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UNIVERSITY OF GAZIANTEP
GRADUATE SCHOOL OF HEALTH SCIENCES

COMPARISON OF STAINLESS STEEL AND TITANIUM ALLOY INFRAZYGOMATIC CREST MINI IMPLANTS BY USING FINITE ELEMENT ANALYSIS

Sarah Faris ALSABUNCHI
MASTER OF SCIENCE THESIS

DEPARTMENT OF ORTHODONTICS

SUPERVISOR
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DECLARATION

I hereby declare that this thesis is my own work and effort, has been generated by me as the result of my own original research.

I confirm that:

1. Where I have consulted the published work of others, this is always clearly attributed;
2. Where I have quoted from the work of others, the source is always given. With the exception of such quotations, this thesis is entirely my own work;
3. I have acknowledged all main sources of help.

28.05.2018

Sarah Faris ALSABUNCHI

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ABBREVIATIONS

Ti	Titanium
SS	Stainless Steel
IZC	Infrazygomatic Crest
MIs	Mini implants
FEA	Finite Element Analysis
TADs	Temporary Anchorage Devices
CBCT	Cone Beam Computed Tomography
ISQ	Implant Stability Quotient
g	gram
N	Newton
Mpa	Megapascal
mm	millimeter
NiTi	Nickel titanium
3D	Three dimensional

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ÖZET

PASLANMAZ ÇELİK VE TİTANYUM ALAŞIMLI İNFRAZİGOMATİK MİNİ İMPLANTLARIN SONLU ELEMANLAR ANALİZİ İLE KARŞILAŞTIRILMASI

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İnfrazigomatik krest mini implant, maksimum ankraj kullanılarak spesifik ortodontik hareketlerin elde edilmesi gereken durumlarda, iskeletsel ankraj amaçlı yaygın olarak kullanılan geçici ankraj apareylerinden biridir. Titanyum ve paslanmaz çelik, kendine has materyal özellikleri nedeniyle ortodontik mini implantlar için tercih edilen malzemelerdir. Bu çalışmanın amacı, titanyum ve paslanmaz çelik infrazygomatic krest mini implantlarının oluşturdukları strese göre, sonlu elemanlar analizi kullanılarak karşılaştırılmasıdır. Maksilla ve zigomatik kemiği içeren bir kafatası modeli ile iki mini implanta ait (Ortho Bone screw, mushroom 2.0 ve Dentos Abso anchor, small head type SH 2018-12) 3 boyutlu modeller oluşturuldu. Her iki mini implant da infrazygomatic krest alanına, oklüzal düzlem ile 70 derecelik açı ile ve oklüzal düzlemin 14 mm yukarısında olacak şekilde yerleştirilmiştir. Daha sonra iki grupta da, close coil spring aracılığıyla, maksiller molar distalizasyon için 300 gr ve maksiller molar intrüzyon için 150 gr kuvvet uygulanarak, kortikal kemikte, kansellöz kemikte, mini implant yüzeyinde ve maksiller birinci molar dişin köklerinde oluşturulan basma ve gerinim dağılımları değerlendirildi. Titanyum ve paslanmaz çelik mini implantların arasındaki PMax, PMin ve von Mises streslerinin pik değerleri karşılaştırıldı. Her iki grupta da, kortikal kemik, kansellöz kemik ve mini implant yüzeyinde ortaya çıkan stresin tüm tepe değerlerinin, titanyum modellerinde daha yüksek olduğu görülmüştür. Titanyum mini implantlar, infrazigomatik krest bölgesinde ankraj amaçlı kullanıldığında paslanmaz çelik mini implantlardan daha fazla stres oluşumuna neden olmaktadır.

Anahtar Sözcükler: İnfrazigomatik krest, Titanyum, Paslanmaz çelik, Mini implant, Ankraj.

ABSTRACT

COMPARISON OF STAINLESS STEEL AND TITANIUM ALLOY INFRAZYGOMATIC CREST MINI IMPLANTS BY USING FINITE ELEMENT ANALYSIS

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The infrazygomatic crest mini implant is one of the temporary anchorage devices that have been widely used as skeletal anchorage in cases where specific orthodontic movements have to be obtained with absolute anchorage. Titanium and stainless steel are the materials of choice for orthodontic mini implants due to the specific properties of each of them. The purpose of this study was to compare between titanium and stainless steel infrazygomatic crest mini implant regarding the stress generated by each of them using finite element analysis. Three dimensional models of maxillary and zygomatic bone with mini implants (Ortho Bone screw, mushroom 2.0; Dentos Abso anchor, small head type SH 2018-12) were constructed, both mini implants have been inserted in the infrazygomatic crest area at angle of 70 degree and 14 mm above the occlusal plane. Two different scenarios then performed on each mini implant with forces of 300 g for maxillary molar distalization and 150 g for maxillary molar intrusion using close coil spring. Stress/strain patterns generated on cortical bone, cancelous bone, mini implants and roots of the maxillary first molar were evaluated, then the peak values of PMax, PMin and von Mises stresses were compared between titanium and stainless steel mini implants. In both scenarios peak values of stress were greater in titanium than in stainless steel for all study models. Titanium mini implants create greater stress than stainless steel mini implants when used as anchorage device in the infrazygomatic crest region.

Key Words: Infrazygomatic crest, Titanium, Stainless steel, Mini implant, Anchorage.

1. INTRODUCTION

The aim of orthodontic treatment is to make patient's life better through improving their dentofacial functions and esthetics. Orthodontic anchorage, which is defined as a endurance to unfavorable tooth movement (1), is considered as an important factor for prosperous treatment of dental and skeletal malocclusions (2, 3). Anchorage has been also defined by Freudenthaler et al. as the amount of allowed movement of the reactive unit in a force system (4).

Keeping anchorage under control helps to evade unfavorable tooth movements. In some cases while, even a small amount of force can cause unfavorable movements; it is very consequential to have an absolute anchorage to prevent them (5). Absolute anchorage has been defined as no movement of the anchorage unit which means zero anchorage loss when forces applied for teeth movement (1). This type of anchorage can only be achieved by using ankylosed teeth, dental implant or mini plate as anchors, both depending on bone to avoid movement (6). This type of anchorage which depended on devices such as implants or mini implants placed in the bone, can be acquired by increasing the support to the reactive unit as indirect anchorage or by immobilizing the anchor units themselves which recognized as direct anchorage.

Traditional means of anchorage for many years were either extraoral which included: high pull, cervical and J-hook headgear; or intraoral like: elastics, undertying, distal-jet, bonding 7s, TPA, lingual arch and Nance appliances. Marcotte has defined anchorage requirements in three categories: type A, the posterior teeth immobile during anterior teeth retraction; type B, reciprocal closure of space where equivalent amount of force from both sides have applied; type C, anterior teeth not moving during protraction of the posterior dentition (7).

With increasing the need for more stable and stronger anchorage unit, skeletal anchorage devices have been widely used. Generally, the skeletal anchorage devices include bone plates and mini implants. These implantable temporary skeletal anchorage devices (TADs) become very popular and favorable for orthodontists because of the greater effect in controlling the anchorage, they are more acceptable by patients as they require less cooperation regarding wearing the appliances, they can decrease the overall treatment time sometimes and they permit orthodontic treatment for some cases which thought to be impossible without surgery. TAD is successfully useful for solving vertical type malocclusion in cases of open bite and overeruption of teeth due to loss of antagonists (8). Traditionally, skeletal open bite known to be treated with surgical techniques like maxillary impaction to decrease the dentoalveolar height. Introducing TADs to orthodontics has eleminate the need for surgery and became an alternative option when used for molar intrusion which provides a conservative treatment manner with an irreversible damage (9, 10).

Temporary anchorage devices can also eliminate the need of patient compliance in cases of sagittal dental movements such as mesializing or distalizing the whole dentition both with and without tooth extraction (11), TADs considered to be the best and most proper mean to correct all types of malocclusions if only the correct diagnosis and mechanical design have been made, except for those cases with facial deformities which need invasive surgeries to obtain a compatible skeletal relationship (12). Several types of TAD have been used in orthodontics (13). Roberts et al. reported conventional osseointegrated implants (14), while Costa et al. (15) and Freudenthaler et al. (4) reported mini and microimplants as orthodontic anchorage devices.

Mini implants (MIs) have been utilized as orthodontic anchorage device for over years and rapidly spread (16, 17), as they successfully used for intrusion, extrusion, space closure, molar uprighting, molar distalization, open-bite closure, and a variety of other anchorage purposes (18, 19). Material of mini implant consider as one of the crucial factors that affect their success rates and tip breakage during insertion. Elastic modulus, resistance to corrosion, strength and biocompatibility are important consideration in selecting an orthodontic mini implants material, also the material chosen should have sufficient mechanical strength to provide resistance to stress and strain during clinical placement and removal of mini implant without persistent deformation (20). As titanium (Ti) is a highly biocompatible material, most mini implants are made of titanium or titanium alloys (21), but some manufacturers have recommended stainless steel (SS) mini implants which offer better mechanical properties when used as anchorage in en mass retraction (22). While some comparative studies focused on the mechanical and physical properties between titanium and stainless steel mini implants and effects on primer stability (20, 23, 24), there is obvious lack in studies that compare Ti and SS stresses or stress pattern generation.

Mini implants have superior properties as they can be placed in various unusual sites intraorally such as the alveolar bone of adjacent teeth without damage effect on the roots or waiting time for osseointegration, they can also be placed in the palatal area, maxillary tuberosity and recently the infrazygomatic crest region (25-27). The infrazygomatic crest mini implant has been successfully used to provide a perfect skeletal anchorage for the purpose of en-masse anterior retraction, intrusion of the maxillary posterior teeth and maxillary dentition distalization (28, 29).

One of the interesting challenges in orthodontics was evaluating the mechanical effects of mini implant on bone. These include the inability to measure stress levels in patient’s bone and how to visualize the stress patterns that are generated clinically. Finite element analysis (FEA) has utilized to overcome these challenges. FEA which is recently used in purpose of determining stress and strain degree in the dental structures, their boundaries areas and determining stress creating on/by the mini implants to provide their mechanical properties. This method includes a series of computational procedures for calculating the stress and strain in every element and performs a model solution. As FEA can solve many problems by calculating physical measurements of stress within the structures, it considered to be a useful application in evaluating aspects of mini implant use in orthodontics (30).

Several studies regarding the infrazygomatic crest mini implants have been accomplished using cone beam computed tomography (CBCT) (31), in vivo (28) and in vitro (32). artificial bone, these studies have been conducted for determining thickness of infrazygomatic crest, biting depth of the mini implant, appropriate insertion site and angulation of the implant in this area (33). While there are no studies conducted in FEA to determine and evaluate the stress created by mini implants inserted in the infrazygomatic crest and despite a lack of evidence from FEA studies to support preferred mini implant material type, the aim of our study is to evaluate and compare stresses generated in mini implant and bone using FEA when two mini implants with varied materials (titanium and stainless steel) placed in the infrazygomatic crest region.

2. GENERAL INFORMATION

2.1 Anchorage

The secret behind any succseeful orthodontic treatment is “anchorage control” which has usually obtained by using teeth, intraoral appliances, and extraoral appliances, for minimizing the movement of specific teeth, while completing the desired movement of other teeth. However, according to Newton’s third law, every action there is an equal and opposite reaction, there are limitations in our ability to absolutely control all aspects of tooth movement. Many case reports and orthodontic articles has been published to enhance or overcome the traditional anchorage’s limitations, through explaining the possibility of useing various types of anchorage devices which are placed temporarily in ultimate contact with bone. Although the standards of orthodontic anchorage have been perfectly understood since the 17th century, it has not been clearly broght to light until 1923 when Louis Ottofy described it as “the base against which orthodontic force or reaction of orthodontic force is applied” (34). Lately, in 2009 Daskalogiannakis defined anchorage as “resistance to unwanted tooth movement” (1). Many classifications based on anchorage needs have appeared over the years. Ottoffy summarized the anchorage categories previously mentioned by E.H Angle and others as simple, stationary, reciprocal, inetrmaxillary or extraoral (34). Moyers then updated Ottoffy’s classification and made different subcategories of extraoral anchorage also he subclassified the simple anchorage into single, compound and reinforced (35). Gianelly and Goldman, for the purpose of indicating the extent to which the teeth of the active and reactive units should move when a force is applied, they propose the terms maximum, moderate, and minimum (36). Marcotte and Burstone in their classification of anchorage they depended on the amount of participation of the anchorage unit in closing the space, and they classified the anchorage into three categories A, B, and C (37, 38).

Tweed defined anchorage preparation, or the uprighting, and even the distal tipping of posterior teeth to utilize the mechanical advantage of tent peg before retracting anterior teeth (39). In 1728, Fauchard in his study he found that the use of the expansion arch through ligating the teeth to a properly shaped rigid metal plate, which widened the crowded dentition to a normal form (40). About 100 years later, Gunnell suggested an occipital region as anchorage in 1822, but did not describe its use until 1841. In 1841, J.M.A. Schange perfected the "crib" of Delabarre and use it in the palatal plate as anchorage, which enable the use of a labial arch and ligatures of silk or gold wires to obtain several tooth movements (40). Occipital anchorage was later described by Angle in 1891. Desirabode, in 1843 suggested the use of teeth teeth which have and stronger roots as anchorage. Orthodontic anchorage's discussion would not be completed without the participate of E.H. Angle, who also introduced the concept of occlusal anchorage in 1891 and stationary anchorage in 1887 (40). Traditional means of anchorage may be extraoral (high pull, cervical, and J-hook headgear) or intraoral (elastics, undertying, distal-jet, bonding 7s, TPA, lingual arch, Nance appliances). The need for skeletal anchorage in orthodontics has been significantly increased with the growing numbers of adult patients looking for orthodontic treatment. In addition, complex treatment aims, patients with missing teeth, not appropriate or uncooperative to use extra-oral anchorage all added to increase the requirement for the skeletal anchorage. The recent advent of temporary skeletal anchorage devices in orthodontics provides an additional method for anchorage control to assist in obtaining desired tooth movement in three dimensions.

2.2 Temporary Anchorage Devices

A temporary anchorage devices (TADs) reffers to those devices which are temporarily fixed to bone to enhancing orthodontic anchorage through supporting the teeth of the reactive unit or through obviating the need for the reactive unit completely, and which is removed after use. They can be located in the transosteal, subperiosteal, or endosteal region; and they can be also fixed to bone either mechanically (cortically) or biochemically (osseointegrated).

Basically, the incorporation of TADs into orthodontic treatment facilitate the infinite anchorage and made it possible, in terms of implants, the infinite anchorage is defined as showing no movement (zero anchorage loss) as a consequence of reaction forces (1).

Generally, there are two types of TADs in orthodontics: mini implants and bone plates. The use of bone plates has obviously decreased due to their disadvantages of limited placement location. Alternatively, mini implants can be widely used much more frequently in practice due to their ease of application and removal also their wide range of placement location.

2.3 Mini Implants

The idea of using bone screws as temporary skeletal anchorage dates back to 1945, when Gains forth and Higley placed vitallium screws in the ascending ramus in dogs for canine retraction (41). The first clinically reported use in humans came from Creekmore and Eklund in 1983 for the purpose of treating a patient with deep bite; they inserted vitallium bone screws in the anterior nasal spine. An elastomer has been used from the screw to intrude the incisors just 10 days after insertion of the screw (42). However, the use of mini-implants was not embraced until 1997, when Kanomi described a mini-implant specifically made for orthodontic use (18). Soon after in 1998, Costa et al. presented a screw with a bracket like head (15). These mini-screws which were temporarily fixed to bone for the purpose of enhancing orthodontic anchorage came to be known as TADs and ever since, rapid developments ensued in this area of skeletal anchorage.

Mini implants has been also called screw-type implants, which have also been called: micro-implants, micro screws and miniscrews (43).

An ideal mini implant should be simple to use, has small dimensions, can be loaded immediately, can wear orthodontic forces, immobile, not need compliance, biocompatible, provides clinically similar or superior results when compared with traditional anchorage devices and it should not have high prices. For integrated implants, the maximum load is commensurate with the degree of osseointegration, whereas for nonintegrated implants the maximum load is commensurate with the surface area contact of the cortical bone to the implant.

2.3.1 Classification of mini implants

Mini implants considered as biocompatible TADs which in turn can be classified as biological or biocompatible in nature (39). Both groups can also be subclassified depending on their mode of attachment to the bone, which could be either biochemical (osseointegrated) or mechanical. For example, an ankylosed tooth is considered as a biological TAD which has biochemical contact while fixed to the bone, it is also temporarily used for the aim of orthodontic anchorage and replaced after ending the treatment. On the other hand, a dilacerated tooth which can be used as a biological TAD that is basically fixed to bone mechanically. The biocompatible anchorage devises are either a modification of a dental implant, or a surgical fixation process. For example, a mini implant is a fixation device placed in various regions for anchorage control only by mechanical stability. On the other hand, a palatal implant is a miniaturized dental implant placed in the palate with the intention of osseointegration and use for orthodontic anchorage. Labanauskaite suggested the following classification of the implants uesed for orthodontic anchorage. He excluded dental implants which are placed only for the purpose of suppostıng a prosthesis as they did not considered as TADs because they are not displaced after orthodontic treatment (44).

I. According to the shape and size:
1) Conical (cylindrical)
  a) Mini screw implants
  b) Palatal implants
  c) Prosthodontic implants
2) Mini plate implants
3) Disc implants (onplants)
II. According to the implant bone contact:
  1) Osteointegrated
  2) Non-osteointegrated
III. According to the application:
  1) Used only for orthodontic purposes (orthodontic implants)
  2) Used for prosthodontic and orthodontic purposes (prosthodontic implants)
TADs has also been classified according to the pattern of their insertion
  I- Self-drilling (self cutting)
  II- Self-tapping: require pre-drilling at full length

Orthodontic anchorage may be used either directly or indirectly. When using as direct anchorage, a force is applied directly from the mini implant to gain the desired biomechanical result. While in indirect anchorage, a mini implant is connected to a tooth or segment of teeth in order to fixed them and prevent their moving by using an archwire or ligature wire, thereby incorporating them into the anchorage unit (45).

2.3.2 Indications and contraindications of mini implants

Using mini implants has significantly increased in order to obtain orthodontic tooth movements in cases including insufficient number of teeth and/or lack of occlusion in the anchorage unit, also for extrusion or intrusion of single tooth or units of teeth; asymmetric tooth movements, unilateral expansion, asymmetric cant correction; retraction and/or intrusion of anterior teeth, mesial or distal movement of molars, proclination of anterior teeth in cases where no posterior anchoring element is available, closing space with maximum anchorage, non-compliant patients. In order to their smaller sizes, wider applications in tooth movements, simpler placement and immediate loading, mini implants have become very important in contemporary orthodontics (15, 43, 46-48).

Contraindications of mini implants include, but not restricted to:

- Patients with systemic conditions such as metabolic bone diseases
- Immune suppressive therapy
- Bisphosphonate medication,
- Poor quality bone tissue
- Hypersensitivity
- Radiotherapy in the head and neck damage (49).

2.3.3 Risk of failure

Failure cases has been extensively reported as when correlated with: increased peri-implant inflammation (50, 51), placement in the mandible (52), small diameter (53), root proximity (54), insufficient primary stability (55), placement in the right side of the mouth (50), placement in nonkeratinized tissues (56), tightness of implant insertion (57), self-drilling mini implant design (51), mandibular retrusion (51), uprighting movement (51), when cortical bone is thinner than 0.5 mm and decreased bone density (49), over insertion (58) and loading within two or three weeks (59).

2.3.4 Material used in mini implants

Pure titanium (Ti) and stainless steel (SS) mini implants have excessively been used for oral surgical and orthopedic internal fixation. As Ti considered as a highly biocompatible material, it allows direct bony contact (osseointegration) between dental implants and the adjusent bone. Also, Ti has the characteristic of oxidizing in the presence of air and aqueous electrolytes to form a passive titanium dioxide film which participate in its biocompatibility and corrosion resistance but needs to be alloyed to improve its strength and fatigue resistance.

The best and most successful material of choise for mini implants should possess sufficient mechanical properties, corrosion resistance and should be biocompatible, thus generally most of the orthodontic mini implants composed of titanium- aluminum-vanadium (Ti6-Al4-V) alloys, primarily Ti6-Al4-V (titanium grade IV- V) (21). Pure Tİ is not favorable for mini implant because of alloy’s superior characteristics regarding strength, which make it to overcome prevalent problems with pure Ti mini implants such as fractures or distortions. As increasing the demand for Ti which has widely been used in various industial fields, so it becomes more expensive that’s why in some countries SS preferred over Ti alloy mini implants.

Stainless steel is one of the most frequently used implant materials due to several favourable properties which include mechanical strenght, biocompatibility, manufacturing ease and cost-effectiveness (60). Unlike titanium alloy, SS bone screws tend to make a fibrous integration with the bone (61), which reduces removal torque and leads to easier withdrawal of mini implant (62). Many literatures focused on Ti alloy mini implants explaining their types, properties, and loading behavior (63). In contrast, reports regarding the use of SS mini implants are usually rare.

Both materials ensure a relatively good and successful clinical outcomes regarding the biomechanics and stability of the mini implants. Although SS screws were marketed initially, the greater biocompatibility of Ti made it a more attractive option (17, 64, 65). Also, Ti is lighter than SS, and so it has a yield strength which is twice that of SS. This makes Ti having the highest strength to weight ratio of any metal suited for dental use. We can also mention that modulus of elasticity and thermal expansion factors of Ti are similar to bone, which reduce the potential of implant failure.

2.3.5 Components of mini implant

Basically a typical mini implant has three main components: head of the mini implant, neck and body or core of the mini impalnt, which consists of a helix (the thread) (66). Each component has its importance in the function of the mini implant.

The head: is the extraosseous part of the screw which provides a mean for applying torsion to the core and thread, and it is also act as a point of orthodontic force application. Most mini screw implants have a male type head design which provides an articulation point to the screwdriver and gives control during implantation.

The core or the body: is the intraosseous part which provides the support of the screw and it is wrapped in a helical thread. As the core diameter ultimately determines the torsional strength so the cross-sectional area of the core is very important to the strength of the mini screw implant, which means the incidence of screw fracture during the implantation procedure decrease by increasing core diameter.

The shank: is that part of screw which extends from head to the beginning of the threads.

The pitch: represents the distance between the threads on the screw.

