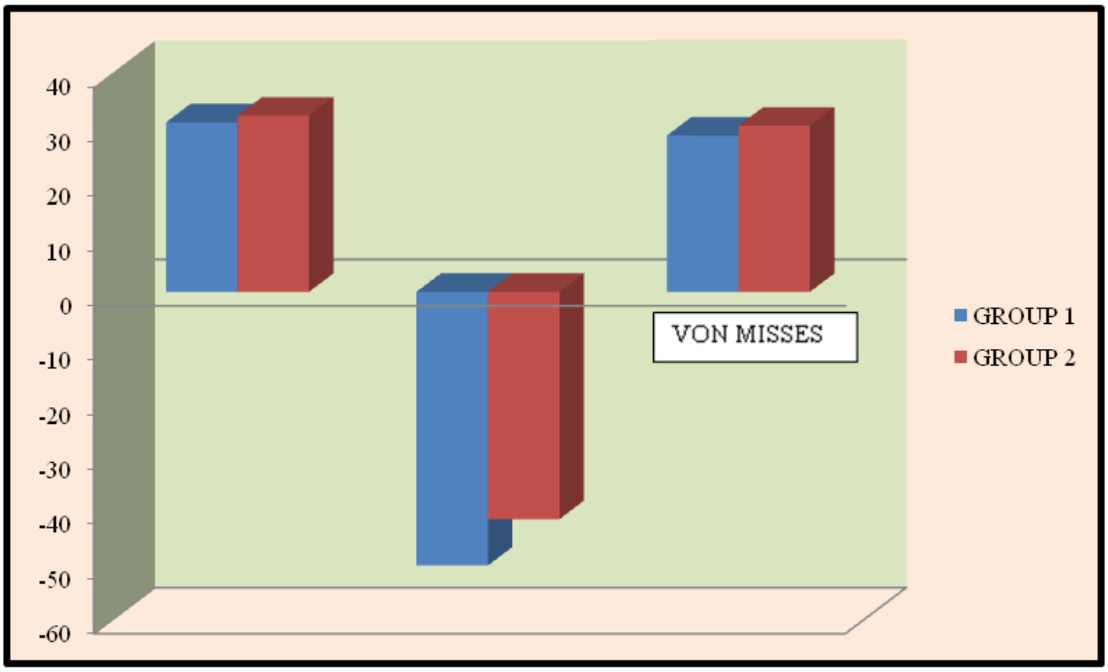



Fig-33: Comparison of Stress in Group-I and Group-II in Mini Screw with Cortical bone thickness as variable

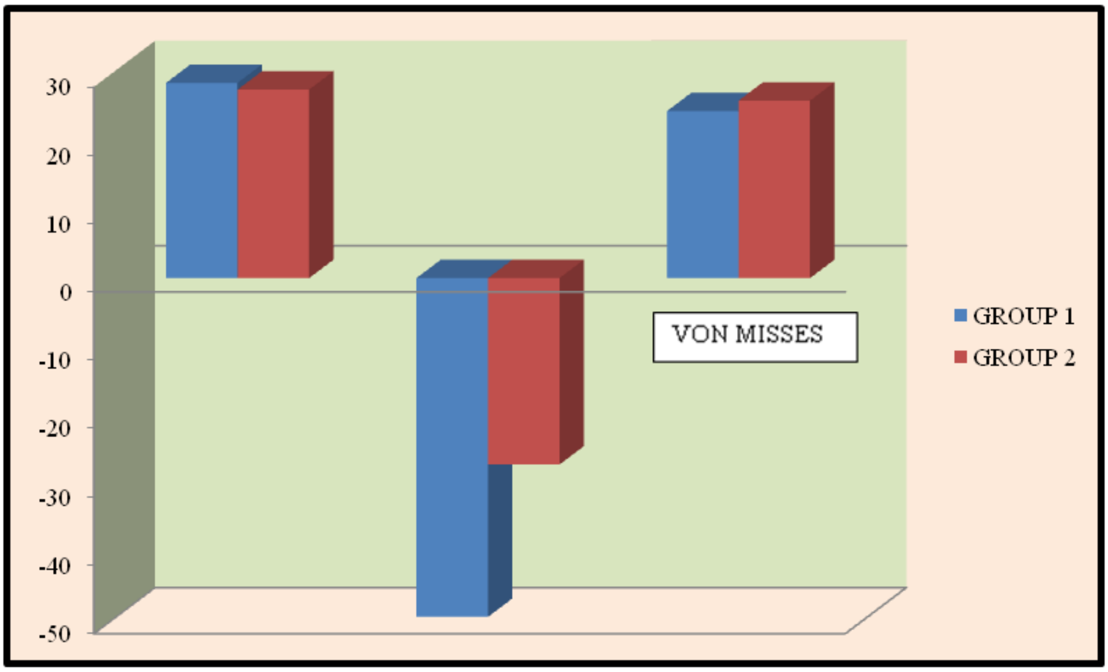



Fig-34: Comparison of Stress in Group-I and Group-II in Mini Screw with Pitch of the mini screw as variable

