“COMPARATIVE EVALUATION OF BIOMECHANICAL PERFORMANCE OF TITANIUM AND STAINLESS STEEL MINI IMPLANTS AT DIFFERENT ANGULATIONS IN THE MAXILLA: A FINITE ELEMENT ANALYSIS”

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List of Abbreviations

Sr. No.	Abbreviations	Full Form
1.	3D	3 Dimensional
2.	Ti	Titanium
3.	TAD	Temporary Anchorage device
4.	N	Newton
5.	CEJ	Cement Enamel Junction
6.	FEA	Finite Element Analysis
7.	PDL	Periodontal ligament
8.	FEM	Finite Element Method
9.	CT	Computed Tomography
10.	CBCT	Cone Beam Computed Tomography
11.	MPa	Mega Pascal
12.	RMO	Rocky Mountain Orthodontics
13.	P	Probability of occurring of an event
14.	Mm	Millimeter
15.	gm/gms	Grams
16.	.stl	Standard triangulation language
17.	SD	Standard Deviation

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Introduction

Miniscrews have become a reliable and accepted means for providing temporary anchorage during orthodontic and orthopaedic treatment. Mini implants have several advantages over the conventional methods of skeletal anchorage. Miniscrews are easier to place, have versatility in varied locations, are both cost effective and smaller, and have the possibility of immediate or early loading. However, the clinical behaviour of miniscrews is still unclear. Various authors have reported loosening and failure during clinical orthodontic treatment. Also, various factors such as implant type and dimensios, design and shape of the screw thread, implant surface characteristics, amount of orthodontic force, soft tissue characteristics, cortical bone thickness and insertion have been reported to influence stability of miniscrews. Miniscrews have been placed in virtually every bony location in the mouth; however, the angle of insertion reported in the literature varies from 30 to 90 degrees relative to the alveolar bone.[1]

The proper angle of insertion is important for cortical anchorage, patient safety, and biomechanical control. The use of a proper insertion angle reduces the risk of damaging the dental roots of adjacent teeth and also provides increased surface contact area between the microimplant and the bone. But the actual impact of different insertion angulations on microimplant stability is unknown. It is virtually impossible to measure stress accurately around microimplants 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. It appears to be suitable for simulating complex mechanical stress situations in the maxillofacial region.[2]

One aspect of miniscrew use that lacks significant evidence is the evaluation of the miniscrew placement angulation and the subsequent stress generation in both miniscrew and bone when placement angulation is varied. Some researchers have advocated placement angulation that avoids tooth roots or increases cortical bone contact. Others advocate for angulation that facilitates hygiene or ease of placement. The stresses generated around the miniscrew are a result of the structure and properties of the materials/tissue involved. Although bone, teeth, and other anatomical structures are predetermined in each patient, the material or design used in a miniscrew is potentially alterable. Typically titanium (Ti) has been the material of choice; however, alternative materials such as stainless steel (SS) can also be used and may prove useful if they can help reduce the stress generated both within the miniscrew and within the cortical and medullary bone.[3]

Many studies have been done in the past considering the angulation and the material of mini implants , but evidence lacks in comparative evaluation of titanium and stainless steel mini implants together in a finite element model which is anatomically accurate and with 5 different implant angulations.

Aim and Objectives

AIM:

To assess the tensions and deformations (stresses and strains) generated after the application of two types of forces (traction and torsion) in miniscrews of two different materials (Titanium & stainless steel).

OBJECTIVES:

- To assess the tensions and deformation (stresses and strains) generated after the application of two types of forces (torsion & traction) in Titanium miniscrew placed at different angulation.
- To assess the tensions and deformation (stresses and strains) generated after the application of two types of forces (torsion & traction) in stainless steel miniscrew placed at different angulation.

- To compare the tensions & deformation (stresses & strains) after application of torsion & traction in titanium & stainless steel miniscrews placed at different angulation.

Review of literature

The literature has been received under following headings

1. History of Temporary Anchorage Devices (TAD’s)
2. Success rate of TAD’s
3. Risk factors for mini implant failure.\
4. Studies on implant angulation
5. Introduction to finite element methods.
6. History of finite element method.

1.HISTORY OF TEMPORARY ANCHORAGE DEVICES

Conservation of anchorage is one of the most important factors in rendering orthodontic treatment. As DeVan’s dictum says, “we must maintain what is good rather than replacing what is impaired”. Various methods have been employed in orthodontics to preserve anchorage. Orthodontic mini implants are the latest additions in the armamentarium, by use of which, absolute anchorage can be achieved. The concept of skeletal anchorage was introduced in the year 1945 by **Gainsforth** and **Higley**.[4]They placed 85 mini implants in the dog ramus and observed them 1 month post placement. All these implants failed subsequently.