Most common used mini implants have diameters ranging from 1.2 to 2 mm. A tapered mini implant which has an initial diameter of 1.5 mm usually has a decreased diameter at the tip to 1.2–1.1 mm. The difference between the initial diameter and the tip is commonly 0.3–0.4 mm (23). The length of the mini screw implants could be varied from 4 mm to 12 mm.

2.3.6 Stability of mini implants

Insertion of mini implant usually leads to trauma that include both soft and hard tissue, which means damage to the cells and matrix will occur, so it initiates the healing process of bone and soft tissue.

The healing process passes through three distinct phases: inflammation phase, reparative phase and remodeling phase (67). The stability of the mini implant comes from the characteristics of the interface between the implant and the tissue that resulted from the healing process. Usually an osseointegration is highly more desirable than a fibrous integration (68). Producing fibrous tissue at the bone-implant interface could be one of the most serious risk factors of mini implant failure. In order to get more stable and successful mini implant it is important to have sufficient support from bone rather than tissue. For the best bone repair a sufficient amount of reparative cells and nutrients are required, that's why local circulation is extremely important, and bone repair will not be initiated unless local circulation has been created (69). Many factors could affect the healing process of bone, like; osteoporosis, osteopenia, hyperthyroidism and radiation therapy.

2.3.6.1 Primary stability:

It is usually created by mechanical retention of the mini implant in the bone. The peak of it will be immediately after insertion but it decreases concomitantly with bone remodeling that occurs around the mini implant (65). This is related to the tension-compression state that forms at the bone-TADs interface (70).

2.3.6.2 Secondary stability:

It is determined by the biologic bind of mini implant to surrounding bone and it increases over time with the following remodeling that occurs. The overall stability that resulted from both primary and secondary stability is called clinical stability. This is usually decline to minimum at 2 weeks post insertion and finally stabilizes greater than initial primary stability at about 6 weeks (65). Site characteristics is one of the factors which are related to primary stability, also the geometric design of the screw which includes the pitch of screw threads, length, diameter, shape and tip form of the mini implant are affect the primary stability (71-73).

2.3.6.3 Evaluation stability of mini implant

The amount and patterns of stress conducted during placement of mini implants and after application of orthodontic force are crucial factors affecting the stability of mini implants. There are several methods available for studying the stress and stability of implants which can be classified to invasive and non-invasive techniques (74). Invasive methods which intervene with the osseointegration process of the implant such as histologic and histomorphometric technique (75), cutting torque resistance analysis (76) and pullout test (70, 77).

Radiographical examination was one of the first methods used to evaluate the dental implants and the physical properties of surrounding bone immediately after implant placement. Hermann et al. described the bitewing radiographs as a successive technique for evaluation the height of crestal bone around dental implants (78). As mini implants are differently oriented when compared with dental implants, so bitewing technique is not useful to evaluate the bone level in mini implant.

A percussion test considered as simple method to be used in order to estimate osseointegration. It is based on vibrational acoustic science and impact response theory. A clinical judgment about osseointegration is made according to the sound heard upon percussion with a metallic instrument. However, this method mostly depend on the experience level of the clinicians and it is not possible to be standardized clinically.

Pulsed oscillation waveform performs by recording the frequency of implant vibration induced by a small pulsed force (79).

The impact hammer method is a modefied version of the percussion test. The method wich also named periotest has been reported for evaluating implant stability (80). Resonance frequency analysis (RFA) is a method used to evaluate the stability of implant based on vibrations conducted within the bone. The current device which uses resonance frequency for determining the stability of dental implants is the Osstell Mentor device (81). Results of this method are presented as the implant stability quotient (ISQ). ISQ is based on the resonance frequency which ranges from 1 (lowest stability) to 100 (highest stability). A horizontal force which is perpendicular to the long axis of the miniscrew implant is used for the measurement of mini implant stability in bone with RFA. Some clinical studies represented the superiority of the Osstell device over the periotest device for measuring dental implant stability (82). Also Veltri et al. showed that resonance frequency could be used for different types of miniscrew implants (83).

2.3.7 Loading of mini implant

After insertion of the mini implant, it can be immediately loaded or just loaded after a healing period. Both loading mechanisms have shown similar success rates with an orthodontic force of 250 g. In 2008, Garfinkle (84) had not find statistically significant difference between the immediately loaded mini implants and delayed loaded mini implants in mean of success rates. Therefore; neither the timing of force application, nor the force itself could precipitated failure of the mini implant. Clinical orthodontic forces can be applied immediately to mini implants with high clinical success rates (84).

2.3.8 Site selection

Site of insertion is also a crucial point for installation of a mini implant. There are many factors which should be kept in mind while selecting the location of mini implant. These factors include: the quality and quantity of bone (85) (morphology, density and depth of the bone), cortical bone and thickness of soft tissue, proximity to the surrounding anatomical structures such as roots, ligaments, and nerves and finally the malocclusion site. So, TADs which are inserted in the maxilla have higher success rates than the mandible. In the maxilla, the maximum amount of mesiodistal bone has been detected on the palatal side, between the second premolar and the first molar. The minimum amount of bone has found in the tuberosity. The maximum thickness of bone in the buccolingual dimension detected between the first and second molars, while the minimum was determined in the tuberosity. In the mandible, the maximum amount of mesiodistal dimension was between first and second premolar. The minimum amount of bone was between the first premolar and the canine. In the buccolingual dimension, the maximum thickness was between first and second molars. The minimum amount of bone was between first premolar and the canine (86). Regarding soft tissues, the attached gingiva is usually most favorable site for insertion followed by mucogingival line.

The most common location for mini implant placement is the buccal surface of the posterior maxillary alveolus which can help in facilitating numerous biomechanical movements including; leveling gingival contour, en masse distalization of the maxillary arch, retraction of anterior teeth, mesialization of maxillary molars for class III correction, intrusion, correction of canted occlusal planes, molar uprighting, extraction space closure and midline correction. Extrusion and de-impaction of canines and molars could indicate mini implants use in other areas of the alveolus and jaws (51, 52).

2.3.8.1 Infrazygomatic crest region

The infrazygomatic crest (IZC) is one of the important anatomical sites which used as anchorage region for placement of orthodontic mini implants and miniplates. The IZC region has important specifications such as the thicker cortical bone, which helps in good primary stability. IZC mini implants have been successfully used for providing skeletal anchorage for en-masse anterior retraction, maxillay molar distalization and also intrusion of the maxillary posterior teeth (87-89).

Anatomically, the IZC can be defined as a column of cortical bone at the zygomatic process of the maxilla which are distant from the dentoalveolar region, permit to unobstructed tooth movement, and reducing the chance of root contact. Clinically, it is an obvious prominent bony ridge elongated the curve region between the alveolar and zygomatic processes of the maxilla. In adults, it is located just above the maxillary first molar.

The most common complications during mini implant insertion in the IZC of adults is injury to the mesiobuccal root of the maxillary first molar and maxillary sinus perforation, that’s why it is more favorable to insert mini implants in higher positions rather than lower, such as midroot, at or beyond the root apex where interseptal bone is usually thicker and there is less chance of root injury. Therefore, to avoide injuring the vital tissues , the mini implant should be inserted in the IZC fo about 14 to 16 mm above the maxillary occlusal plane, and the insertion angle should be 55° to 70° to the maxillary occlusal plane (33).

2.4 Finite Element Analysis

Finite element analysis (FEA) is a computer-based numerical simulation technique that is commonly used for predicting the mechanical behavior of engineering structures. It could be used to calculate many phenomena’s such as deflection, stress, vibration and buckling behavior. It has also been used for analyzing either small or large-scale deflection because of loading or applied displacement. At the same time, it could be used to analyse elastic deformation, or “permanently bent out of shape” plastic deformation. Finite element analysi uses a complex system of points (nodes) and elements, which make a grid called as mesh. This mesh is programmed to contain the material and structural properties (elastic modulus, Poisson’s ratio, and yield strength), which define how the structure will react to certain loading conditions. The mesh acts like a spider web, in that, from each node there extends a mesh element to each of the adjacent nodes. The FEA method was first developed in 1943 by R. Courant (90). His main point was to decrease the various calculative processes to have ultimate solution to bio-mechanical system. He used Ritz method to solve such numeric equations. Later in, Turner et al. (91) tried to describe this method by developing wider definition of these numeric analyses. Since the 1980's, FEA has been rapidly performed into computer programs. Such computer based numerical stress analysis methods make the complex distributions of elastic and inelastic stresses in engineering components to be more easily calculated and allow analyses to be accomplished for static and dynamic loads. The process of analysis has been facilitated through the processes of component drawing and reverse engineering through using Computer Aided Design (CAD) systems.

FEA has been then first used in dentistry when Thresher's showed that it can be used for stress analysis of human teeth in 1972 (92), then Yettram's evaluation about the restored tooth crowns in 1976 (93). Many studies have been reported regarding the stress analysis in dentistry including stress analysis of inlays, crowns, bases supporting restorations, fixed bridges, complete dentures, partial dentures, periodontics and endodontic, also studies of teeth, bone, and oral tissues (94-97).

FEA has also been used to incorporate and predict the biomechanical performance of many different implant designs which are used in dentistry and medicine (98). These predictions may be used to find out the clinical factors effect on the success of implants. For example, many researchers have used the FEA for evaluating the interfacial stress distribution which formed in area where the endosseous implants contact cortical and trabecular bone (99).

2.4.1 Finite element analysis in orthodontics

The importance of FEA in orthodontics is that evaluating the loading and deformation properties of complex structures after the orthodontic forces application by analytically apply different force systems at any point and in any direction. Orthodontic tooth movement occurs through remodeling processes of the alveolar bone, which are stimulated by changes in the stress/strain distribution in the periodontium. The FEA method usually used to describe the complex stressed situation within the periodontal ligament (PDL) and implant-bone interaction, also in determining the patterns of orthodontic tooth movement (100, 101). FEA has been generally used in orthodontics applications specifically for mini implants (102). The results of FEA are represented and evaluated both visually and numerically, so the stress / strain of the biomedical system can be visually evaluated. By using illustration, FEA describe the concentration and distribution of stress within boundaries by utilizing a continuum of color to reflect varying stress levels, then makes the modeled region to be solved as numerically described. FEA allows inferences and impressions to be made from the single mathematical solution for each variable or scenario. Likewise, effect of 'clinical significance' may be made from these results when combined with other existing data. Although there are some limitations, FEA is considered to be a useful tool in many applications. Since there is only one unique solution which is possible for any specific set of conditions, statistical analysis is not necessary to be used in FEA.

The time and effort which is necessary to create a realistic model may be one the most important disadvantages of FEA. Generating a completed maxillary or mandibular arch, complete with enamel, dentin, pulp and PDL for each tooth, as well as lamina dura, cortical and trabecular bone could take hundreds of hours. One other obvious shortcoming in utilizing FEA simulation of the mechanical behavior of mini implants is the representing of real anatomy of human hard and soft tissues, and also their response to mechanical force over time including healing and remodeling.

2.5 Maxillary Molar Distalization

Most of the cases which have Class II molar relationships tend to have anteroposterior skeletal discrepancy, Class II canine relationship and excessive overjet. When the extraction is contra indicated, molar distalization or in other word whole dentition distalization will be the best way for correcting the Class II malocclusion and somehow obtaining an ideal profile. Different mechanisms and appliances can been used for the purpose of molar distalization, the extra oral appliances such as headgear has the disadvantage of patient's compliance, to minimize the need for patient's compliance many intraoral appliances have been introduced such as repelling magnet (103, 104), coil springs on a continuous arch wire (105, 106), superelastic nickel-titanium arch wires (107), coil springs on a sectional arch wire (108, 109), and springs in beta titanium alloy (pendulum, K-loop, intraoral bodily molar distalizer) (110-112).

After introducing the TADs many studies had been reported regarding molar distalization using TADs. Byloff et al. used a Graz implant–supported pendulum appliance (113). Keles et al. designed a slide appliance with osseointegrated palatal implant instead of a Nance button (114). Carano et al. invent a distal-jet combined with a mini implant anchorage system (25). Gelgor et al. combined distal force mechanics using a compressed coil spring with an intraosseous screw (115). Inter radicular mini impalnts are also used for molar distalization, Lin proposed a new method for the maxillary dentition distalization by using mini implant inserted in the IZC using a sliding mechanism (116).

2.6 Maxillary Molar Intrusion

One of the most complicated malocclusions is the skeletal anterior open bite, in which overgrowth of the maxillary and mandibular posterior dentoalveolar heights is the main characteristic feature, which in turn result in longer vertical facial dimension and a steeper mandibular plane (117). In order to solve open bite problem, the heights of posterior dentoalveolar regions should be decreased, which is difficult and needs the intrusion of the posterior molars.

Many methods for intrusion of the posterior teeth in young patients have been introduced such as passive bite blocks (118), active bite blocks with magnets or springs (119, 120), high-pull headgear, fixed appliances and vertical elastics (121, 122). Generally, these traditional techniques usually failed to intrude the molars, especially in adult patients. The only alternative treatment method and the most commonly recommended over many years was the surgical impaction of the maxilla, to obtain counterclockwise rotation of the mandible, thus reduction of anterior facial height in adult patients who complain from severe skeletal open bite (123). After invention of the skeletal anchorage devices, mini implant and miniplates have become very popular in treating cases with skeletal open bite in adults. Several studies have reported the successful treatment of anterior open bite using miniplates for intruding the mandibular or maxillary molars (18, 124-126).

3. MATERIALS AND METHODS

Two different orthodontic treatment scenarios were performed to compare the stress distribution between SS and Ti alloy infrazygomatic crest mini implants after force application by using FEA.

3.1 Study Design and Scenarios

- Two mini implants resemble in length (12 mm) and diameter (2 mm) with varied materials has been placed separately in the IZC region of a three-dimensional (3D) model that includes the maxillary and zygomatic bones.
- Mini implants used

a. Ortho Bone screw (mushroom 2.0) (iNewton Dental, Inc. Taywan, Taepei) which is made of stainless steel (SS).

b. Dentos Abso anchor (small head type SH 2018-12) (Dentos Inc. Taegu, Korea) which is made of titanium (Ti).

A B

Figure 1. (A) Ortho Bone screw (mushroom 2.0); (B) Dentos Abso anchor (small head type SH 2018-12).

Each mini implant has been inserted into the IZC region just above the maxillary first molar at angle of 70^0 to the occlusal plane and 14 mm above the occlusal plane, the insertion depth for both mini implants was 8 mm so 4 mm was stayed out of the bone.

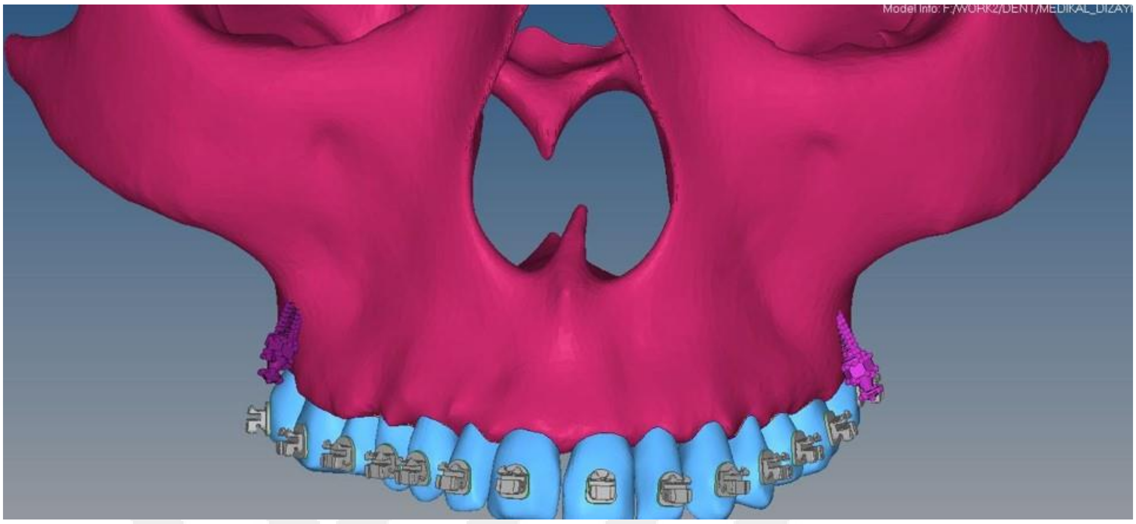

Figure 2. Frontal view of completed maxillary and zygomatic model after mini implants placement

Two different treatment scenarios carried out to establish and evaluate the stress variations while applying the same amount of force on both mini implants:

Molar distalization (Scenario 1)

Force of 300 g (3N) applied from the hook of maxillary canine to the head of mini implant placed in the IZC region using Nickel Titanium (NiTi) close coil. 0.017 X 0.025 SS arch wire has been used for sliding mechanism (Figure 3, 4, 5).

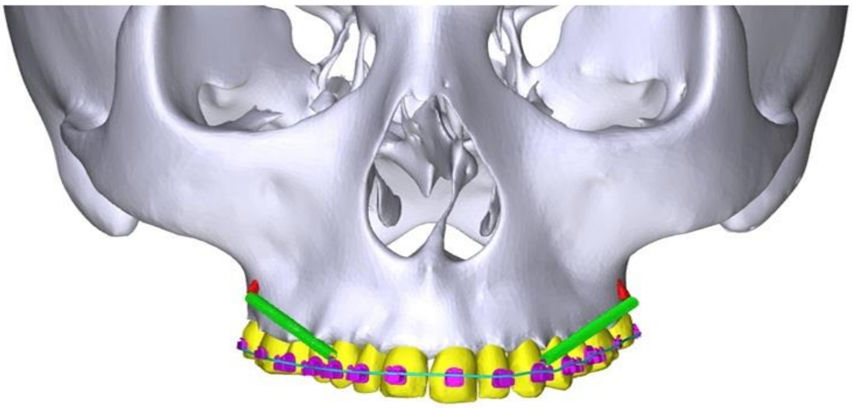

Figure 3. Frontal view of mini implant placement and force application in Scenario 1

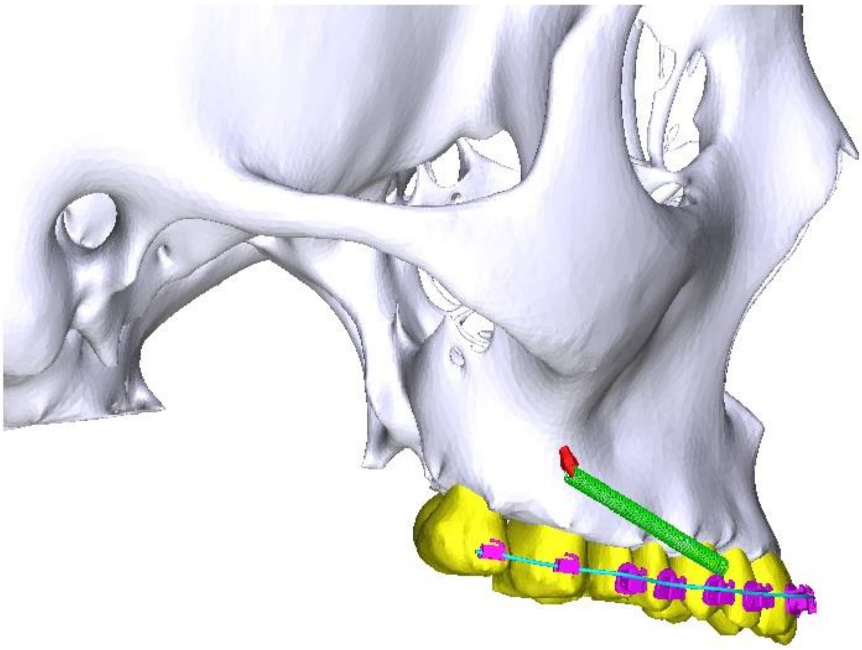

Figure 4. Lateral view of mini implant placement and force application in Scenario 1

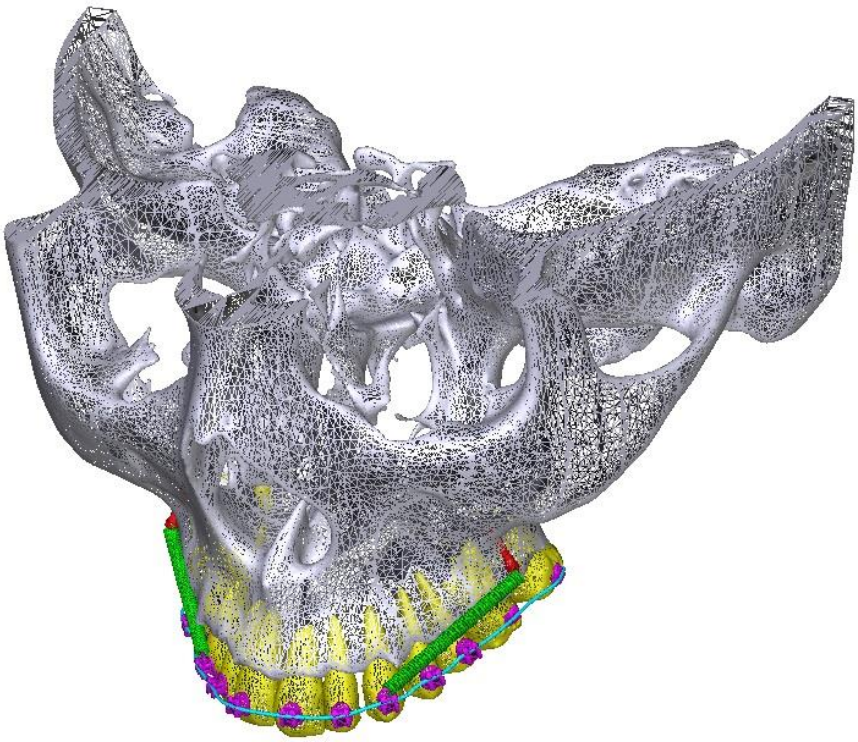

Figure 5. Scenario 1 represented with mesh overlay model in FEA

Molar intrusion (Scenario 2)

Force of 150 g (1.5 N) has been applied from the hook of the first molar to the mini implant placed in the IZC region using NiTi close coil (Figure 6, 7, 8).