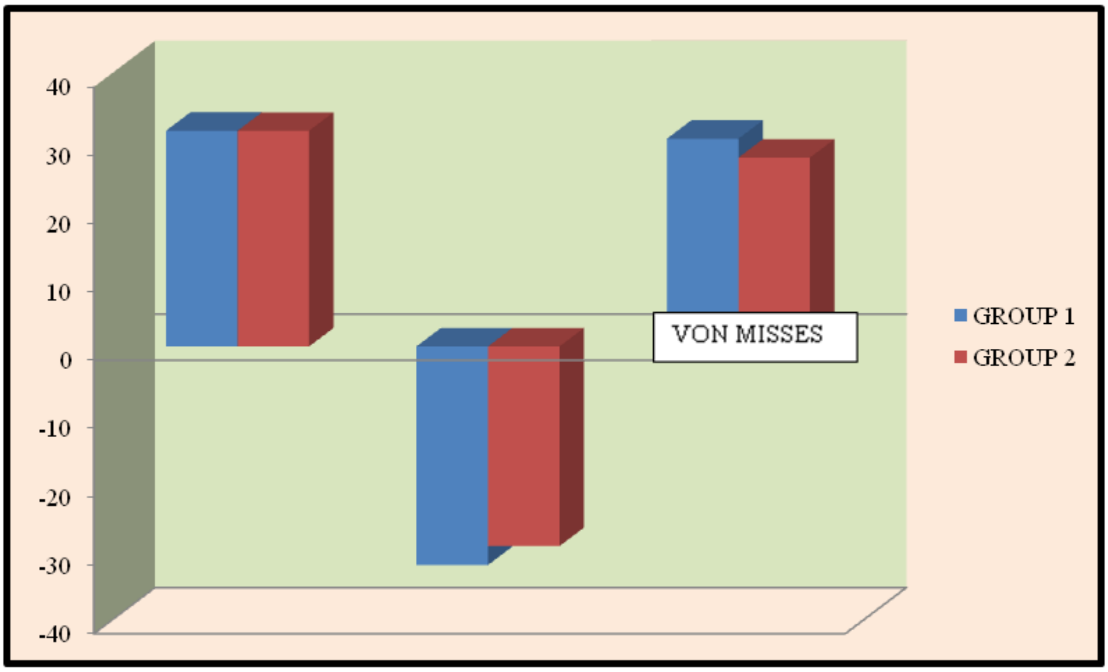



Fig-35: Comparison of Stress in Group-I and Group-II in Mini Screw with Angulation of Force as variable

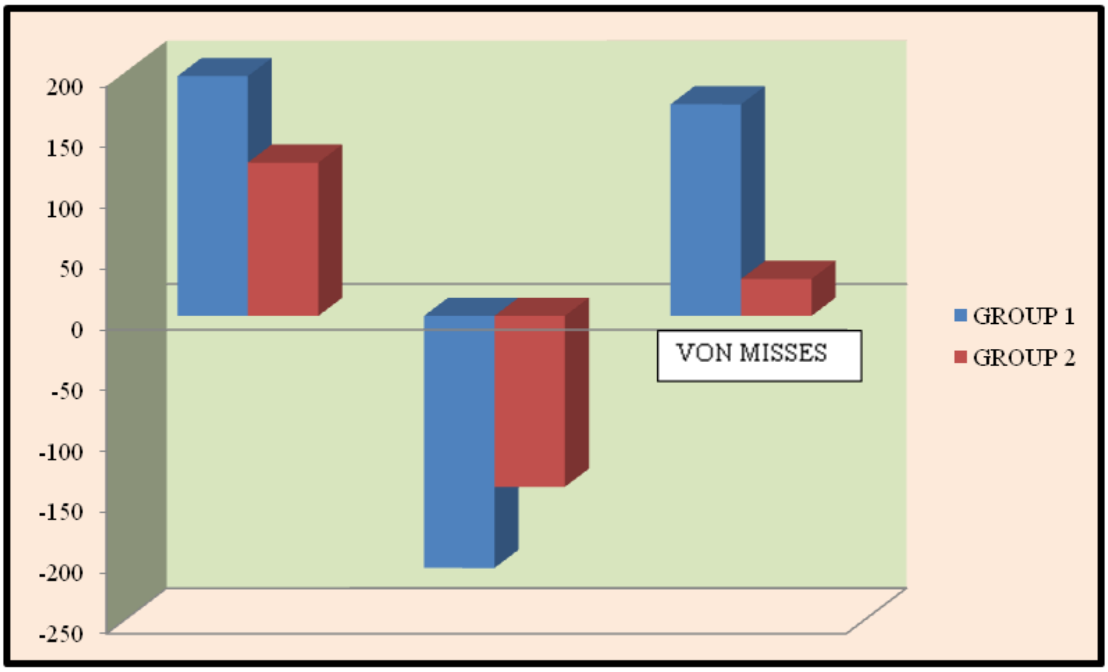



Fig-36: Comparison of Stress in Group-I and Group-II in Mini Screw with Exposure length of the mini screw as variable


Fig-37: Comparison of Stress in Group-I and Group-II in Cortical Bone for Control Simulations

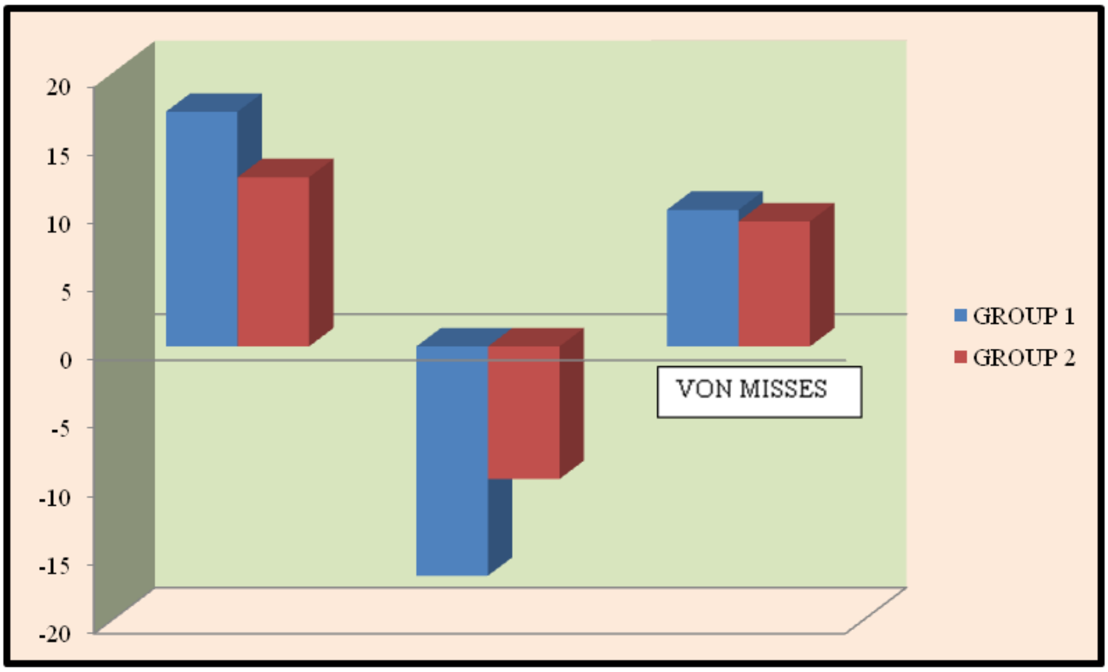



Fig-38: Comparison of Stress in Group-I and Group-II in Cortical Bone with diameter as variable

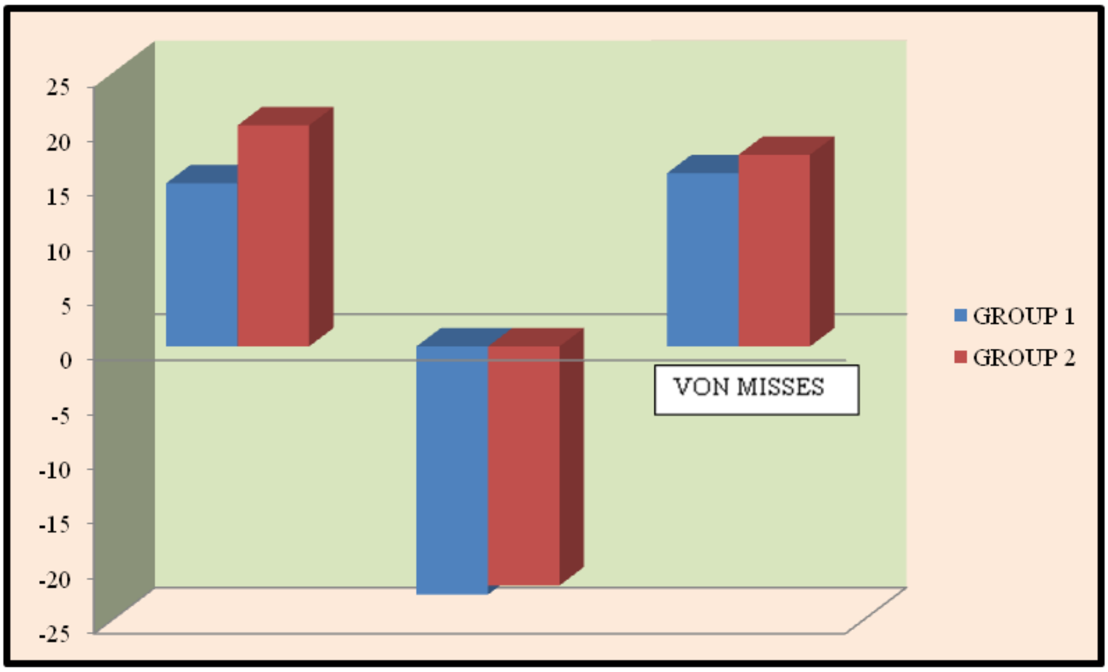



Fig-39: Comparison of Stress in Group-I and Group-II in Cortical Bone with Cortical bone thickness as variable


Fig-40: Comparison of Stress in Group-I and Group-II in Cortical Bone with Pitch of the mini screw as variable