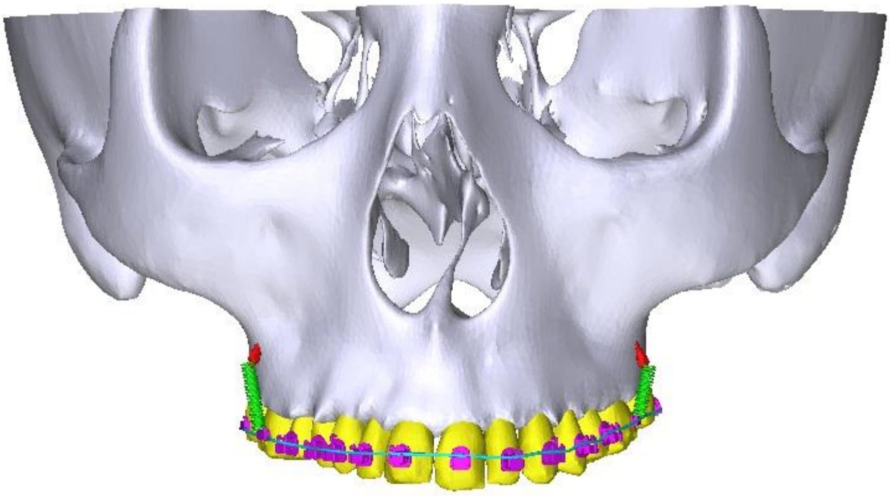

Figure 6. Frontal view of mini implant placement and force application in Scenario 2

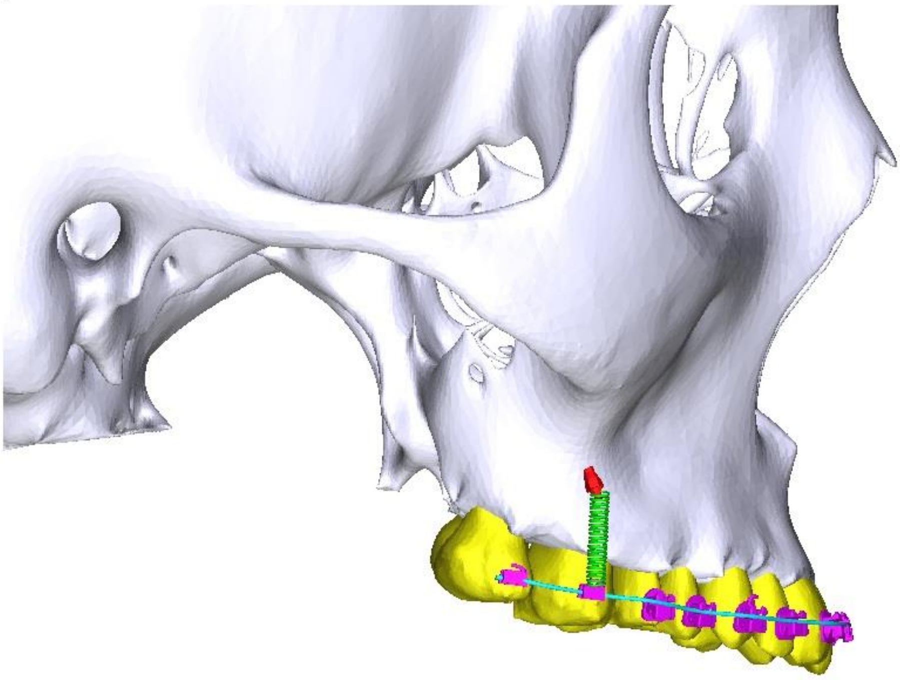

Figure 7. Lateral view of mini implant placement and force application in Scenario 2

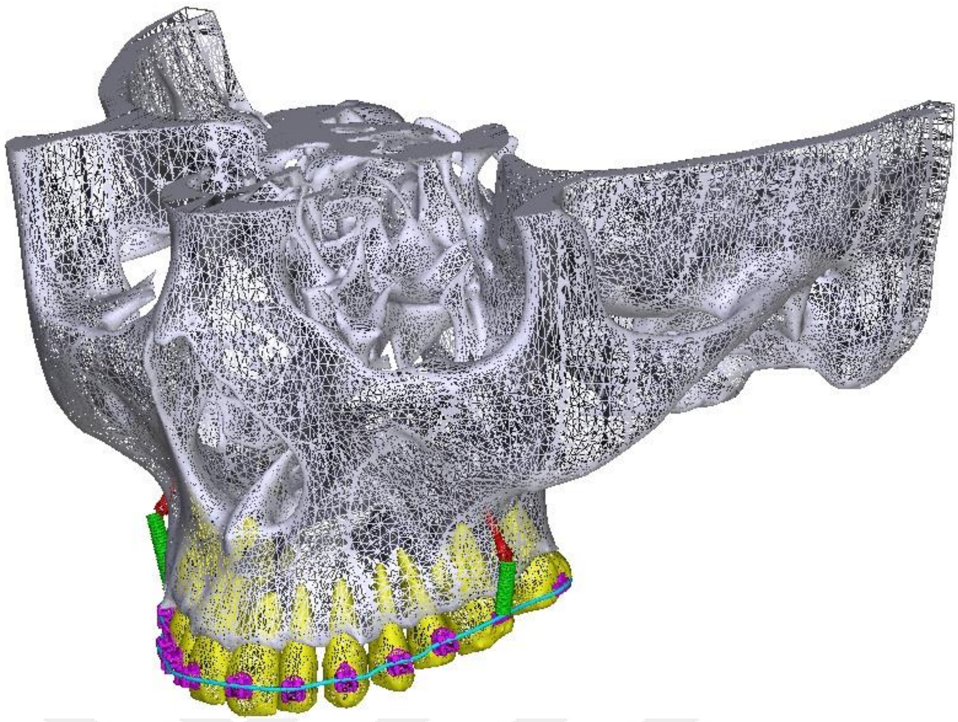

Figure 8. Scenario 2 represented with mesh overlay model in FEA

3.2 Preparation of Geometric Models and Construction of Boundary Network Structure

- 3D model of the zygomatic bone and maxilla including all maxillary teeth were constructed based on the Sobotta Human Anatomy Atlas and the data were prepared to FEA using the professional computer-aided design & computer-aided manufacturing (CAD/CAM) based software Computer Aided Three-Dimensional Interactive Application (CATIA), (Dassault Systèmes, France).
- All materials that used in the research (Ortho Bone screw mushroom 2.0, Dentos Abso anchor small head type SH 2018 – 12 screws, arc wire, brackets, NiTi coil springs) were also drawn and placed on the corresponding places on the 3D models according to the treatment scenarios by using the same software.
- Variations due to materials properties differences were also evaluated, subsequent models were created where mini implants material was varied from Ti to SS.
- The prepared 3D models then transferred to the Altair Hypermesh software in stl format in order to be analyzed. The stl format is the universal value for 3D

modelling programs. Because of securing the coordinate information of the nodes, no information loss occurs during the transfer of the data between different software programs owing to stl format. After rendering and making all the components compatible with each other and Altair Hypermesh, compositions of materials were introduced to the software. In our case these materials were the maxillary bone, teeth, mini implants, arch wire, brackets and NiTi coil spring.

- Material values (elasticity modulus and Poisson ratio) of each structure, which defines the physical characteristics of each model were provided (Table 1).
- Evaluation of the data in FEA program: As noted earlier, at this phase the effects of the different mechanics according to the treatment scenarios, examined on prepared 3D models. In this study, strain and displacement values occurring after activation of the applied mechanics monitored and saved. To reflect bone structures in different areas and of different density values better, a mesh system consisting of node elements (nodes) has been used. It was crucial to immobilize the model to perform the FEA. For this reason, the amount of mobility of the model constrained in appropriate places.
- After applying all material properties, the created Finite Element model was analysed with Dassault Systems Abaqus 6.14 _1 software. Three different types of stress were evaluated:

P Max: maximum principle stress (tensile stress).

P Min: minimum principle stress (compressive stress).

Von Mises stress (equivalent stress).

- P Max represents the peak value at point that the tensile stress in a material is exceeded. P Min represents the peak value of compressive stress resulting from a force applied to a material; the greater negative value of the stress means the greater 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 results in permanent deformation.

Table 1. The "Isotropic Material Properties" of the Materials used in the study.

Material	Young's modulus (MPa)	Poisson's ratio
Cortical bone	1.37×10^4	0.26
Cancellous bone	1.85×10^4	0.30
Teeth	1.96×10^4	0.30
Titanium	113000	0.34
Stainless Steel	200000	0.29
Ligature wire/braces	21×10^4	0.30
Mini implant	110	0.35

4. RESULTS

After force application in the mentioned scenarios, stress patterns on cortical and cancellous bone of IZC region, roots of the maxillary first molar and mini implant surfaces were captured. Then the peak values of stress of each model's stimulation were evaluated individually in both types of mini implant Ti and SS in three manners P Max, P Min and von Mises.

The results of all measurements were recorded as coloured images. In these images, each colour corresponds to a numerical value range. The colour codes corresponding to the range of values are indicated on the scale on the left side of the images.

4.1 Findings Related to Scenario 1

In this scenario 300 g (3N) force has been applied to the head of mini implant in purpose of molar distalization.

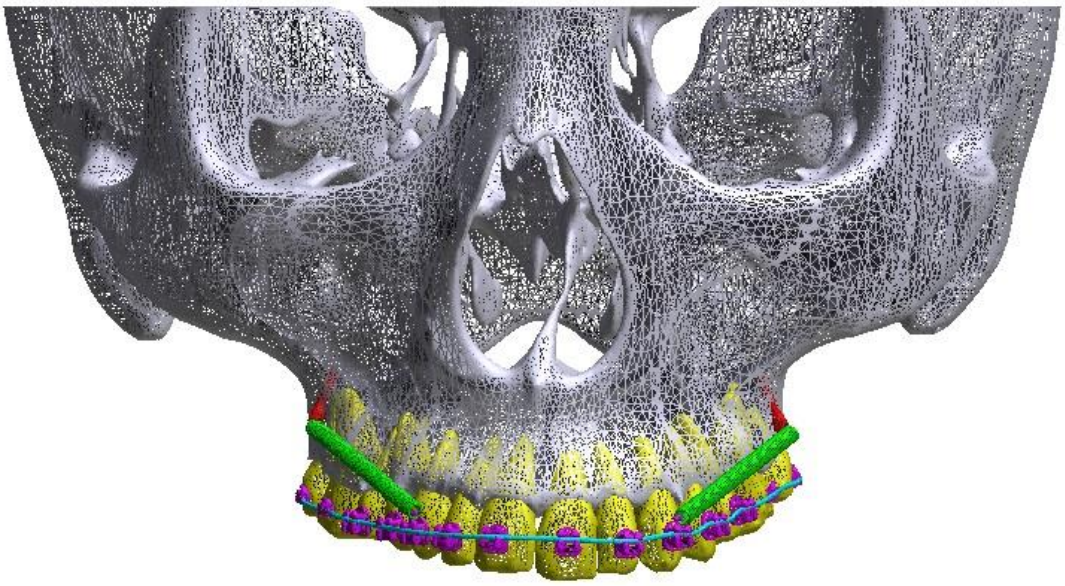

Figure 9. Scenario 1 represented with mesh overlay model in FEA

4.1.1 Findings related to the cortical bone of infrazygomatic region

Peak P Max value in the cortical bone measured as 18.6 MPa for Ti and 9.6 MPa for SS (Figure 10, 11, 12).

Peak P Min value in the cortical bone measured as -30.1 MPa for Ti and -9.8 MPa for SS (Figure 13, 14, 15).

Peak von Mises value in the cortical bone measured as 25.9 MPa for Ti and 10.3 MPa for SS (Figure 16, 17, 18).

When stress generated by both mini implants in the cortical bone compared, all stress types analysed showed higher stress value for Ti than SS mini implant (Table 2).

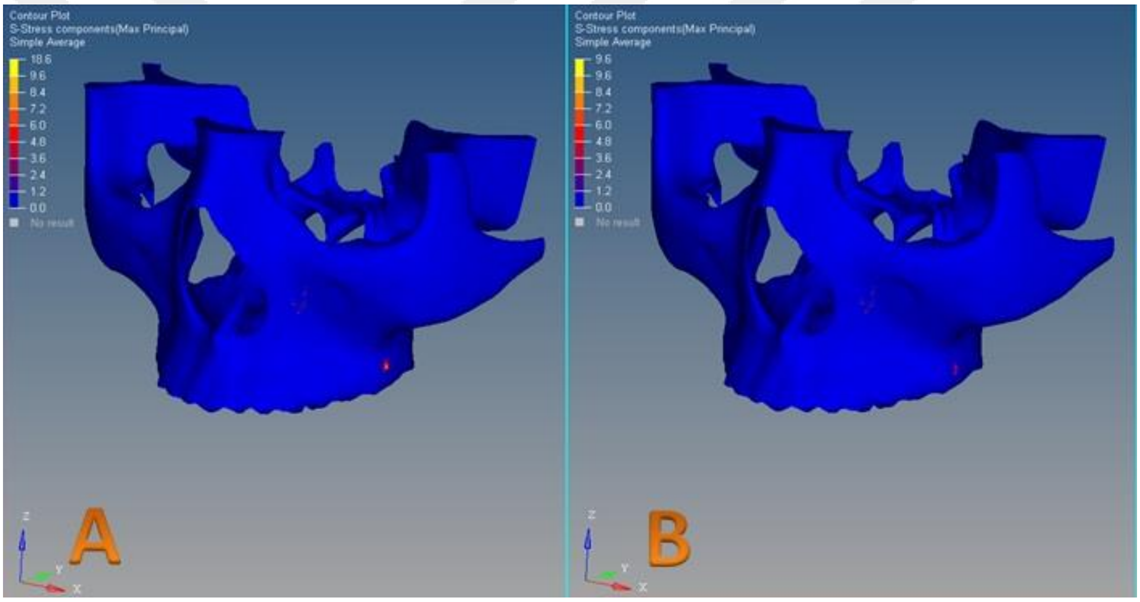

Figure 10. P Max stress patterns generated in cortical bone in Scenario 1: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

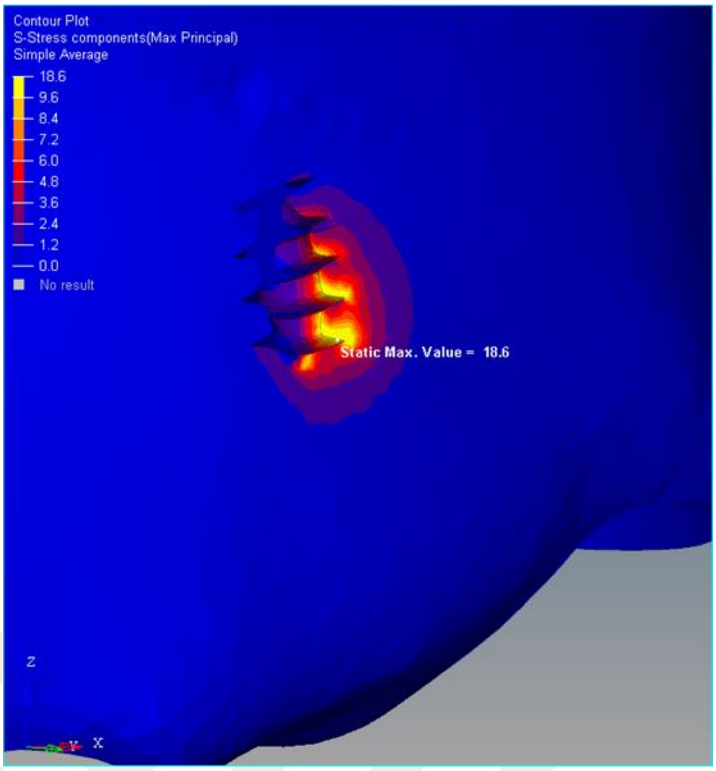

Figure 11. P Max stress patterns generated in cortical bone using Dentos Abso anchor Ti mini implant in Scenario 1

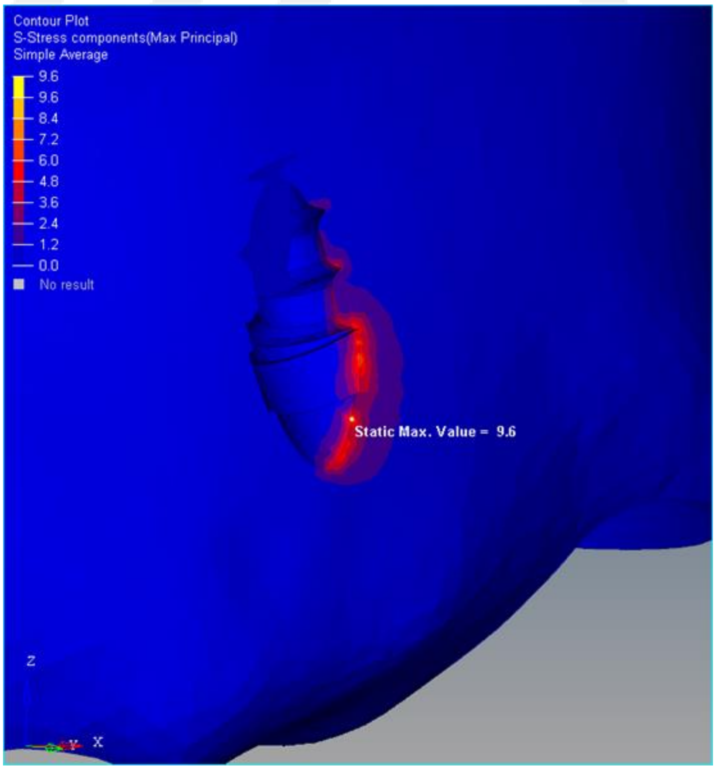

Figure 12. P Max stress patterns generated in cortical bone using Ortho Bone SS mini implant

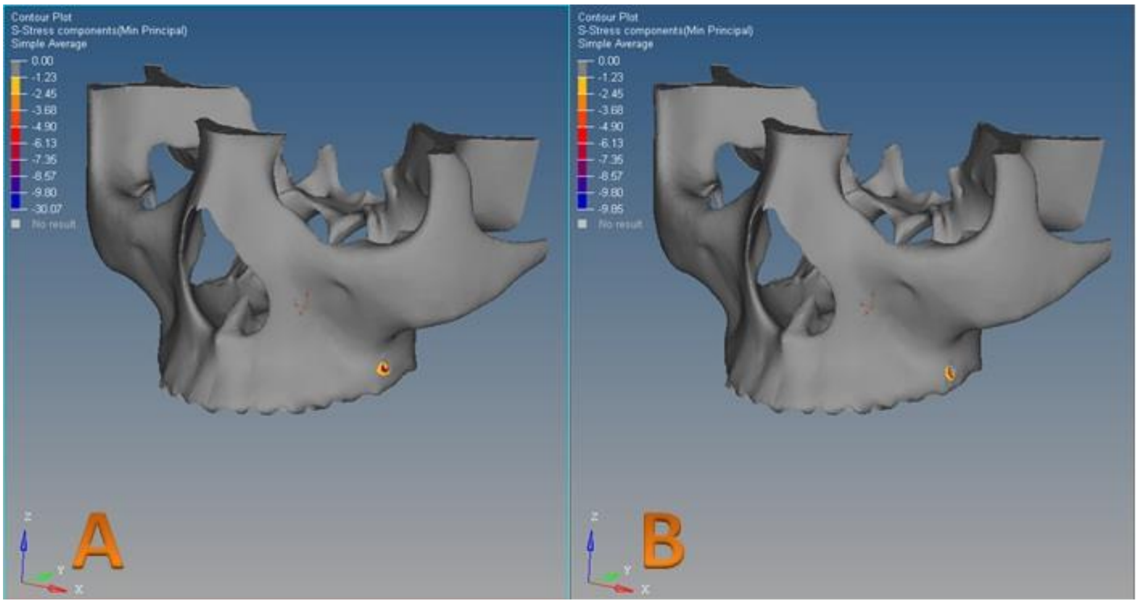

Figure 13. P Min stress patterns generated in cortical bone in Scenario 1: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

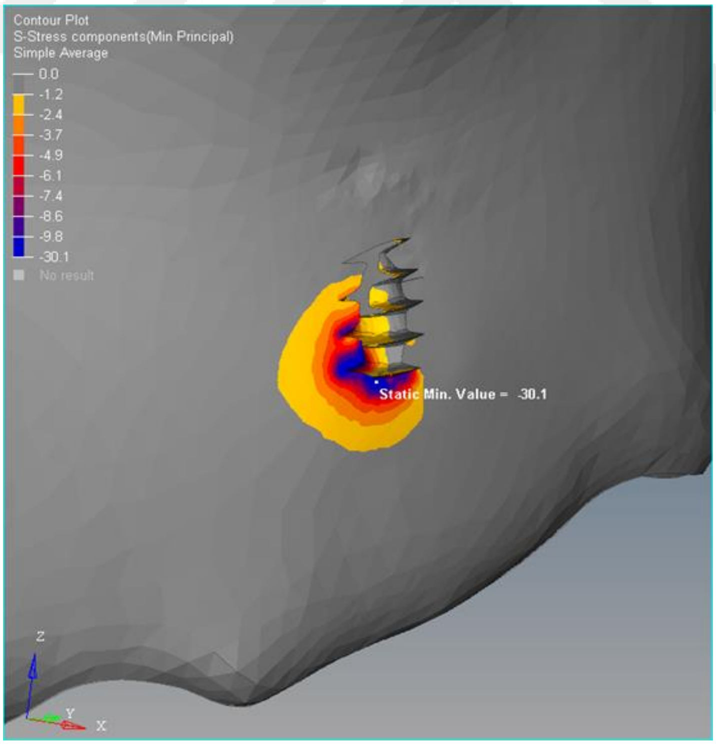

Figure 14. P Min stress patterns generated in cortical bone using Dentos Abso anchor Ti mini implant

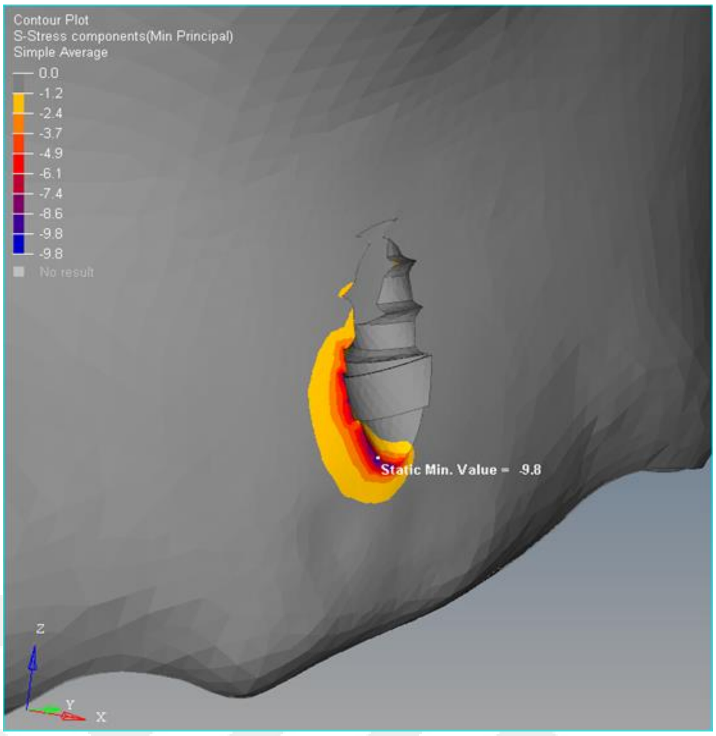

Figure 15. P Min stress patterns generated in cortical bone using Ortho Bone SS mini implant