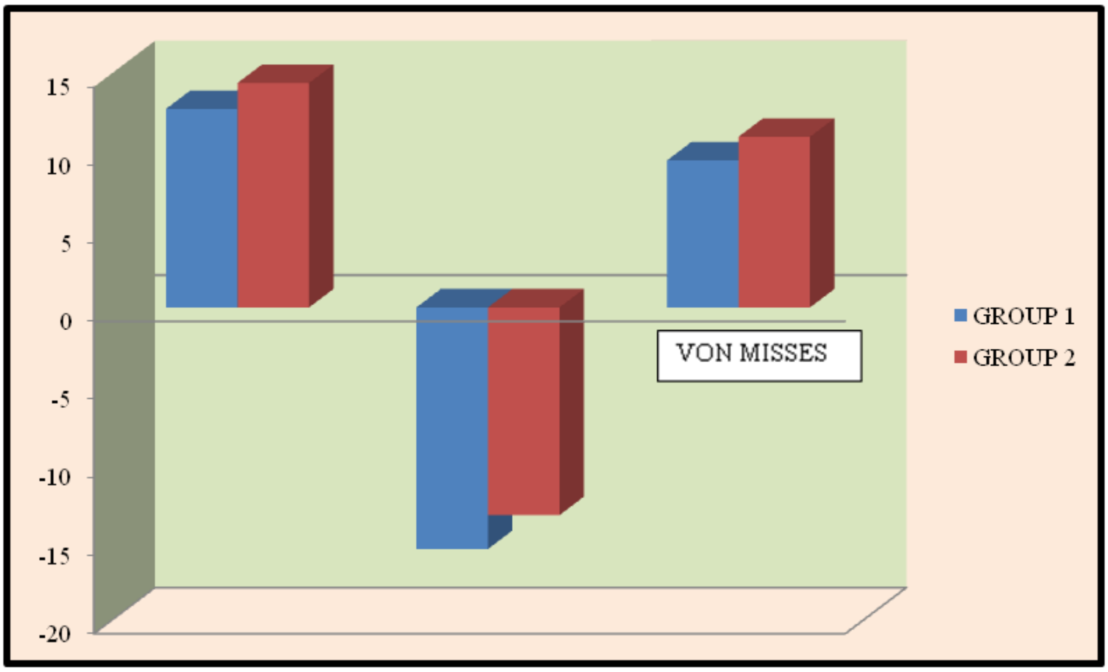



Fig-41: Comparison of Stress in Group-I and Group-II in Cortical Bone with Angulation of Force as variable


Fig-42: Comparison of Stress in Group-I and Group-II in Cortical Bone with Exposure length of the mini screw as variable

# DISCUSSION

Proper anchorage control is a prerequisite in orthodontic treatment to achieve successful outcome of an orthodontic therapy. Most of the time the orthodontist is confronted with control of reactionary forces which causes undesirable movement of the anchor teeth which is supposed to be stable. Alternative methods include a non-extraction approach[62], which by itself may not be applicable in all cases. Other methods include a headgear[63], intra-oral elastics where compliance may be an issue[64]. Differential moments may be applied but control of reactive forces may be difficult[65].

With the advent of miniscrews the above mentioned problems can be managed efficiently and has also broadened the scope of orthodontic treatment. Absolute anchorage is now possible because of the availability of these screws. Border line surgical cases can also be managed with miniscrews avoiding the possibility of orthognathic surgery thus reducing the subsequent morbidity and associated costs[66].

With the miniscrew embedded in the bone, biomechanics of the bone plays a very important role in the stability of a miniscrew[67]. Bone remodels in response to a mechanical load. Abnormal stresses around a bone may lead to a failure of the miniscrew. Optimal stress conditions in bone may lead to a favorable response in bone with subsequent retention of the miniscrew[55]. Therefore the present study is aimed at analyzing the stress distribution in bone around a miniscrew in response to various variables that are in play for a successful retention of a miniscrew in bone.

The study was done taking into consideration the various variables such as diameter, pitch, cortical bone thickness, force angulation and exposure length.

**Interpretation of results**

The results are interpreted as follows;

***I) Distribution & magnitude of stress in Group I***

A) Distribution & magnitude of stress in implant

B) Distribution & magnitude of stress in cortical bone

C) Distribution & magnitude of stress in cancellous bone

***II) Distribution & magnitude of stress in Group II***

Distribution& magnitude of stress in implant

Distribution & magnitude of stress in cortical bone

Distribution& magnitude of stress in cancellous bone

***III) Comparison of stress in Group I and Group II***

The distribution of maximum, minimum principle and von mises stress in implant, cortical and cancellous bone are presented in the form of color bands and the legend on the right side of each diagram indicates the magnitude of stress.

Application of stress in any body causes elongation or compression along the principal axes of the body in three dimensions namely x, y, z directions. By sign convention the first principle stress indicates the maximum tensile stress and the third principle stress indicates the maximum compressive stress acting perpendicular to the cross section in which the shear stress is zero. The von mises criterion combines all three principal stress into an equivalent stress when exceeds the yield strength of a material causes fracture of the material.

***I) Distribution & magnitude of stress in Group I:***

***A) Distribution& magnitude of stress in implant:***

***a) Maximum principal stress:***

The maximum principal stress was found to be near the point of application of force. The magnitude of stress was 30.975 MPa (Table 4). The first thread of the implant showed slightly lesser stress of 26.353 MPa (Figure 7b top panel) for the control simulation.