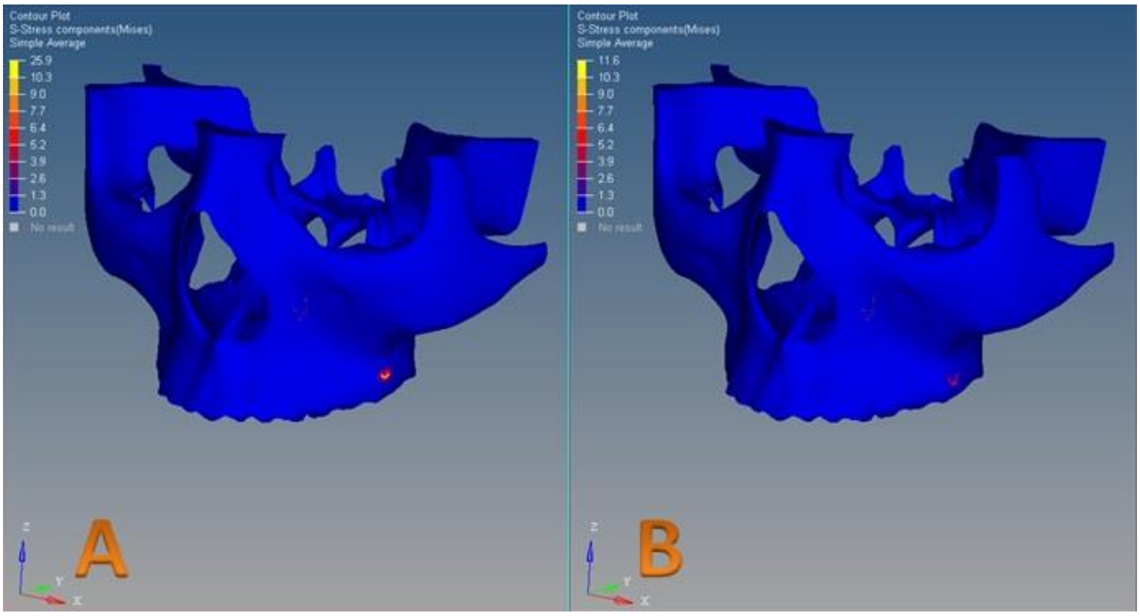

Figure 16. Von Mises stress patterns generated in cortical bone in Scenario 1: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

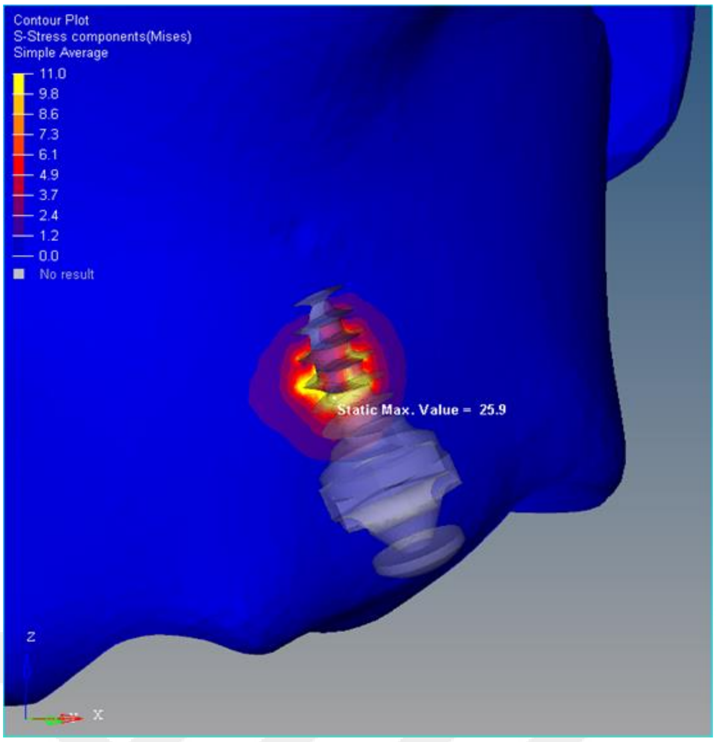

Figure 17. Von Mises stress patterns generated in cortical bone using Dentos Abso anchor mini implant

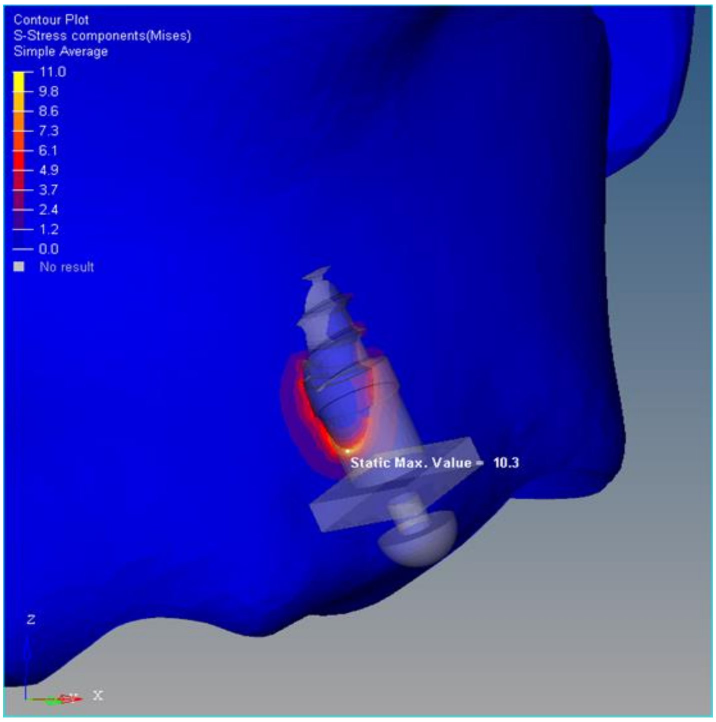

Figure 18. Von Mises stress patterns generated in cortical bone using Ortho Bone SS mini implant

4.1.2 Findings related to the cancellous bone of IZC region

Peak P Max value in the cancellous bone measured as 0.47 MPa for Ti and 0.19 MPa for SS (Figure 19, 20, 21).

Peak P Min value in the cancellous bone measured as -0.81 MPa for Ti and -0.29 MPa for SS (Figure 22, 23, 24).

Peak von Mises value in the cancellous bone measured as 0.91 MPa for Ti and 0.29 MPa for SS (Figure 25, 26, 27).

When stress generated by both mini implants in the cancellous bone compared, all stress types analysed showed higher stress value for Ti than SS mini implant (Table 2).

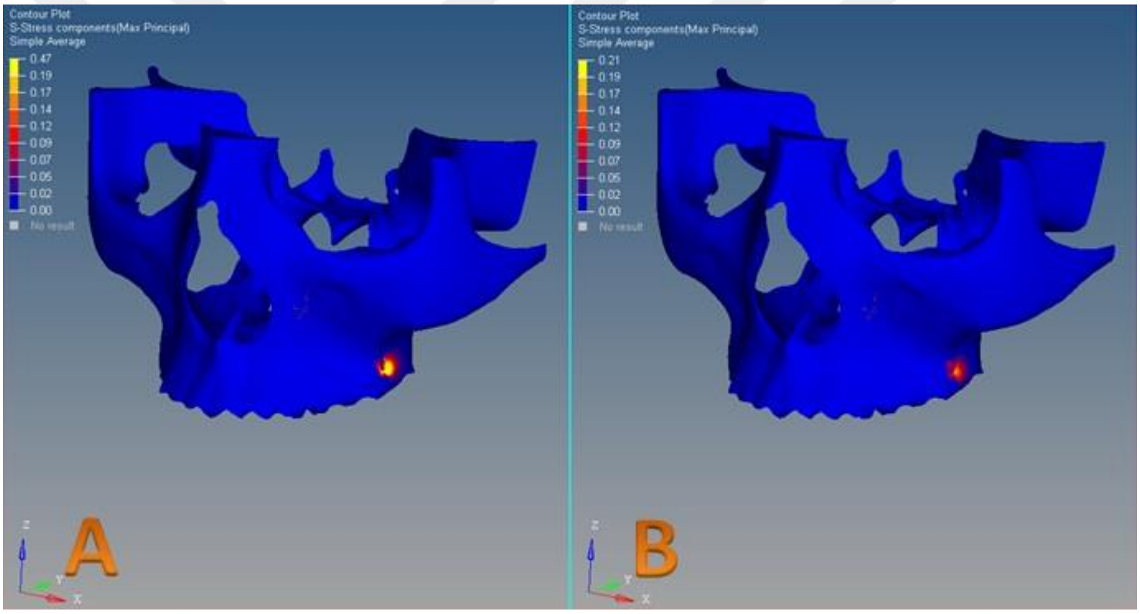

Figure 19. P Max stress patterns generated in cancellous bone in Scenario 1: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

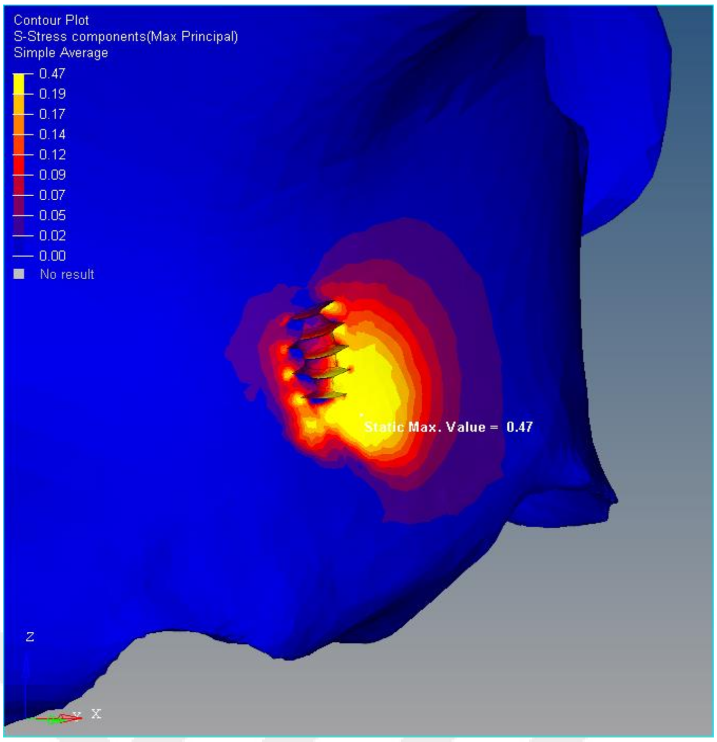

Figure 20. P Max stress patterns generated in cancellous bone using Dentos Abso anchor Ti mini implant

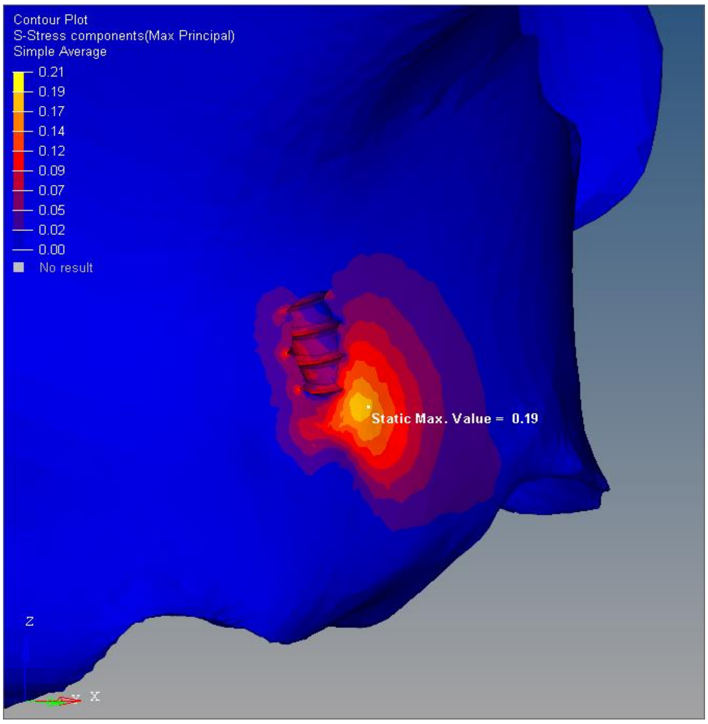

Figure 21. P Max stress patterns generated in cancellous bone using Ortho Bone SS mini implant

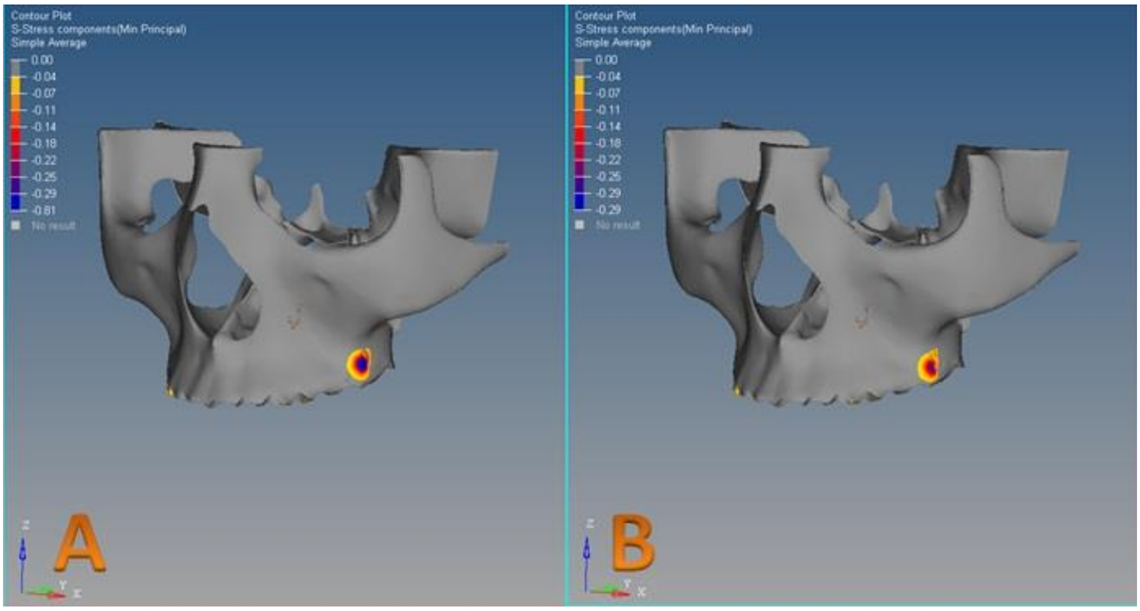

Figure 22. P Min stress patterns generated in cancellous bone in Scenario 1: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

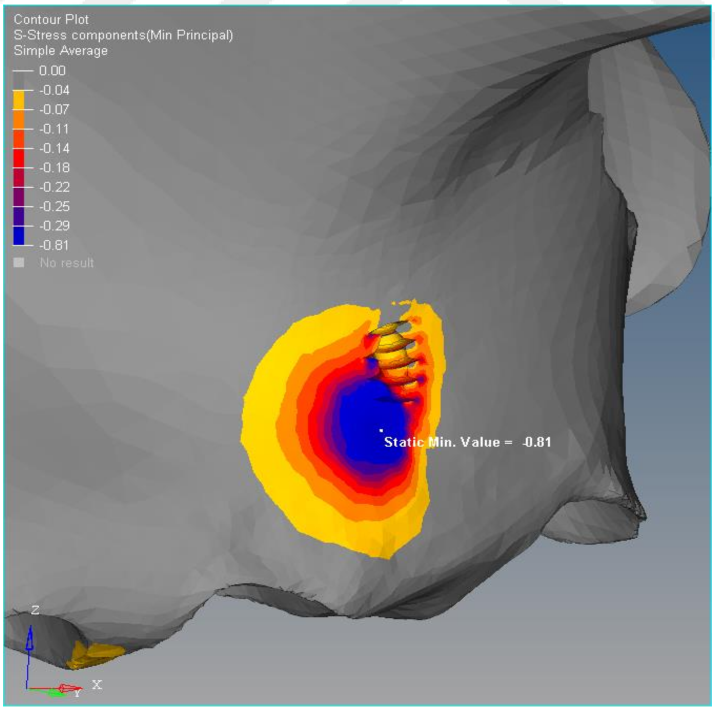

Figure 23. P Min stress patterns generated in cancellous bone using Dentos Abso anchor Ti mini implant

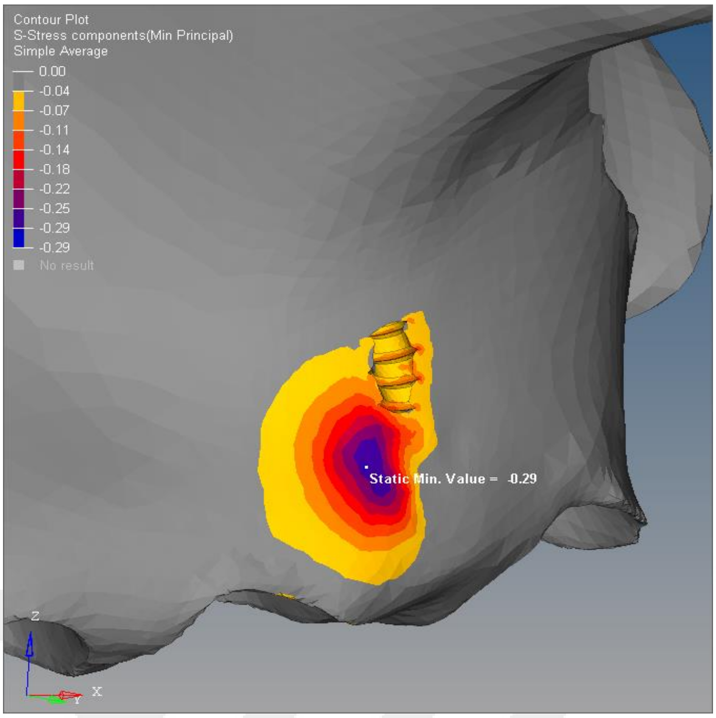

Figure 24. P Min stress patterns generated in cancellous bone using Ortho Bone SS mini implan.

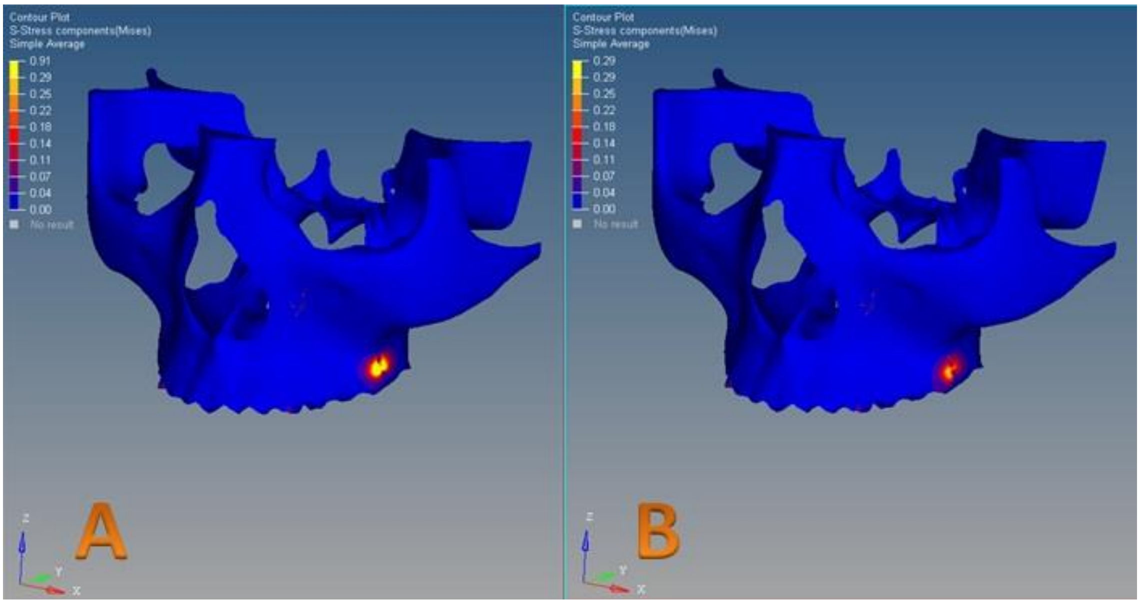

Figure 25. Von Mises stress patterns generated in cancellous bone in Scenario 1: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

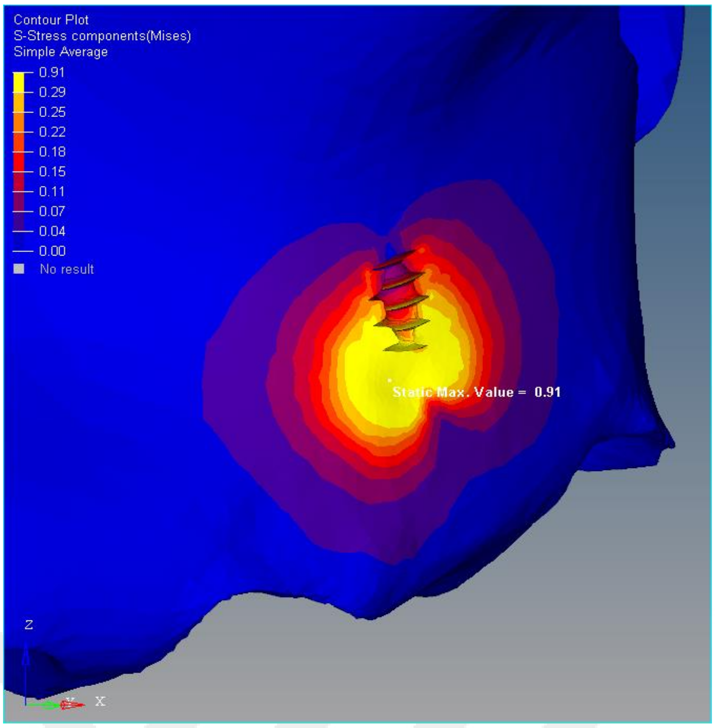

Figure 26. Von Mises stress patterns generated in cancellous bone using Dentos Abso anchor Ti mini implant

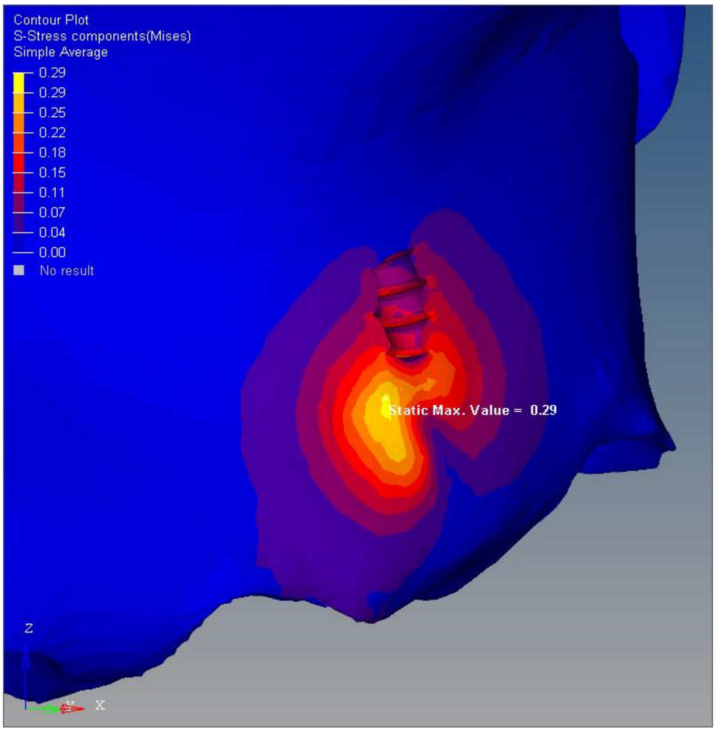

Figure 27. Von Mises stress patterns generated in cancellous bone using Ortho Bone SS mini implant

4.1.3 Findings related to mini implant

Peak von Mises value on the surfaces of mini implants measured as 100.0 MPa for Ti and 19.0 MPa for SS (Figure 28, 29, 30). Therefore, Ti mini implant has a higher stress than SS one when exposed to 300 g of force in case of distalization (Table 3).