***Effect of variables:***

Increasing the diameter of the implant to 1.5 mm increases the stress value to 61.4 MPa (Table 5). The distribution of stress was similar to that seen in as the control value. Decrease in the cortical bone thickness to 1mm did not produce any significant decrease in the stress value 30.987 MPa (Table 6). The magnitude and stress pattern was similar to that of the control. Changing the angulation of force application to 60 degrees from 45 degrees and increase in pitch to 1mm from 0.5 mm produced similar stress values of 30.987 MPa and 31.568 MPa respectively (Table 7, 8). Increase in exposure length to 5mm dramatically increased the stress values to 197.142 MPa (Table 9). In contrast to other variables the distribution of stress was found to be in the implant thread embedded in the cortical bone.

***b) Minimum principal stress:***

The minimum principal stress was- 50.059 MPa (Figure7c, top panel) in the mesial surface of the implant and was seen at the first thread of the implant for the control.

***Effect of variables:***

Changing the diameter of the implant to 1.5mm increases the stress value to- 60.518 MPa (Table 5). In contrast to the control the minimum principal stress and was seen at the point of application of force. Changing the cortical bone thickness, angulation of force application showed no significant decrease in the stress values. However increase in pitch to 1mm and increase in exposure length to 5mm showed a decrease of -47.583 MPa (Table 7) and an increase of- 207.281 MPa (Table 9) respectively.

***c) Von Mises stress:***

The equivalent stress was found to be 28.598 MPa (Figure 7a, top panel) and was seen at the point of application of force.

***Effect of variables:***

Increasing the diameter increase the stress value to 56.222 MPa (Table 5). Cortical bone thickness, pitch, force angulation did not result in significant changes in the stress value. Increasing exposure length resulted in an increased value of 174.006 MPa. In all the simulations the equivalent stress was seen at the point of application of force except in the case of increased exposure length which was seen at implant thread embedded in the cortical bone.

### *B) Distribution & magnitude of stress in cortical bone:*

#### *a) Maximum principal stress:*

The maximum principal stress 11.478 MPa (Figure 8b, top panel) was seen distal to the implant for the control simulation. The stress pattern was seen as a localized band distal to the implant surrounded by a decreasing concentric gradient of stress pattern.

#### *Effect of variables:*

Increasing the diameter of the implant to 1.5mm increased the stress values to 17.176 MPa (Table 5). Decreasing the cortical bone thickness to 1mm also increased the stress values to 14.913 MPa though not as much increase when compared to that of increasing the diameter. Force angulation and pitch did not result in a significant change in the values of stress. Increase in exposure length to 5mm increased the stress value to 59.628 MPa.

#### *b) Minimum principal stress:*

The stress value of -17.596 MPa (Figure 8c, top panel) was seen in the mesial surface of the bone for the control. The stress pattern was similar to that of the maximum principal stress except that it was seen in the mesial surface of the bone.

#### *Effect of variables:*

Diameter of the implant in this case did not result in increase in stress values -16.775MPa (Table 5). Force angulation and pitch of the implant also did not result in an increase of stress values. Decreasing cortical bone thickness increased the stress values -22.704 MPa (Table 6) so as increased exposure length -53.748 MPa (Table 9). Increasing pitch of the implant decreased the stress values to -15.468 MPa.

***c) Von mises stress:***

A stress value of 28.598 MPa (Figure 7a, top panel) was obtained. The pattern of distribution was seen on both the mesial and distal surface of the bone.

***Effect of variables:***

The magnitude of stress was increased in the case of diameter 56.222 MPa (Table 5) and increased exposure length 174.006MPa (Table 9). All the other variables resulted in similar stress as that of control. The pattern of distribution of stress was similar in all simulations.

***C) Distribution & magnitude of stress in cancellous bone:***

***a) Maximum principal stress:***

Stress value of 0.080562 MPa (Figure 9c,top panel) was obtained for the control simulation. The pattern of stress was more localized than that seen in cortical bone.

***Effect of variables:***

An increase in diameter to 1.5mm resulted in increase in stress to 0.045327 MPa (Table 5). Reducing the cortical bone thickness resulted in a very significant increase in stress value to 0.248057 MPa (Table 6). Angulation of force, pitch and exposure length did not result in a significant increase of stress values compared to control.

***b) Minimum principal stress:***

A stress value of -0.027194 MPa was obtained (Figure 9c, top panel) .

***Effect of variables:***

Cortical bone thickness -0.128302 MPa (Table 6) and exposure length -0.142616 MPa (Table 9) had the highest impact on stress values when compared to the control. The other variables stress level was similar to that of control.

***c) Von mises stress:***

The equivalent stress was found to be 28.598 MPa (Figure 9a, top panel) and was seen at the point of application of force.

***Effect of variables:***

Increasing the diameter increase the stress value to 51.57 MPa (Table 5). Cortical bone thickness, pitch, force angulation did not result in significant changes in the stress value. Increasing exposure length resulted in an increased value of 76.68 MPa. In all the simulations the equivalent stress was seen at the point of application of force except in the case of increased exposure length which was seen at implant thread embedded in the cortical bone.