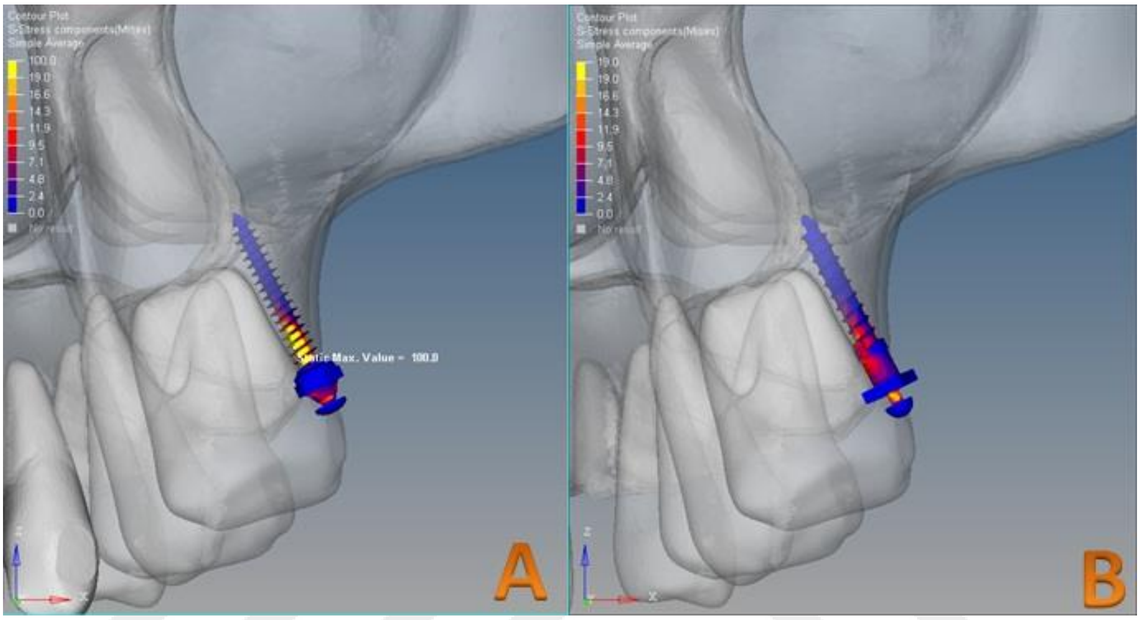

Figure 28. Von Mises stress patterns generated on mini mplant in Scenario 1: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

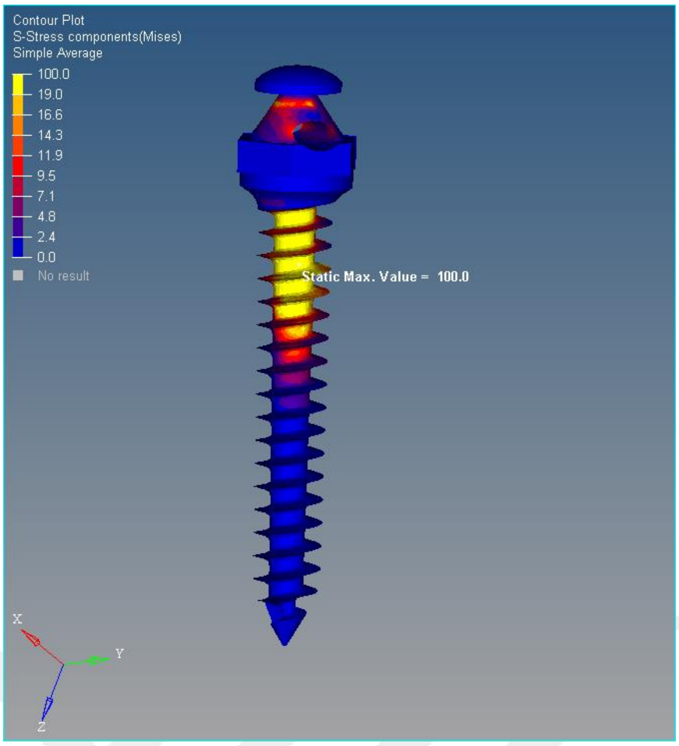

Figure 29. Von Mises stress patterns generated on the Dentos Abso anchor Ti mini implant

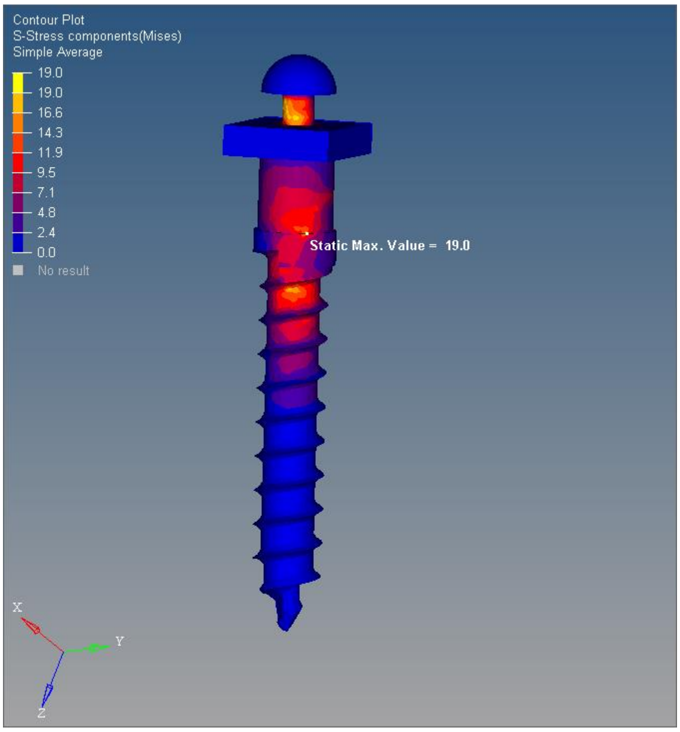

Figure 30. Von Mises stress patterns generated on the Ortho Bone SS mini implant

4.1.4 Findings related to the roots of maxillary first molar

Peak P Max value in the buccal roots of maxillary first molar measured as 0.07 MPa for Ti and 0.19 MPa for SS (Figure 31, 32).

Peak P Min value in the buccal roots of maxillary first molar measured as -0.16 MPa for Ti and -0.22 MPa for SS (Figure 33, 34).

Peak von Mises value in the buccal roots of measured as 0.16 MPa for Ti and 0.22 MPa for SS (Figure 35, 36).

Even though stress patterns observed in the buccal roots were low in value, SS mini implant showed higher stress generation when compared to Ti, while stress was not observed in the palatal root of maxillary first molar.

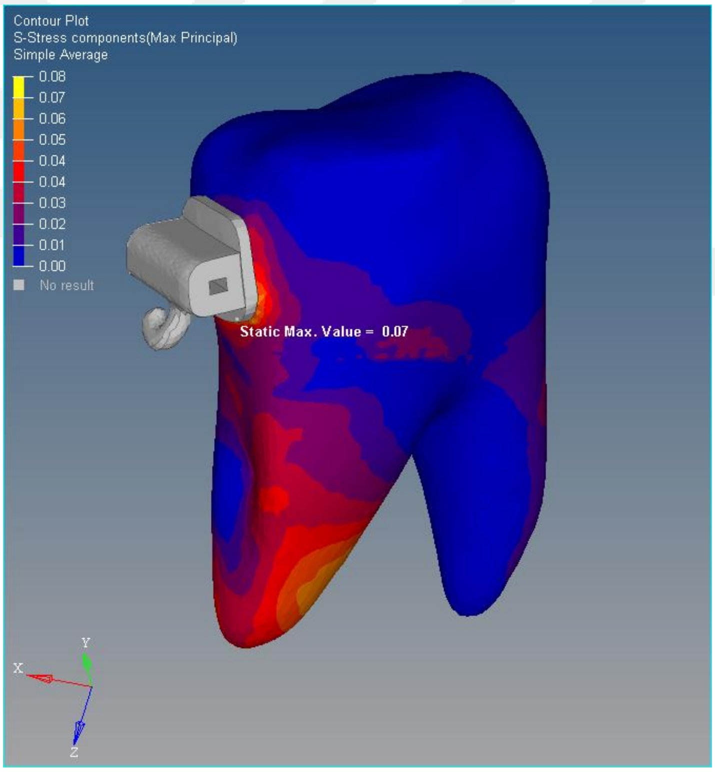

Figure 31. P Max stress patterns generated in the roots of maxillary first molar using Dentos Abso anchor Ti mini implant

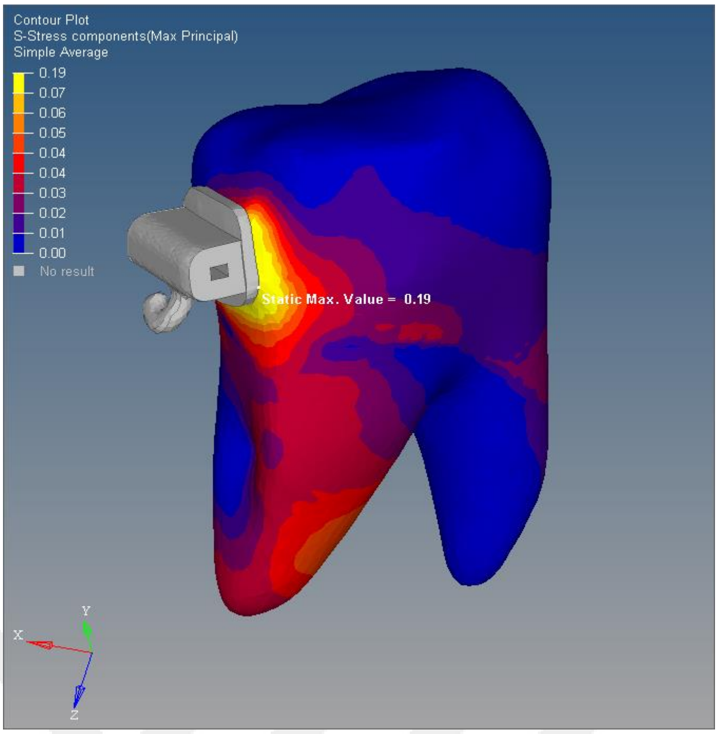

Figure 32. P Max stress patterns generated in the roots of maxillary first molar using Ortho Bone SS mini implant

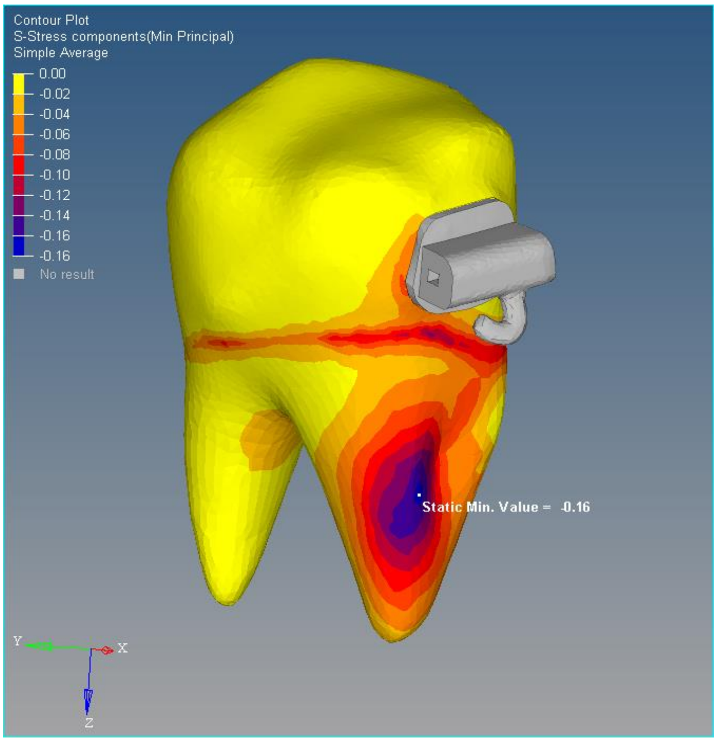

Figure 33. P Min stress patterns generated in the roots of maxillary first molar using Dentos Abso anchor Ti mini implant

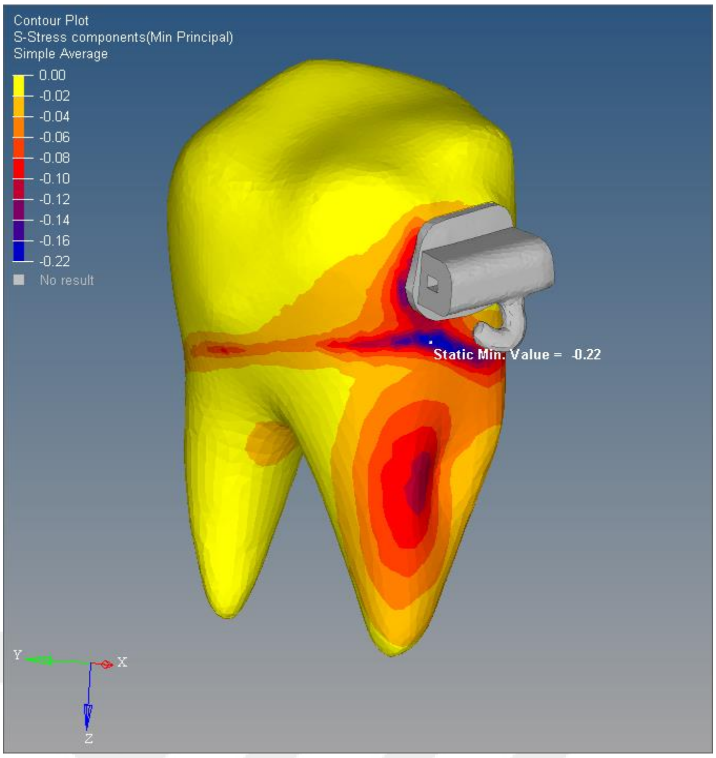

Figure 34. P Min stress patterns generated in the roots of maxillary first molar using Ortho Bone SS mini implant

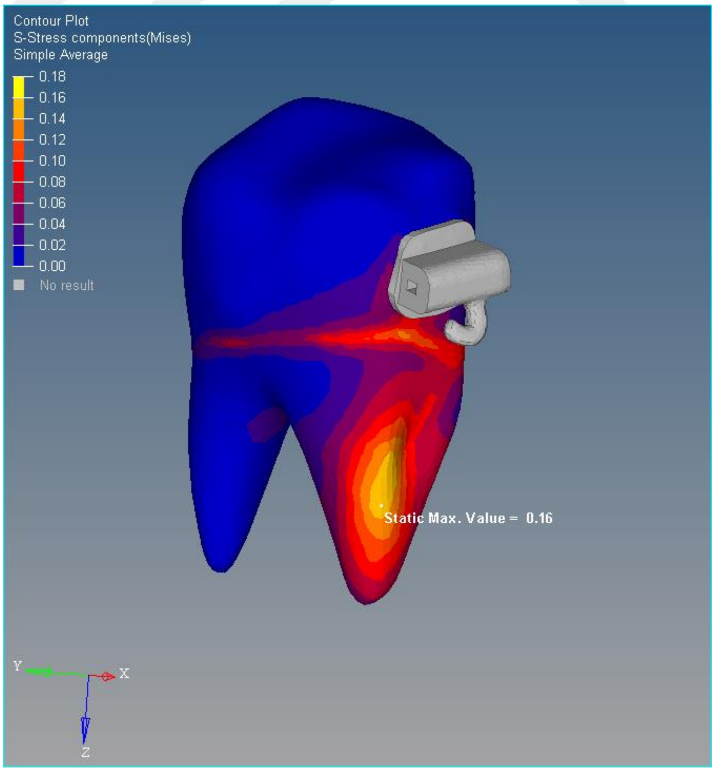

Figure 35. Von Mises stress patterns generated in the roots of maxillary first molar using Dentos Abso anchor Ti mini implant

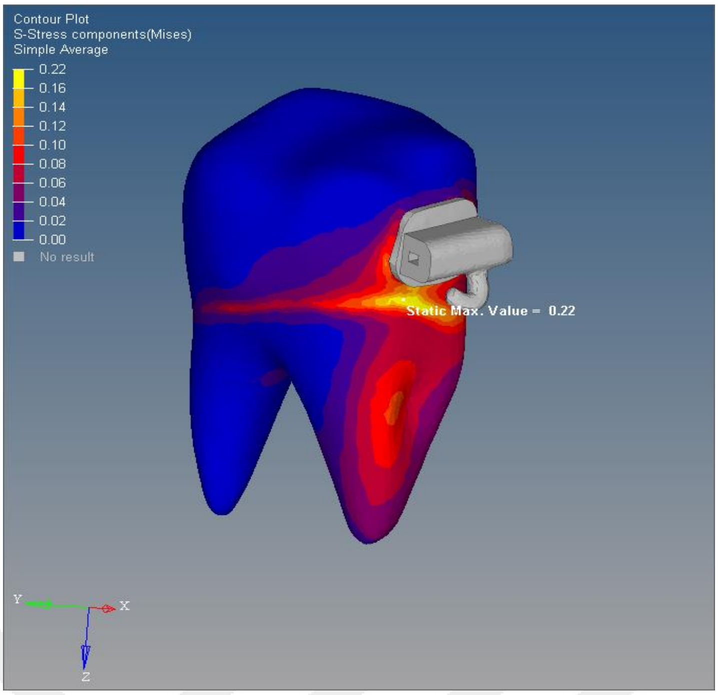

Figure 36. Von Mises stress patterns generated in the roots of maxillary first molar using Ortho Bone SS mini implant

Table 2. Comparision of Stress (MPa) in Cortical and cancellous bone between Ti and SS mini implants.

	Distalization			Intrusion		
	P Max	P Min	von Mises	P Max	P Min	von Mises
Titanium						
Cortical	18.6	-30.1	25.9	6.3	-15.4	11.6
Cancellous	0.47	-0.81	0.91	0.17	-0.46	0.50
Stainless steel						
Cortical	9.6	-9.8	10.3	2.5	-5.2	4.1
Cancellous	0.19	-0.29	0.29	0.07	-0.17	0.16

Table 3. Comparision of Stress (MPa) on mini implants between Ti and SS.

	Distalization	Intrusion
	von Mises	von Mises
Titanium	100.0	40.4
Stainless steel	19.0	7.9

4.2 Findings Related to Scenario 2

In this scenario 150 g (1.5N) force has been applied to the head of mini implant in purpose of molar intrusion.

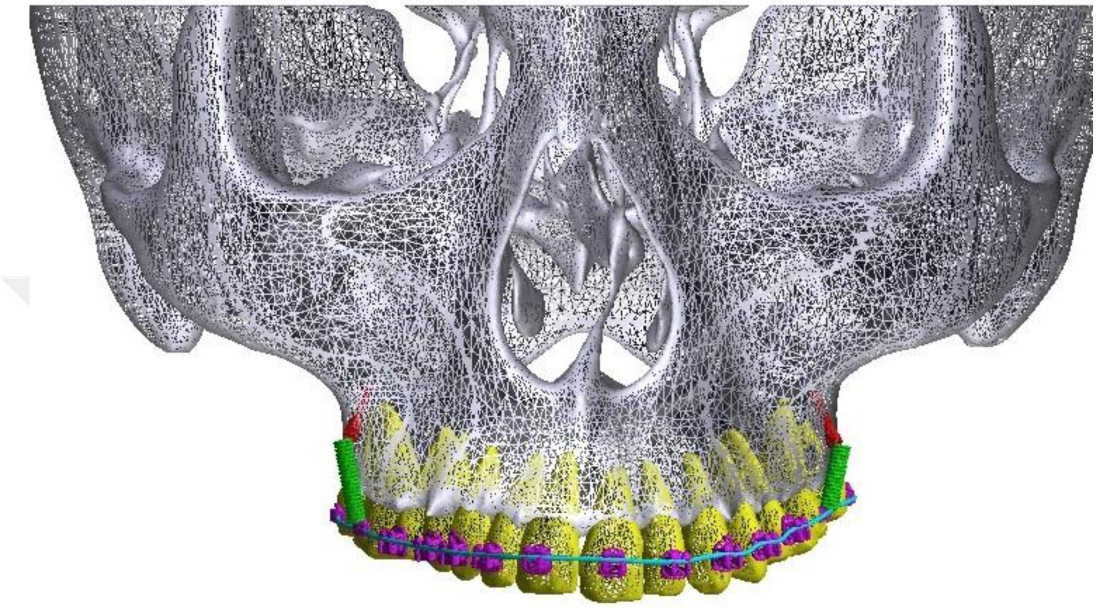

Figure 37. Scenario 2 represented with mesh overlay model in FEA

4.2.1 Findings related to the cortical bone of IZC region

Peak P Max value in the cortical bone measured as 6.3 MPa for Ti and 2.5 MPa for SS (Figure 38, 39, 40).

Peak P Min value in the cortical bone measured as -15.4 MPa for Ti and -5.2 MPa for SS (Figure 41, 42, 43).

Peak von Mises value in the cortical bone measured as 11.6 MPa for Ti and 4.1 MPa for SS (Figure 44, 45, 46).

When stress generated by both mini implants in the cortical bone compared, all stress types analysed showed higher stress value for Ti than SS mini implant (Table 2).

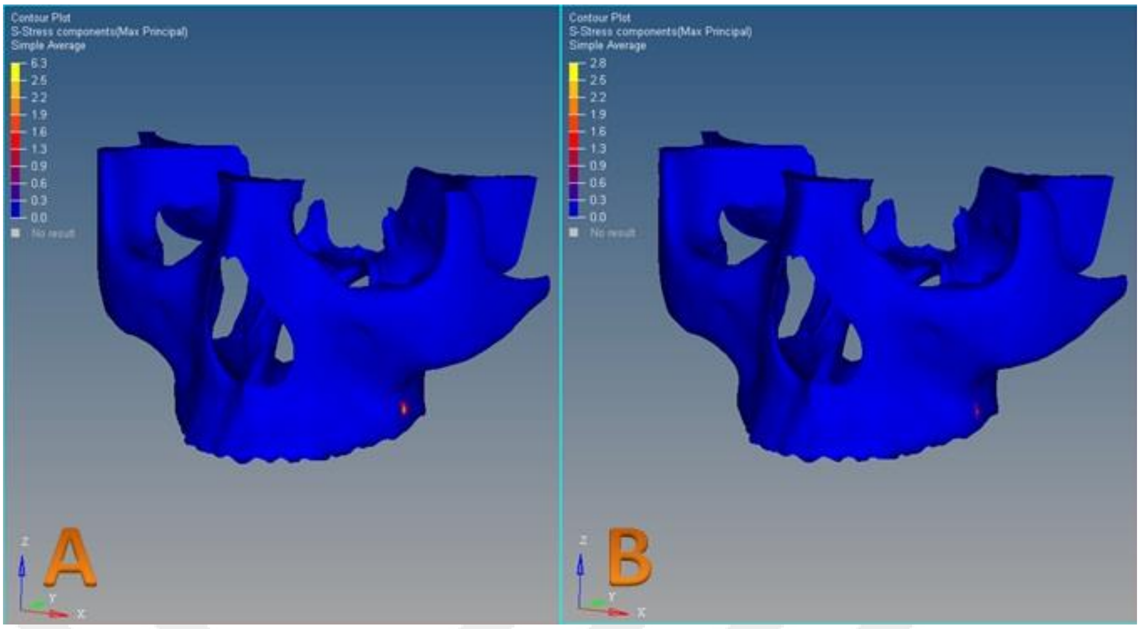

Figure 38. P Max stress patterns generated in cortical bone in Scenario 2: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

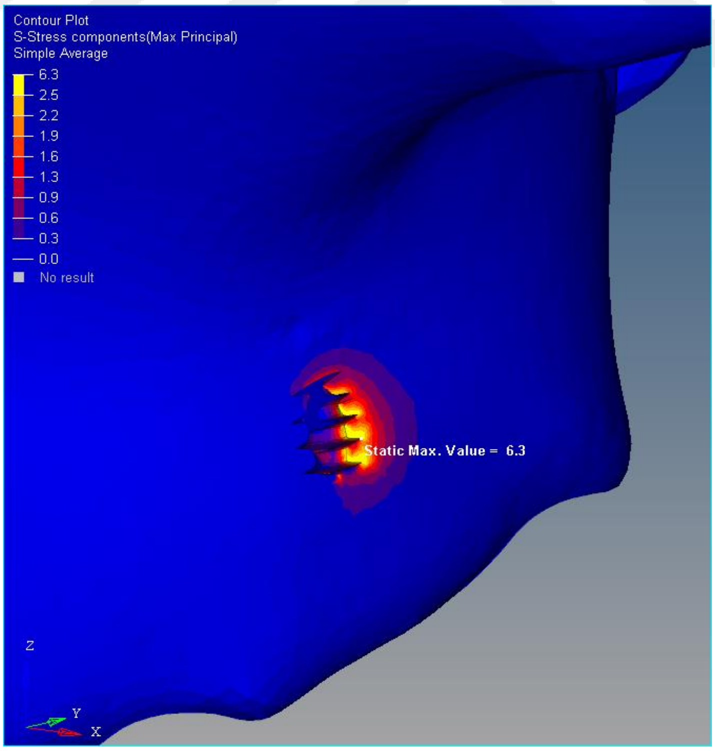

Figure 39. P Max stress patterns generated in cortical bone using Dentos Abso anchor Ti mini implant

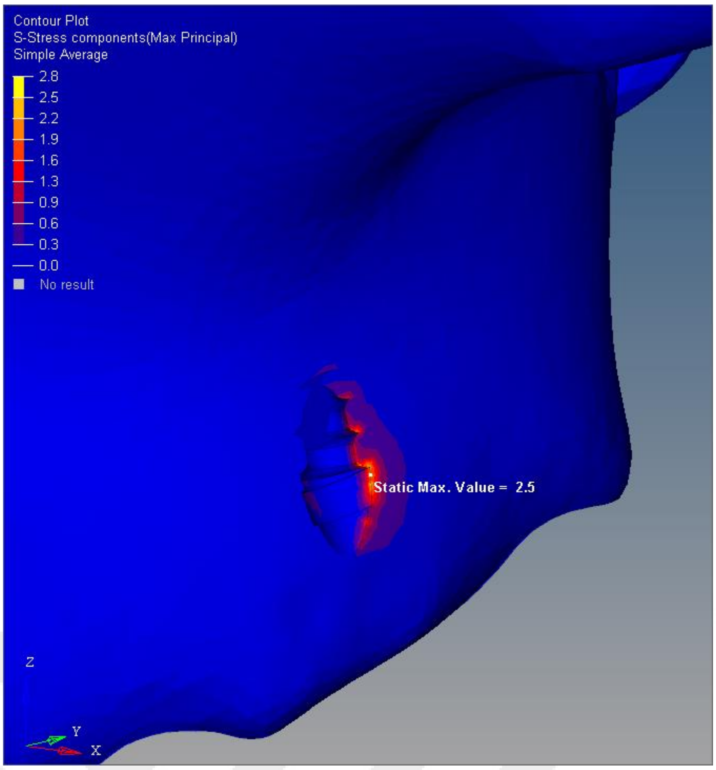

Figure 40. P Max stress patterns generated in cortical bone using Ortho Bone SS mini implant

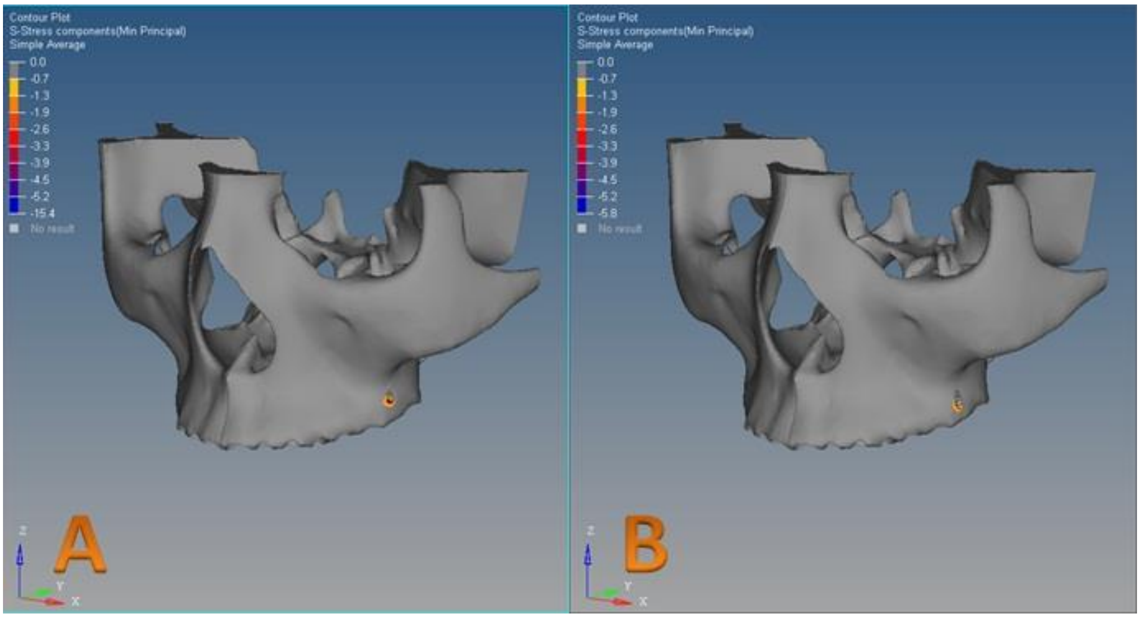

Figure 41. P Min stress patterns generated in cortical bone in Scenario 2: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

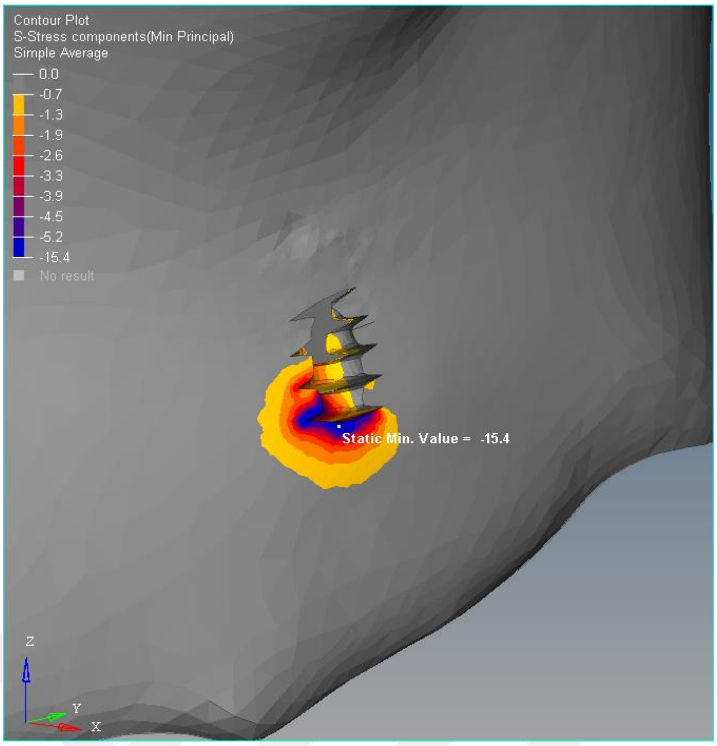

Figure 42. P Min stress patterns generated in cortical bone using Dentos Abso anchor Ti mini implant

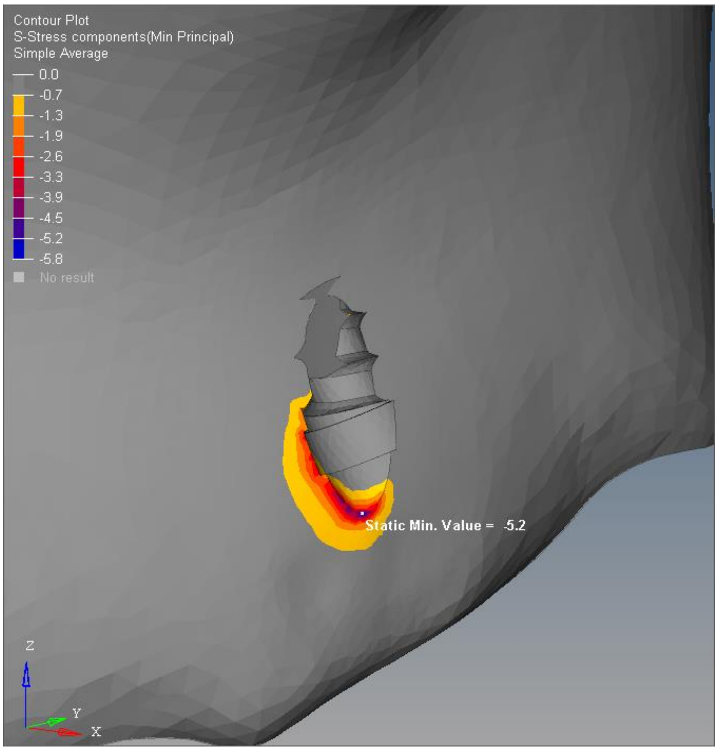

Figure 43. P Min stress patterns generated in cortical bone using Ortho Bone SS mini implant

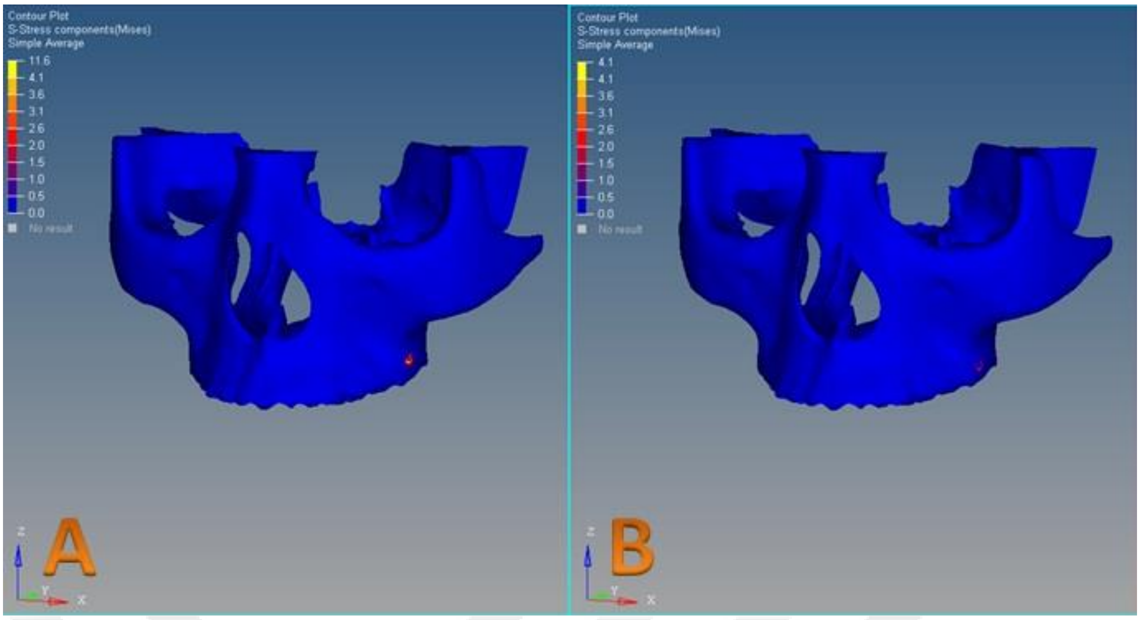

Figure 44. Von Mises stress patterns generated in cortical bone in Scenario 2: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

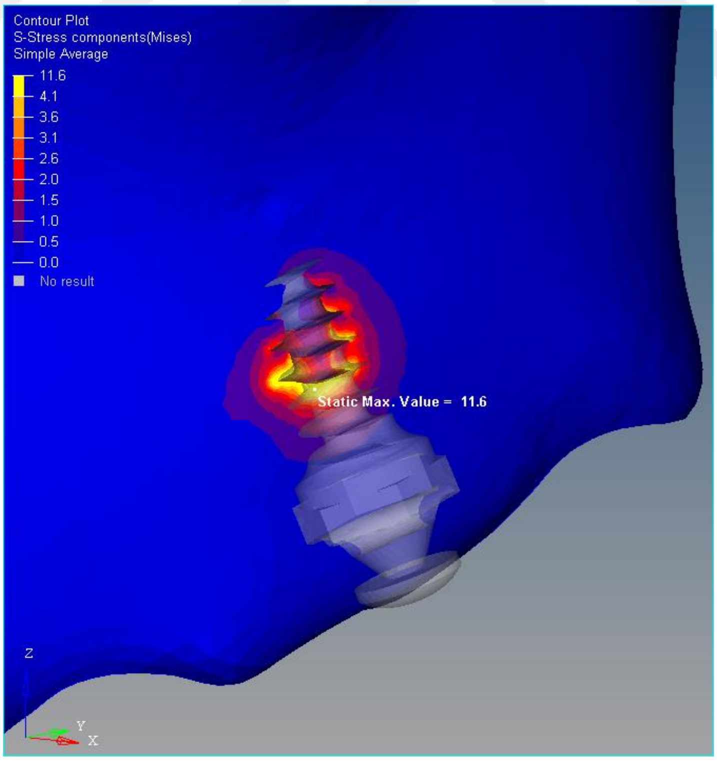

Figure 45. Von Mises stress patterns generated in cortical bone using Dentos Abso anchor Ti mini implant

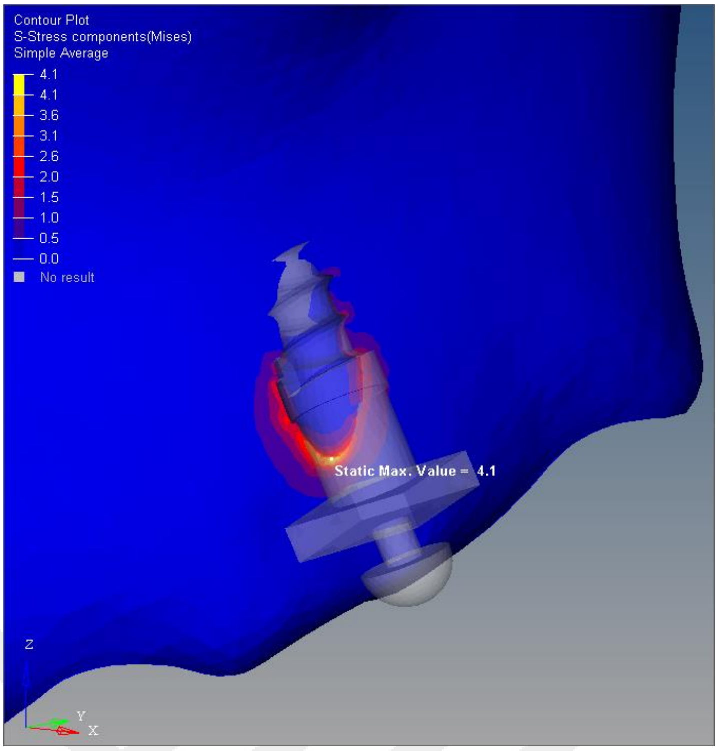

Figure 46. Von Mises stress patterns generated in cortical bone using Ortho Bone SS mini implant

4.2.2 Findings related to the cancellous bone of IZC region

Peak P Max value in the cancellous bone measured as 0.17 MPa for Ti and 0.07 MPa for SS (Figure 47, 48, 49).

Peak P Min value in the cancellous bone measured as -0.46 MPa for Ti and -0.17 MPa for SS (Figure 50, 51, 52).

Peak von Mises value in the cancellous bone measured as 0.50 MPa for Ti and 0.16 MPa for SS (Figure 53, 54, 55).

When stress generated by both mini implants in the cancellous bone compared, all stress types analysed showed higher stress value for Ti than SS mini implant (Table 2).