***II) Distribution & magnitude of stress in Group II:***

***A) Distribution & magnitude of stress in implant:***

***a) Maximum principal stress:***

A stress value of 32.23 MPa (Figure 7b, lower panel) was obtained for the control simulation. The area near the point of application of force showed the maximum stress value.

***Effect of variables:***

An increase in diameter to 1.5mm and increased exposure length to 5mm raised the stress values to 51.105 MPa (Table 6) and 126.045 MPa (Table 9) respectively. Other variables did not show an increase in stress values. The region of point of application of stress and the first thread of the implant showed the maximum stress similar to that of control in the case of cortical bone thickness, pitch and force angulation.

***b) Minimum principal stress:***

A stress value of -41.582 MPa (Figure 7c, lower panel) was obtained for the control simulation. The mesial surface of the implant first thread showed the minimum stress value.

***Effect of variables:***

Force angulation and pitch showed a decrease in stress values to -27.282 MPa (Table8) and -29.215 MPa (Table 7). Diameter and exposure length showed an increase in stress values of -54.731MPa (Table 5) and -140.561MPa (Table 9). The implant first thread on the mesial surface showed the minimum stress value in the case of cortical bone thickness and force angulation. The point of application of force showed the minimum stress in the case of diameter and pitch and in the case of exposure length at the implant fifth thread.

***c) Von mises stress:***

The equivalent stress was found to be 30.356 MPa (Figure 7a, lower panel) and was seen at the point of application of force.

***Effect of variables:***

The magnitude of stress was increased in the case of diameter 51.57 MPa (table 5) and increased exposure length 74.59MPa. All the other variables resulted in similar stress as that of control. The pattern of distribution of stress was similar in all simulations.

***B) Distribution & magnitude of stress in cortical bone:***

***a) Maximum principal stress:***

The maximum principal stress 16.434 MPa (Figure 7b lower panel) was seen distal to the implant for the control simulation. The stress pattern was a seen as a ring of decreasing gradient similar to that seen in group I control simulation.

***Effect of variables:***

Increasing the diameter of the implant to 1.5mm decreased the stress values to 12.381 MPa (Table 5). Decreasing the cortical bone thickness to 1mm increased the stress values to 20.219 MPa (Table 7). Force angulation and pitch did not result in a significant change in the values of stress. Increase in exposure length to 5mm increased the stress value to 46.752 MPa.

***b) Minimum principal stress:***

The stress value of -17.674 MPa (Figure 8c, lower panel) was seen in the mesial surface of the bone for the control. The stress pattern was similar to that of the maximum principal stress except that it was seen in the mesial surface of the bone.

***Effect of variables:***

Increase in diameter of the implant reduced the stress values -9.7MPa (Table 5). Force angulation did not result in an increase of stress values -16.491MPa (Table 8). Decreasing cortical bone thickness increased the stress values -21.862 MPa (Table 6) so as increased exposure length -19.268 MPa (Table 9). Increasing pitch of the implant decreased the stress values to -13.298 MPa.

***c) Von Mises stress:***

The equivalent stress was found to be 32.213 MPa (Figure 8a, lower panel) and was seen at the point of application of force.

***Effect of variables:***

The magnitude of stress was increased in the case of diameter 51.105 MPa (Table 5) and increased exposure length 126.045Mpa (Table 9). All the other variables resulted in similar stress as that of control. The pattern of distribution of stress was similar in all simulations.

***C) Distribution & magnitude of stress in cancellous bone:***

***a) Maximum principal stress:***

Stress value of 0.055537 MPa (Figure 9c, lower panel) was obtained for the control simulation. The pattern of stress was more localized than that seen in cortical bone.

***Effect of variables:***

An increase in diameter to 1.5mm resulted in a slight increase in stress to 0.071025 MPa (Table 5). Reducing the cortical bone thickness resulted in a very significant increase in stress value to 0.135663 MPa ( Table6). Angulation of force, pitch and exposure length did not result in a significant increase of stress values compared to control.

***b) Minimum principal stress:***

The minimum principal stress was -0.035422 MPa (Figure 9c, lower panel) in the mesial surface of the bone for the control simulation.

***Effect of variables:***

Increasing the diameter resulted in a decrease in the stress level to -0.02634Mpa (Table 5). Cortical bone thickness increased the stress levels to -0.131533 MPa. Pitch and angulation of force showed a similar stress level to that of control. Exposure length caused an increase of stress level to -0.114158 MPa (Table 9).

***c) Von mises stress:***

The equivalent stress was found to be 0.047493 MPa (Figure9a, lower panel) and was seen at the point of application of force.

***Effect of variables:***

The magnitude of stress was increased in the case of diameter 0.052328 MPa (Table 5) and increased exposure length 0.047493 MPa (Table 9). All the other variables resulted in similar stress as that of control. The pattern of distribution of stress was similar in all simulations.