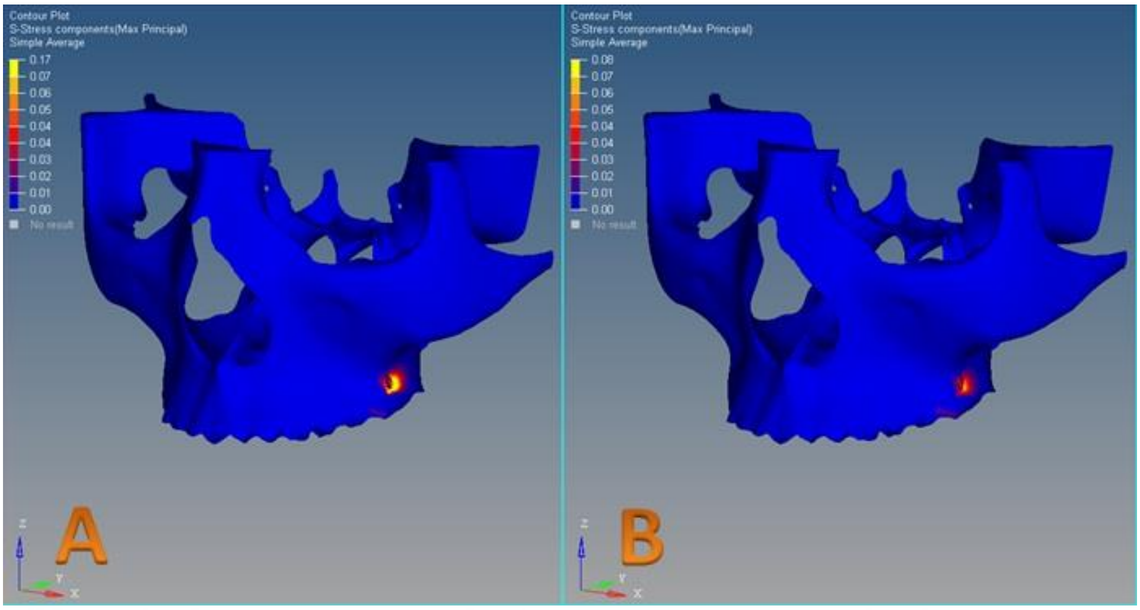

Figure 47. P Max stress patterns generated in cancellous bone in Scenario 2: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

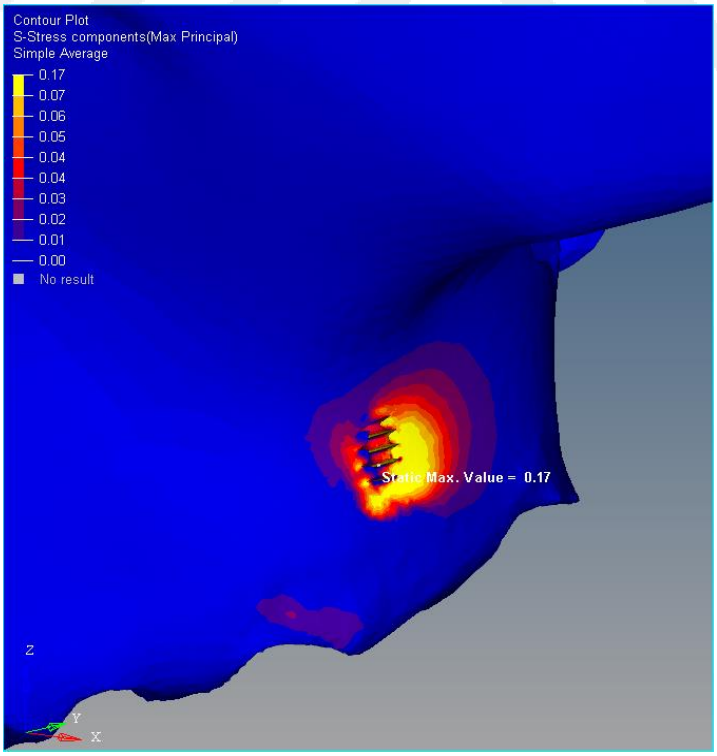

Figure 48. P Max stress patterns generated in cancellous bone using Dentos Abso anchor Ti mini implant

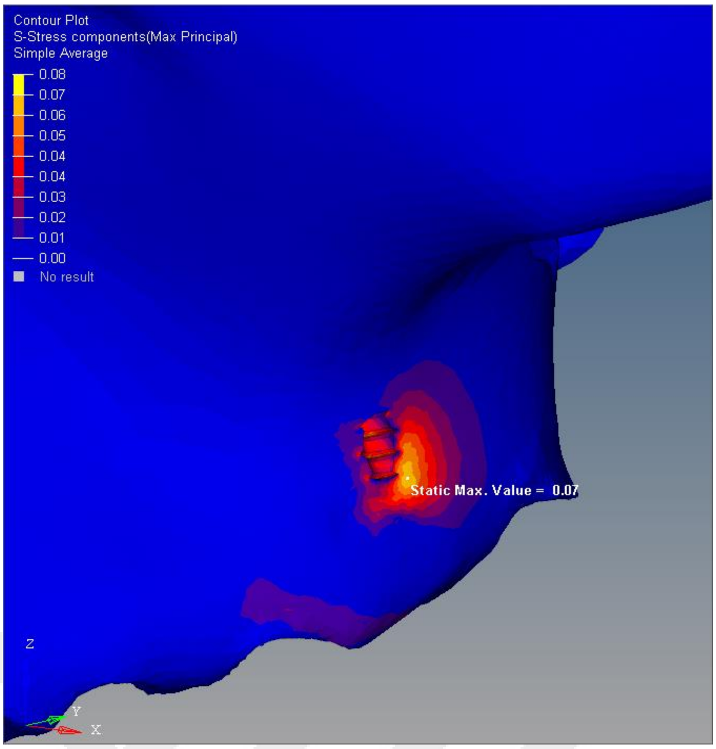

Figure 49. P Max stress patterns generated in cancellous bone using Ortho Bone SS mini implant

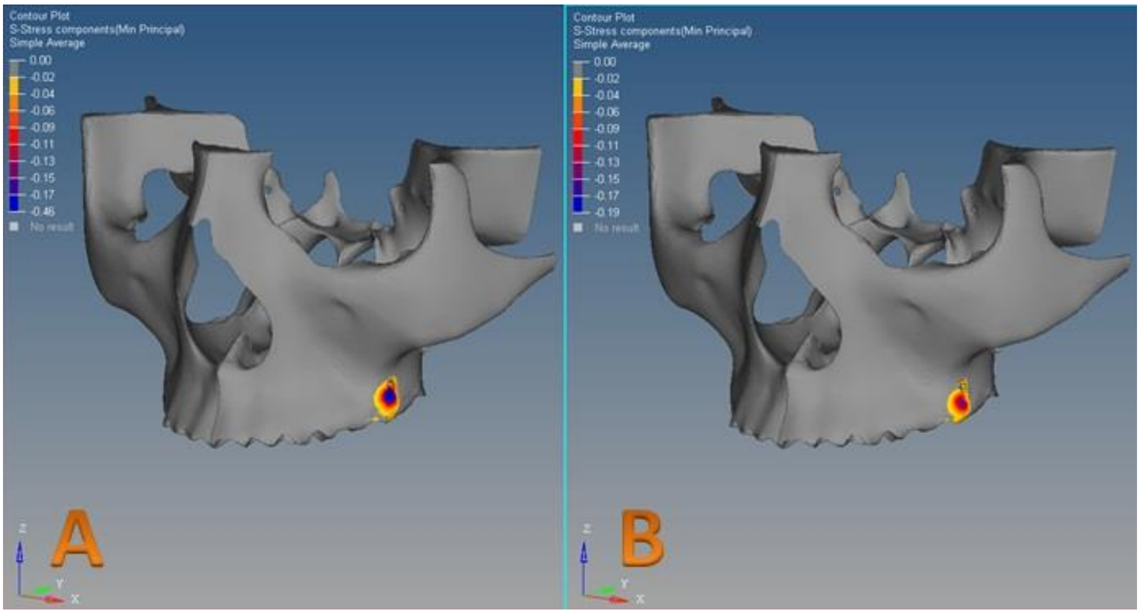

Figure 50. P Min stress patterns generated in cancellous bone in Scenario 2: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

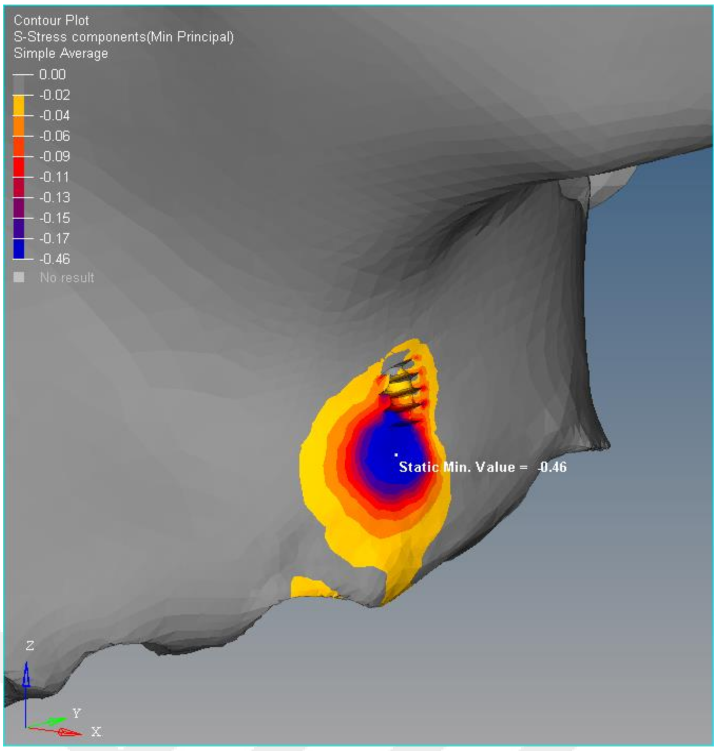

Figure 51. P Min stress patterns generated in cancellous bone using Dentos Abso anchor Ti mini implant

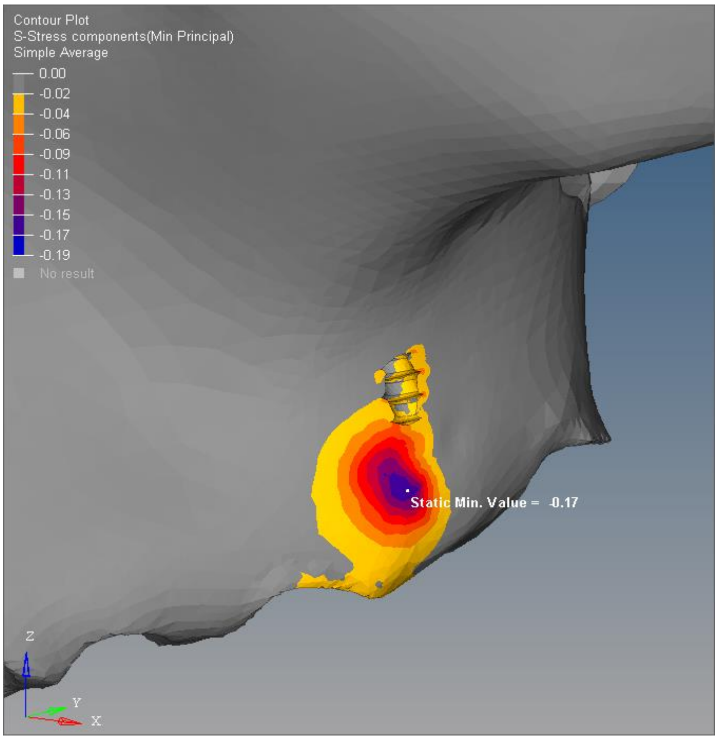

Figure 52. P Min stress patterns generated in cancellous bone using Ortho Bone SS mini implant

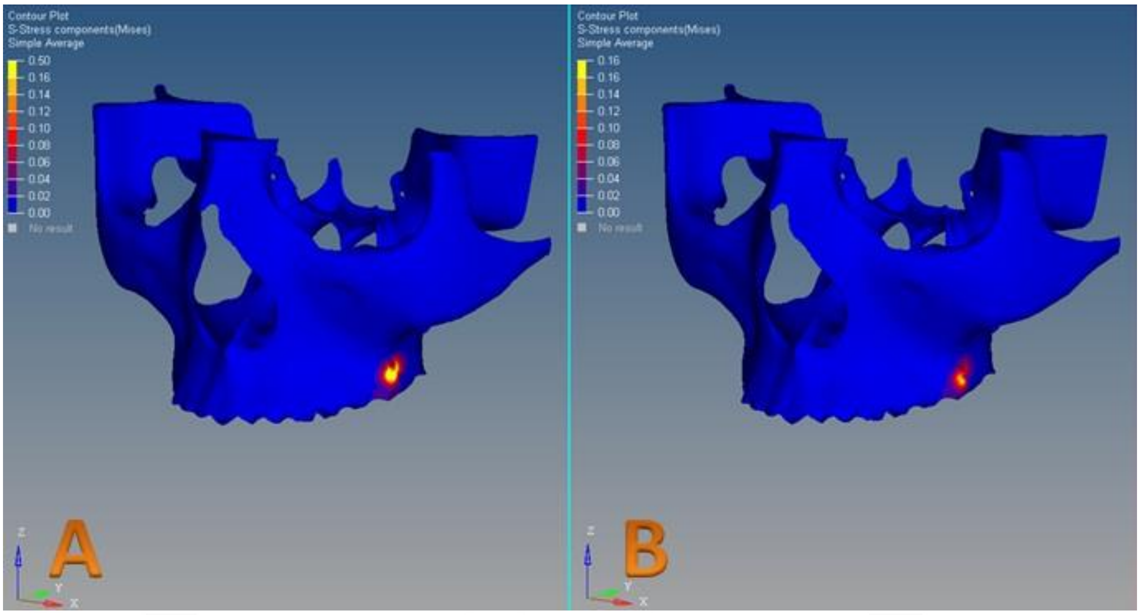

Figure 53. Von Mises stress patterns generated in cancellous bone in Scenario 2: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

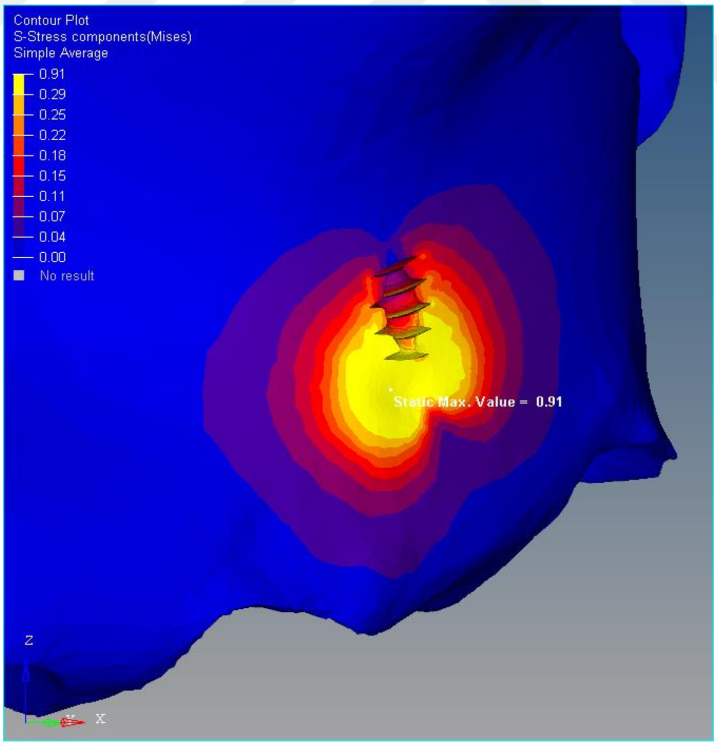

Figure 54. Von Mises stress patterns generated in cancellous bone using Dentos Abso anchor Ti mini implant

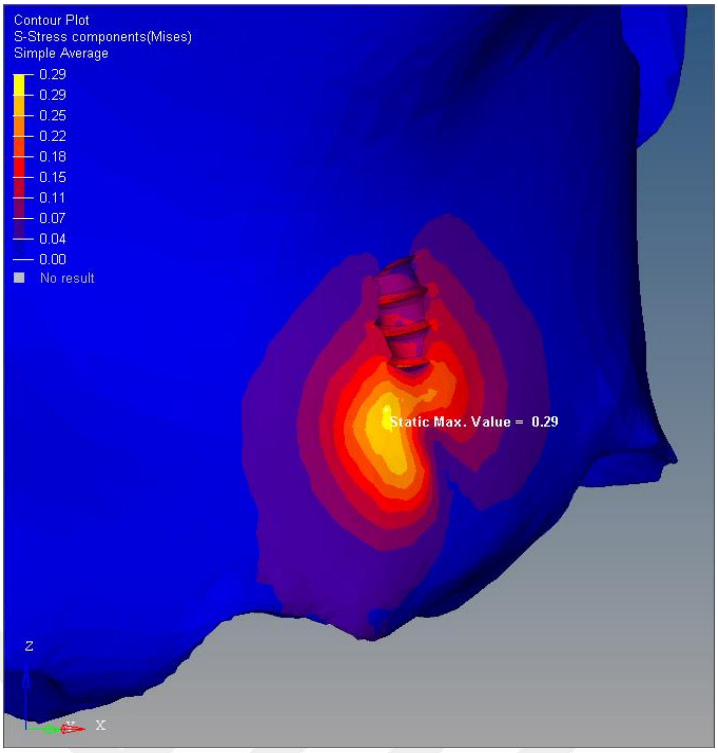

Figure 55. Von Mises stress patterns generated in cancellous bone using Ortho Bone SS mini implant

4.2.3 Findings related to mini implant

Peak von Mises value on the surfaces of mini implants measured as 40.4 MPa for Ti and 7.9 MPa for SS (Figure 56, 57, 58). Therefore, Ti mini implant has a higher stress than SS one when exposed to 150 g of force in case of intrusion (Table 3).

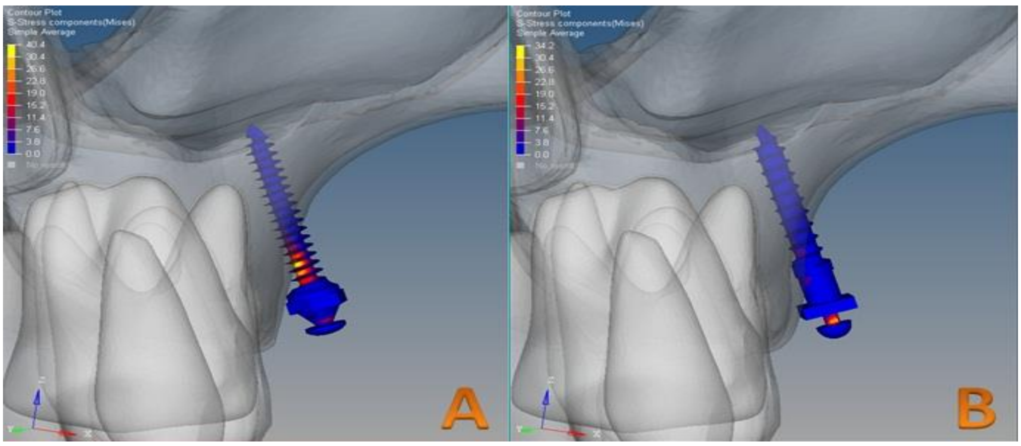

Figure 56. Von Mises stress patterns generated on mini implants in Scenario 2: (A) Ti Dentos Abso anchor; (B) SS Ortho Bone

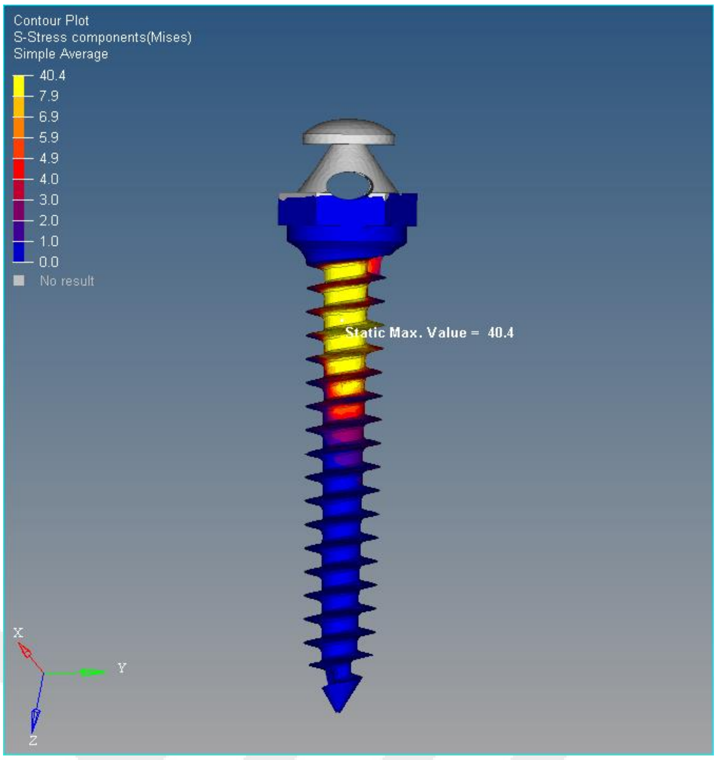

Figure 57. Von Mises stress patterns generated on the Dentos Abso anchor Ti mini implant

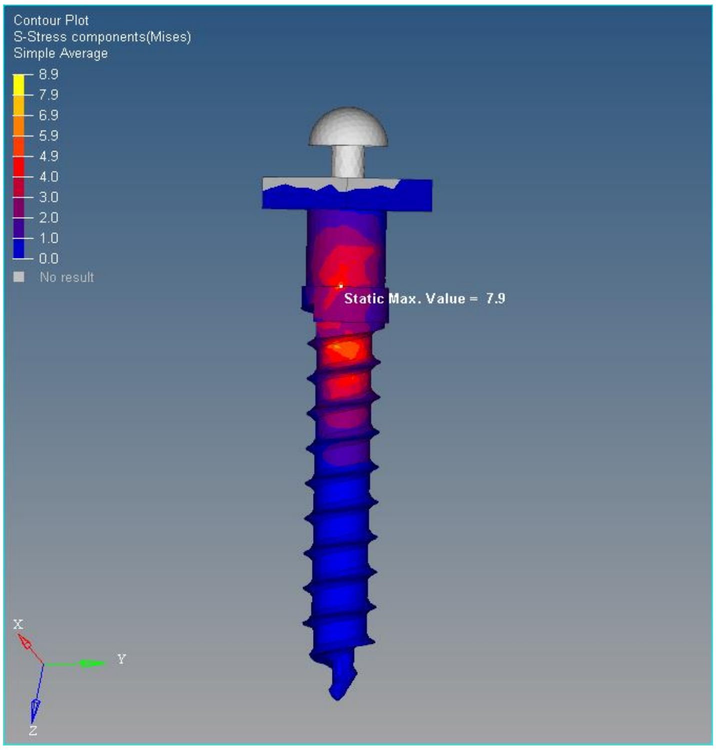

Figure 58. Von Mises stress patterns generated on the Ortho Bone SS mini implant

4.2.4 Findings related to the roots of maxillary first molar

Peak P Max value in the buccal roots of maxillary first molar measured as 0.47 MPa for Ti and 0.42 MPa for SS (Figure 59, 60).

Peak P Min value in the buccal roots of maxillary first molar measured as -0.46 MPa for Ti and -0.45 MPa for SS (Figure 61, 62).

Peak von Mises value in the buccal roots of measured as 0.53 MPa for Ti and 0.52 MPa for SS (Figure 63, 64).

In case of intrusion stress patterns observed in the cervical region of maxillary first molar rather than the roots. Both mini implants generated almost same stress patterns on the maxillary first molar with minimal superiority to the Ti.