***III) Comparison of stress in group I & group II:***

***a) In Implants:***

The stress patterns and magnitude was similar in both groups for all the three types of stress for the control simulation. Increasing the diameter to 1.5mm from 1.3mm resulted in a two fold increase in all the three types of stress for both groups more so for group I than group II. Decreasing the cortical bone thickness from 2mm to 1mm and changing the pitch to 1mm showed similar magnitude of stress as that of the control simulation for both groups. Changing the force angulation to 60 degrees resulted in similar magnitudes for Maximum principal stress and Von mises stress for both groups but Minimum principal stress in group II was half of that compared to that of group I. Increasing the exposure length to 5mm resulted in a six fold increase in all the stresses for group I and a fourfold increase in stresses for group II.

***b) In Cortical bone:***

For the control simulation the magnitude and distribution of stress was similar to both groups. Increasing the diameter caused an increase in Maximum principal stress for group I but Minimum principal stress and Von misses stress in group I resulted in a similar magnitude as that of control. However in group II Minimum principal stress showed a decrease in magnitude. Decreasing cortical bone thickness produced a consistent rise of stresses with Von misses stress increasing two fold in both groups when compared to that of control. Pitch and force angulation did not result in a change in stress values for both groups as that of control. Exposure length caused an increased stress values for both groups more for group I when compared to that of group II.

***c) In Cancellous bone:***

Group I showed an increased stress values of Maximum principal stress and Von Misses stress than group II for control simulation. Diameter caused an increase in Maximum principal stress in group II but a decrease in group I. Cortical bone thickness resulted in an increase in all the stresses for both groups more for group I than group II. Force angulation and pitch did not result in a significant increase of stress in both groups. Exposure length produced a similar increase in stress in both groups when compared to the control.

***Ting-Sheng Lin et al (2013)***[55] compared exposure length, insertion angle and direction of force on cortical bone stresses. The authors used 27 models to simulate the effects of these variables on cortical bone stress. The results showed that exposure length has the highest influence (82.38%) followed by insertion angle (6.03%).Direction of forces had no influence on stress values. All the three parameters had an influence on cancellous bone values. The present study also confirmed the findings of the above mentioned study with exposure length causing an increase in all the three types of stress in implants, cortical and cancellous bone. The present study also concurred that stress in cortical bone is greater than the cancellous bone at all simulations.

***Abhishek H Meher et al (2012)***[51] compared varying cortical bone thickness, applied forces and angulations. They compared cortical bone thickness of 1.5mm and 2mm, force or two different magnitudes

200 and 300 g respectively. The force angulation was maintained at 70, 90,110 and 130 degrees. They concluded that the maximum equivalent stress (MES) and maximum deformation (MD) had an inverse relation with cortical bone thickness. The MES decreased with increased force angulations whereas the MD increased with increasing force angulation. On increase in load the MES showed no change but the maximum deformation MD increased. This study compared the effect of cortical bone thickness at 1mm and 2mm respectively and the force angulation at 45 and 60 degrees. The present study agreed with the fact that decreasing the cortical bone thickness increased the maximum equivalent stress MES in cortical and cancellous bone. Abhishek Meyer et al also concluded that increasing the angulation of force from 70 to 130 degrees decreased the maximum equivalent stress MES but this study did not show any decrease in MES when increasing the force angulation from 45 to 60 degrees. A possible explanation may be that, the effect of force angulation is seen only at higher angulations of force and also the range of force angulation studied in the present study is limited.

***Te-chun Liu (2012)***[50] et al compared the bone quality, loading conditions, screw effects and implanted depth. They utilized a 3 D bone block and used a FEM to analyze the stress patterns produced in bone. The bone blocks were of varying cortex thickness; cancellous bone density and the implants simulated were of varying diameter 1.2mm, 1.5mm, 2mm. The results were as follows, the stress/displacement decreased with increasing cortical bone thickness, but cancellous bone played a minor role in influencing the stress patterns. The two indices were linearly proportional to the force magnitude and produced the highest values perpendicular to the long axis of the screw. They also concluded that a wider screw produced a decreased force and exposure length has the most influence on stress patterns in cortical bone. The present study contradicted the above study that decreasing diameter did not result in a decrease in the stress value. However the present study agrees with the above study that increasing cortical bone thickness resulted in decreased stress.

***Akihiro et al (2011)***[45] measured the stress distribution in cortical as well as cancellous bone using 3 dimensional FEM. The authors used 3 types of implant design cylindrical pin, helical thread, and non-helical thread designs. A force of 2N was applied in all 3 directions and force angulations of 30, 40,45,50,60,70,80,90 were used. They concluded that placement angle; type of miniscrew and direction of the force played a major role in determining the magnitude of stresses in bone. The present study contradicted in the fact direction of force did not result in an appreciable variation of force levels.

***Motoyoshi et al (2005)***[68] compared the effect of pitch and abutment on stress values in miniscrews. The authors modeled six different models with varying pitch 0.5mm, 1mm, 1.5mm and with/without abutments. The authors concluded that the effect of pitch was not significant on stress values. The present study agrees with the fact that pitch of the miniscrew had a negligible effect on the stress values.