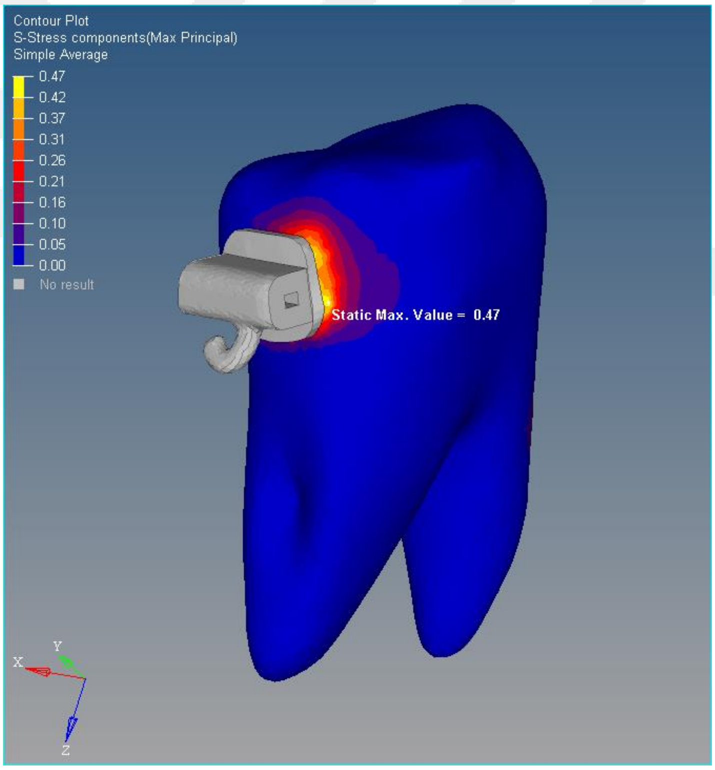

Figure 59. P Max stress patterns generated in the maxillary first molar using Dentos Abso anchor Ti mini implant

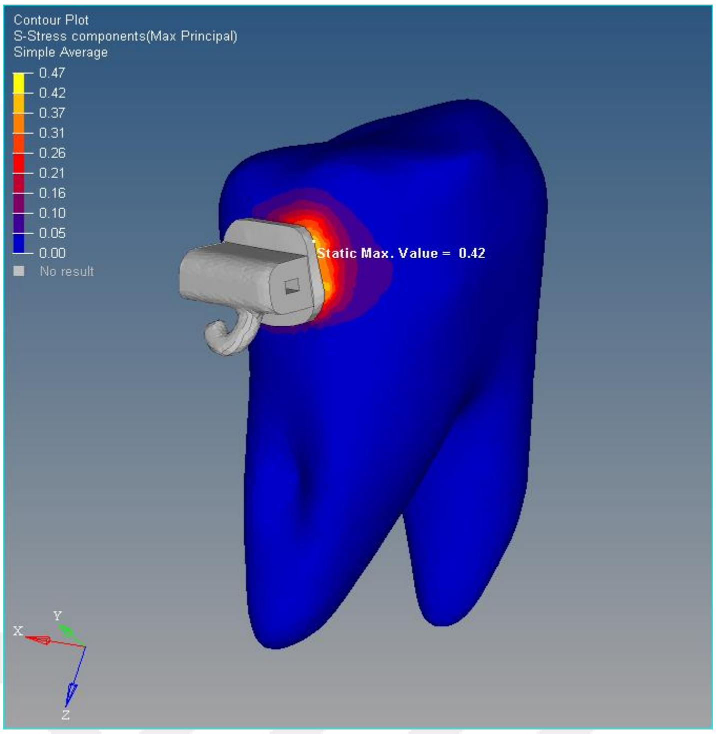

Figure 60. P Max stress patterns generated in the maxillary first molar using Ortho Bone SS mini implant

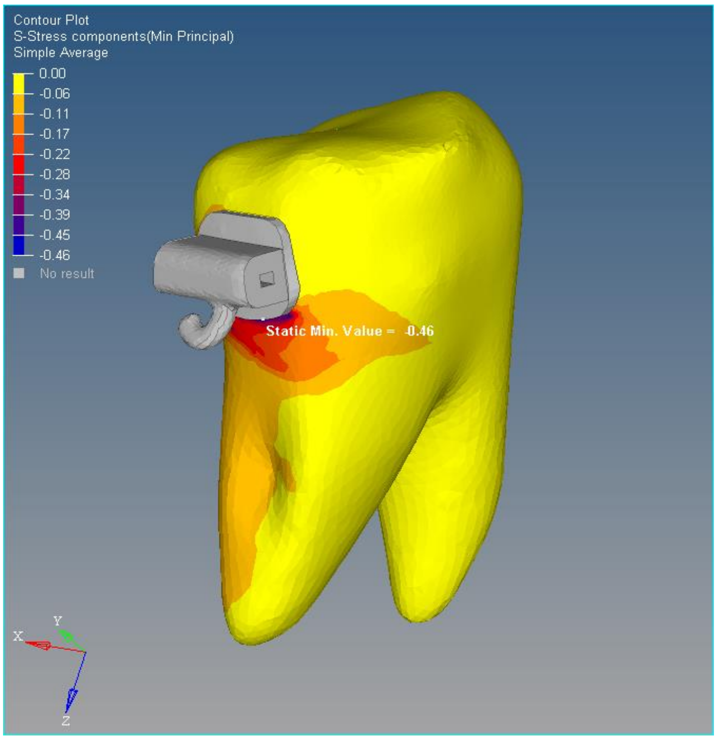

Figure 61. P Min stress patterns generated in the maxillary first molar using Dentos Abso anchor Ti mini implant

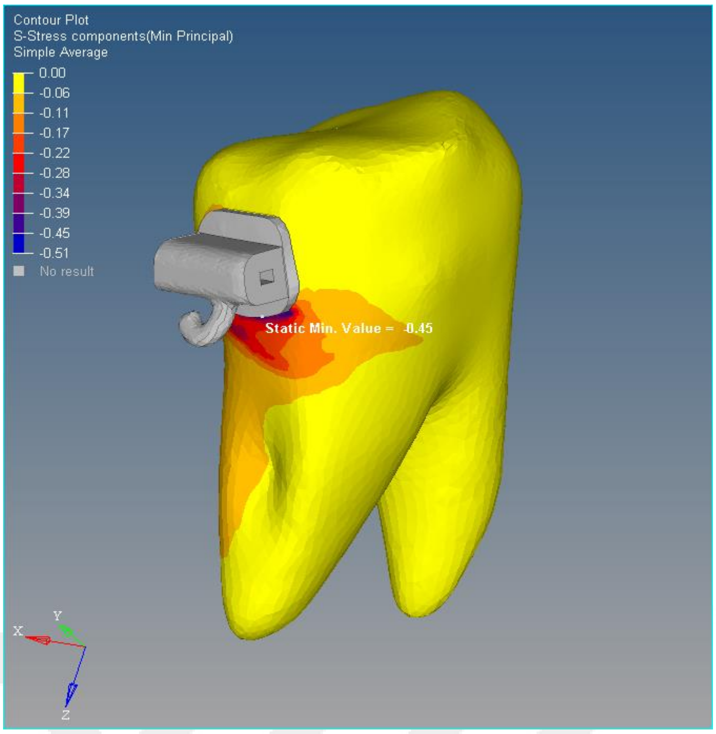

Figure 62. P Min stress patterns generated in the maxillary first molar using Ortho Bone SS mini implant

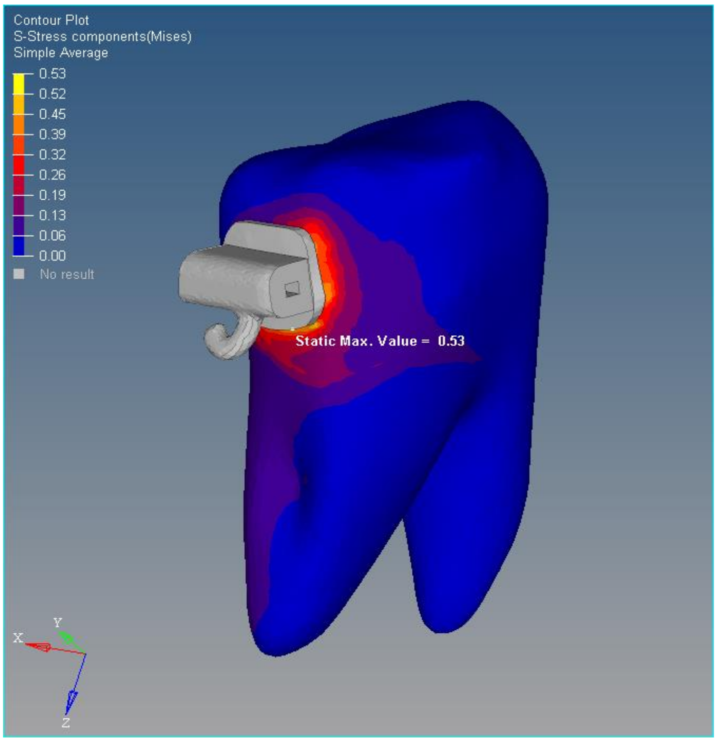

Figure 63. Von Mises stress patterns generated in the maxillary first molar using Dentos Abso anchor Ti mini implant

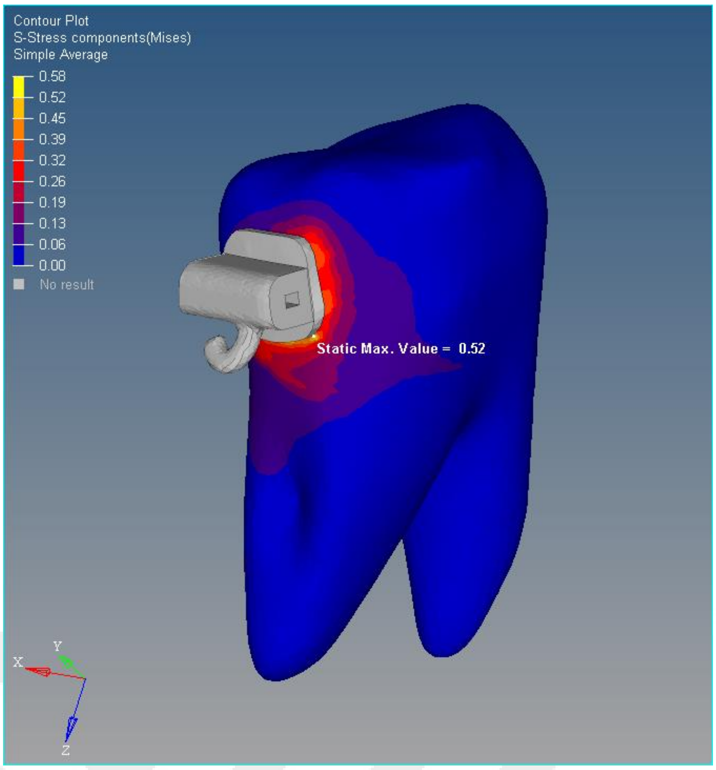

Figure 64. Von Mises stress patterns generated in the maxillary first molar using Ortho Bone SS mini implant

5. DISCUSSION

One of the points of interest in this study is the construction of a model contained maxilla and zygomatic bone, which was included sufficient details to obtain results those, may be valuable and usefull clinically. As there is lack of published orthodontic studies regarding IZC using FEA, outcomes of our study will be compared with the results of those studies which used FEA in cortical bone. Scenarios performed in this study included distalization and intrusion of molars with two different forces, then the generated stresses were separately evaluated, this point provided better comparision between stress patterns generated by Ti and SS mini implants.

Molar intrusion and distalization which represented scenarios of our study, considered as the main orthodontic movements obtained by IZC mini implant. In Class II malocclusion cases in which premolar extraction is an excluded option, usually the treatment procedures will be through the maxillary molars and dentition distalization, the mandibular dentition mesialization or a combination of both. As the mesialization of mandibular dentition has many disadvantages (127), so maxillary dentition distalization has become the treatment of choice in cases of maxillary anterior protrusion and Class II malocclusion (128, 129). Many appliances have been used clinically for distalizing molars and whole dentition distalization. The most traditional appliance was the headgear which has been frequently used for many years as molar distalizer (130). As the headgear has many disadvantage relating the esthetic and patient's compliance so this reason motivate many invetigators to develop intraoral devices with different techniques to distalize the molars, these intraoral devices which eliminate the need of high cooperation from patients such as Wilson arch, repelling magnet, hilgers pendulum appliance, sectional jig assembly, the distal jet, the Keles slider and the first class appliance (103, 131-136). Then recently many researches have been conducted regarding using of the osseointegrated implants, intraosseous screws and plates for the purpose of anchorage in orthodontic patients who need distalization (14, 115, 137). The TADs specifically mini implants become an effective treatment tools to solve Class II malocclusions by maxillary dentition distalization without creating adverse reciprocal movement in cases of no extraction (138). The zygomatic process of the maxilla is considered as a suitable area for skeletal anchorage (89). Now aday, the systems of zygomatic anchorage are vastly used for distalizing the upper molars. Suguwara et al. and Kaya et al. in case of molar distalization they used the zygomatic anchorage (139, 140). Lin developed a modernistic technique for the sake of distalizing the whole maxillary dentition by mini implants inserted in the IZC area using a sliding mechanism, these mini implants known as IZC mini implants (116).

Temporary skeletal anchorage devices, specially the zygomatic anchorage devices including mini implants and mini plates have also been used in treating open bite cases, their mechanism of action depend on intruding the molars into their bony pillar which results in producing mandibular autorotation then eventually bite closure (9, 126, 141). Recently It has been publicized that the TADs can provide treatment results which are similar to those which are gained by orthognathic surgery (142). So, molar intrusion using these TADs has recently become an agreeable and favourable way for treating adult cases with skeletal anterior open bite.

The finite element analysis has been used to determine the compressive and tensile stress areas generated by the mini implants, thus evaluating the mini implants stability. Diameter of the mini implant, bone quality, site preparation and also angulation are important factors which could impact the stability of the mini implant directly or indirectly (55, 143, 144). The angle and position of insertion of the mini implant can determine the bone thickness at the IZC area, the more the thicker bone the greater mini implant biting depth, which provide more favorable bony contact and better primary stability of the mini implant (52). The average IZC thickness superior to the maxillary first molar is about 5 to 9 mm, when the mini implant inserted at 40^0 to 75^0 to the maxillary occlusal plane, with position of 13 to 17 mm above the maxillay occlusal plan (33). The 75^0 postulated insertion angle provides an acute insertion angle between the mini implant and the IZC which in turn make it technically difficult to be obtained and it can cause mini implant slippage and stripping of the bone, also at this insertion angle and position the mini implant could be very close to the mesiobuccal root of the maxillary first molar which may cause root injury. On the other hand the 40^0 postulated mini implant insertion angle and position has a shallower biting depth with a greater possibility of alveolar or buccal mucosal irritation (56). Liou et al. reported that the most appropriate mini implant insertion angle and position in the IZC should range between 55^0 to 70^0 to the maxillary occlusal plane and 14 to 16 mm above the maxillary occlusal plane (33). Based on this fact, in this study the mini implant was inserted at postulated insertion angle of 70^0 to the maxillary occlusal plane, and position of 14 mm above the maxillary occlusal plane which means having biting depth of 8 mm, in respect with the safety zone that is ranging from 0.5 to 2 mm (33, 64).

While there is no definite evidence proving whether the mini implant length is an important factor for primary stability or long-term success (143), implant diameter play a significantly larger role in the stability. The larger diameter of the mini implant permits a more favourable stress distribution (144). From a clinical point of view, mini implant with smaller diameter could be placed and removed easier and be less traumatic, but generally mini implants with a diameter over 1.2 mm are more recommended for orthodontic anchorage. Miyawaki et al. found that all screws with 1.0 mm of diameter have failed, while success rates of screws with 1.5 mm and 2.3 mm diameter were 83.9% and 85%, respectively (55). So, in this study two IZC mini implants of 12 mm in length and 2 mm in diameter have been used.

According to 3D finite element model analysis, a 1.4 mm diameter mini implant that is placed in a cortical bone with 1.2 mm of thickness can withstand 150 g of orthodontic force, while a 1.8 mm diameter implant can withstands 350 g of orthodontic force (145). Park et al. found that 10 to 20 g of force can be used for the intrusion of an incisor and 150 to 200 g of force for intruding a multi-rooted tooth (123). Yao et al. applied 150 to 200 g of force for in intruding an overerupted maxillary first molars from the buccal and palatal aspects at the same time (10). In another clinical study, 150 g intrusive force had also been used for molar intrusion with no significant root resorption in any patient at the end of treatment (146). Over-erupted molar intrusion had also been successfully obtained with 150 g force through the maxillary sinus cortical floor whith preservation of the periodontal health, tooth vitality and root length (147). Cornelis et al. in a prospective clinical study suggested that the application of 150 g force on the maxillary molars is sufficient not only to push the molars backward into an overcorrected Class I position, but also to initiate the retraction of the premolars, canines and incisors with sliding mechanism (148). Wu et al. used 300 g of force per each side for distalizing the maxillay dentition with IZC mini implant (149). Nur et al. used 300 g of force per each side for maxillary molar distalization with zygoma-gear appliance (150). Based on the principles of orthodontic force application and according to the previous studies, we applied 300 g of force on mini implant for molar distalization and 150 g of for molar intrusion in our study.

Since the introduction of mini implant to orthodontic practice, many studies have been conducted for investigating mini implant success rates and factors associated with mini implant failure, there are also many comparative studies for mini implants have been reported regarding different length, diameter, shape, insertion angles and mechanism of insertion (50, 151, 152). Wu et al. evaluated 414 mini implants, including SS and Ti, and reported a success rate of 89.9% (153). Disegi JA. et al. found that the mechanical strength is lower for pure Ti and Ti-alloy mini implants than for SS mini implants (154). Elias et al. in a comparative study between Ti, Ti-alloy and SS mini implants, found that Ti-alloy have higher corrosion resistance and SS mini implants have better penetration (155). In 2012 Pan et al. in his study, he measured the resonance frequencies of mini implants which had inserted into artificial bone blocks using a resonance frequency detection device for this purpose, primary stability has then been compared among mini implants of different sizes, compositions and insertion depths, he found that with respect to the stability, insertion depth was more important than mini implants composition (156). Tsai et al. in a prospective study found that the mini implants success rate and survival did not significantly differ in relation to implant material, which means that the mini implants material did not affect their success rate, regardless of healing time (157). Tseng et al. in a study on the artificial bone discoveed that the material, shape and weight of mini implants has nothing to do with the mechanical strengths (158). Albrektsson and Hansson used light and electron microscopy for the purpose of studying the metal/bone interface between unthreaded Ti and SS implants, they determined direct integration with Ti, while the SS implants had a connective tissue layer surrounding them (61). An in-vivo study on rabbit tibia showed that Ti screws enhanced bone contact and had higher removal torques than SS (159). Brown et al suggested that the SS and Ti alloy miniscrew implants provide the same mechanical stability and similar histologic responses, suggesting that both are suitable for immediate orthodontic clinical loads (24). All mentioned comparative studies which have been performed in vivo and in vitro evaluated the mechanical strength, biocompatibility, stability, and effects of different mini implants materials on healing process, but in FEA field there is an obvious lack of these kind of comparative studies between SS and Ti mini implants regarding the stress created by each of them on the bone and peripheral vital structures. More specifically, there is no any study conducted in FEA to compare stress patterns created by mini implants in the infrazygomatic crest area. For this reason, we performed this study to compare between the stress patterns that are generated by SS and Ti infrazygomatic crest mini implants after force application, and to evaluate the stress/strain effects on the bony structures of the infrazygomatic region including cortical and cancellous bone in adition on the roots of maxillary first molar.

The primary area of focus in the present study is the effect of using different mini implant materials by comparing stress levels generated between Ti and SS mini implants. While there are limited studies regarding comparing the mechanical properties of Ti and SS mini implants, Carano et al. reported that both mini implants could safely be used as skeletal anchorage (23), there are rare studies available which compare Ti and SS stresses or stress pattern generation, so it is difficult to compare our study results directly to the another studies. Regarding to the findings related to the cortical bone in the present study, in case of Ti mini implant peak P Max value was approximately as twice as that in the SS mini implant in both scenarios, while the Peak of P Min was found to be approximately three times greater for Ti when compared to SS in both scenarios, at the same time Ti mini implant has showed a significantly greater value of peak Von Misses than that created by SS in cortical bone. That means that Ti mini implant created greater tensile and compressive stresses on the cortical bone after application of both 150 and 300 g forces when compared with SS mini implant. Since Ti has lower modulus that SS, results of this study matches other studies which considered that materials with lower modulus distributes and transmits forces better to the bone because the bone and implant will flex in a more similar fashion (160).

As the density of cancellous bone plays only a minor role in resisting the applied force (89), Liu et al. in a FEA study found that cortical bone with a higher Young's modulus sustains higher loads than does cancellous bone (161). Also suzuki et al. found that the maximum stress of cancellous bone was significantly lower than that for cortical bone in a FEA study (162). Results of our study were similar to the findings of the previous studies as the P Max, P Min and Von Mises of both SS and Ti mini implants in cortical bone were significantly greater than that in cancelous bone in both scenarios.

Although the maximum stress in cancelous bone for all models of our study were low in value, the peak stress created by Ti mini implant were significantly higher than those created by SS mini implant in both scenarios.

The modulus of elasticity is an important physical property of materials which indicates the flexibility or rigidity of a component before permanent deformation occurs, a material with a low modulus of elasticity possesses the advantage of reduced stress shielding, because more stress will be transferred to the bone (163). Compared to SS, the Ti alloys are roughly equal in strength, but it has half modulus, so Ti devices have the advantage over SS as they have high bioactivity and more flexibility that improve integration and mechanical fixation (160). In consistent with previous studies, in our study the stress created on the surface of Ti mini implant was greater than that observed on the surface of SS mini implant. Peak Von Mises value measured as 100 Mpa for Ti and 19.0 Mpa for SS in case of distalization, while in case of intrusion were 40.4 Mpa for Ti and 7.9 Mpa for SS.

Several studies reported that maximum stresses were located around the neck of mini implant when forces applied perpendicularly to the long axis of mini implant (164). Our results were in good agreement with the results of previous studies as the areas of maximum stress concentration were located at the neck of both Ti and SS mini implants. While the position and angulation of the mini implant placement in our study are considered to be in the safe zone regarding to the surrounding anatomical structures including roots of maxillary first molar, so minimal stress patterns were observed on the buccal roots while stress was not observed in the palatal root in both scenarios. Even though stress patterns observed in the buccal roots were low in value in case of distalization, SS mini implant showed higher stress generation when compared to Ti. In case of intrusion, stress patterns observed in the cervical region of maxillary first molar rather than the roots. Both mini implants generated almost same stress patterns on the maxillary first molar with minimal superiority to the Ti.

6. CONCLUSIONS

1. As the peak values of stress generated did not reach the ultimate tensile strength of cortical bone in any model, titanium and stainless steel can be considered as a successful and safe materials for mini implants used as infrazygomatic anchorage.
2. The peak value of stress generated in the cortical bone of infrazygomatic region is higher in titanium mini implant when compared to the stainless steel, which is related to the higher elasticity modules of stainless steel.
3. Both mini implants created low value of stress in the cancellous bone of infrazygomatic bone with superiority to Ti over SS.
4. The peak value of stress generated on titanium mini implant is obviously greater than that observed on the stainless steel mini implant, which is due to the high elacticity of titanium.
5. Regardless of its composition, infrazygomatic crest mini implants doesn’t show any stress on maxillary molar roots when inserted in a proper angulation and position under certain amount of orthodontic forces.

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