***Jordi Marce-Norgue et al (2013)***[56] evaluated miniscrew design parameters such as screw length, head diameter, shank diameter, shank length, shank shape, thread shape and pitch of ten different commercially available miniscrews. Bio mechanical parameters such as cortical bone thickness, insertion angle, drilling depth was also considered in evaluating the Von-misses stresses and the deformation of the screws. A force value of 1 N was applied perpendicular to the implant was applied to the miniscrews. They concluded that the direction of the applied forces had a limited influence on the stresses produced and geometric parameters such as head diameter, diameter of the shaft, thread pitch had an important relation to the Von-misses stresses produced. According to the authors an ideal implant should have a diameter of 2mm, cylindrical shape, short and wide head, short and wide shank and appropriately sized threads. The present study agrees with two of its conclusions, the diameter affecting the stress values and the direction of applied forces had no effect.

***Ashish Handa et al (2011)***[44] evaluated the influence of thread pitch on maximum effective stress in cortical bone. The authors selected implants of thread pitch of 0.5, 1.0, and 1.5mm placed 1mm mesial to lower mandibular molar with a force magnitude of 2 N at a 45 degree angulation. The Von-mises stresses are maximum at the neck of the implant, bone interface and increased with increasing thread pitch. However the pattern of the stress was not influenced by the thread pitch. The present study resulted that neither the pattern of stress nor stress values was not influenced by the pitch of the miniscrew.

***Ramzi et al (2012)*** [54] investigated the Von mises , principal stresses in cortical bone influenced by design factors of the implant as well as bone factors such as Young’s modulus of the cancellous bone. They used 26 bone blocks modeled 3-dimensionally and a force of 2 N in a linear direction was used. They concluded that certain implant design factors such as implant diameter, head length, thread size and bone factors such as Young’s modulus of elasticity influenced the stresses produced in cortical bone. Other factors such as thread shape, thread size, cortical bone thickness did not have an impact. This study provided similar results in that the design factors such as diameter had an influence on stress and other factors such as pitch did not have an impact. However the present study contradicts that cortical bone thickness did have an influence on stress values.

## SHORTCOMINGS OF THE STUDY

1. Any study using Finite Element Analysis obtains only the initial stress and strain distribution information. At the present state of our knowledge, it is impossible to derive what precisely happens over certain length of time, when the same loading conditions continue. This drawback applies to the present investigation also.

2. Analogous to the previous FE studies, bone was modeled assuming that the cortical and trabecular bone were isotropic, homogenous and linearly elastic. The models did not include the heterogeneous aspects of the surrounding bone, such as osteons, harvesian canals, interstitial lamellae, porosity, etc, because their structures cannot be modeled. Also, the information concerning the precise material properties of these bone microstructures is lacking.

3. Another assumption that had to be made was a hundred percent fixation (total Osseo-integration) of the implant to the bone. This became necessary considering the fact that these implants are mechanically retained in the bone, and that there would not be any interface between the two. This may not be possible to achieve in clinical practice (since it is an ideal and unrealistic assumption). Also, hundred percent bone apposition is not always obtained at the surface of the endosseous implant. This assumption was made because sufficient data concerning osseous healing and the interface between dental implants are still unavailable.

# SUMMARY & CONCLUSION

Anchorage control is one of the challenges for an orthodontist. Efficient attainment and control of anchorage is fundamental to successful orthodontic and dentofacial orthopedic treatment. The clinical success of an implant is largely determined by the manner in which the mechanical stresses are transferred from the implant to the surrounding bone, without generating forces of a magnitude that would damage the integrity of the supporting bone and consequently the longevity of the implant. Hence, this study was carried out to *evaluate the distribution of stresses in implants and in the bone surrounding the orthodontic implant, by testing various parameters* that may affect the stability of the miniscrews in bone

The finite element method was chosen for this study because it has proven to be an useful non-invasive tool in examining the mechanical behaviour of bone, wherein complex structures are formulated as mathematical models in ***Ansys*** and ***Hypermesh*** software on a ***Windows*** platform were used to perform the study.

The study was done by measuring the miniscrews and uploading the measurements to the *solid works modeling software*. The geometric models of the implant were thus obtained using the reverse engineering technique creating a 3D CAD mode. The 3D CAD models in .IGES format were subsequently uploaded to the meshing software *Hypermesh CAE.* A solid bone block of 20mm×20mm×10mm was constructed. The whole assembly of miniscrews and the bone block was imported into the finite element workbench *Ansys*.

Various parameters that may affect the stability of miniscrews such as cortical bone thickness (anatomic variable), pitch, diameter, taper (design variables), insertion depth, angulation of force (clinical variables) were tested. A 2N force which is clinically desirable was applied to the miniscrews. All the three types of stress Maximum, Minimum, Von-mises stress were studied in all the simulations. Additionally the pattern of stress distribution was also studied. All the results were compared to the yield strength of the materials in the model (bone and titanium) was equated with those obtained from existing studies[45]. 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.

The following conclusions were obtained in the study:

a) Between tapered and cylindrical implants no clinically relevant differences were seen in the stress levels obtained.

b) The magnitude of stress generated did not exceed the yield strength of the material (bone, titanium) at all simulations except in the case of exposure length.

c) The magnitude of stress was significantly higher in cortical bone than in cancellous bone for all the simulations.

d) Of all the variables tested cortical bone thickness (anatomic variable) and exposure length (clinical variable) and diameter (design variable) played a very significant role in determining the magnitude of stress in bone.

e) The neck of the implant showed the maximum stress in all the simulations.

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