In 1970. **Leonard Linkow**[5] used an implant for replacement of missing molar as well as anchor for tooth movement. He described in this case report, the use of endosseous blade implants to anchor the rubber bands that were used to retract maxillary anterior teeth.

Wigglesworth[6](1977) in a patient report, “the orthodontic movement of a metal implant” mentioned the use of removable orthodontic device to tip another blade implant in the bone.

Sherman[7](1978) conducted an investigation involving a limited series of vitreous carbon implants in dogs using a standard orthodontic technique with force of 175gms. Tetracycline was used to label new bone formation which has been found to be deposited in vivo in sites of bone formation. The deposition was studied in undecalcified sections by fluorescence microscopy. Two of the six vitreous carbon implants placed resisted the force applied whereas, other four implants loosened or fractured. On the basis of very small sample in dogs, it appeared that orthodontic techniques could not be used to move a vitreous carbon implant.. Therefore implants should not be used before the completion of orthodontic treatment in persons who may require orthodontic service unless the implants are to be used for anchorage.

Smith[8] (1979) studied bone dynamics associated with controlled loading of bioglass – coated aluminium oxide endosteal implants in three male rhesus monkeys. 45S5 bioglass coated Al_2O_3 was utilized which had been proven to produce bone bond. Implants were placed at sites of extracted/enucleated mandibular first molars in two stages with healing period of nine weeks and subjected to orthodontic force of 425 gms by means of a coil spring. The implants remained stationary against the force for the periods up to 6 weeks with minimum changes around the implant posts.

Using vitreous carbon implants in baboons, **Oliver**[9] *et al* (1980) found that all implants adequately resisted forces ranging from 30 to 200gms.

Turley[10] *et al* (1982) and **Turley and Roth**[11] (1983) reported that subperiosteal implants in dogs were stable against 300gms of orthodontic force and 1300 gms of orthopaedic force.

Gray[12] *et al* (1983) studied the abilities of two types of small cylindrical endosseous implants to resist movement when loaded with constant forces of orthodontic magnitude . Pairs of bioglass & Vitallium coated implants of the same size were implanted into the femurs of twelve rabbits. After a 28-day healing period, these implants were loaded with forces of 60 gms, 120gms and 180 gms. Analysis of implant movement after 28-days revealed that no 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.

Creekmore[13] (1983) did a case study in 25-year old female patient to determine if a metal implant could withstand a constant force over a long period of time of adequate magnitude to depress an entire anterior maxillary dentition without becoming loose, infected, painful or pathologic. A Vitallium bone screw was inserted just below anterior nasal spine. Ten days after the screw was placed, a light elastic thread was tied from head of the screw to archwire. During the treatment, the maxillary central incisors were elevated by approximately 6mm and torqued lingually about 25^0. The screw did not move during treatment and was not mobile at the time of removal.

Smalley[14] *et al* (1988) studied the potential of osseointegrated titanium implants for maxillofacial protraction in four pigtail monkeys. Titanium implants were placed surgically into the maxillary, zygomatic, frontal, and occipital bones. The application of force varied among the animals. In animal A, the force was applied to the maxilla. In animal B, the force was applied to the zygomatic bones, in animal C in frontal and in animal D at occipital bone. A tensile force of 600gms/side was maintained until approximately 8mm of maxillary anterior displacement had occurred. The amount of movement required 12 weeks of force application in animals A and B, and 18 weeks in animals C and D. Cephalometric and dry skull analyses showed that the amount of skeletal protraction was significant.

Turley[15] *et al* (1988) did a pilot study on six mongrel dogs, to investigate the use of titanium endosseous implants as orthodontic and orthopaedic anchors using radiographic, histologic and vital bone marking techniques. The use of titanium endosseous implants as anchors for orthodontic tooth movement along with their placement and stability in non-traditional areas was studied. A total of 42 implants were placed, out of which, 24 remained stable and 8 of these implants received activation. All loaded implants remained stable throughout the period of force activation. The large two-stage titanium endosseous implants showed 100% success compared to 47% for smaller one stage implants. Titanium implants resisted 300gms of orthodontic force and 1000gms of orthopaedic force.

In 1989, **Roberts**[16] did a study of rigid endosseous implants for orthodontic and orthopaedic anchorage. Titanium implant (13 x 3.75 mm) were placed in edentulous premolar areas of four beagle dogs. Experimental side was loaded in compression by 3N closed coil stainless steel springs. At two intervals in 20 weeks, all bone forming sites were labelled with different fluorochrome. The analysis of results obtained by various methodologies led to conclusion that all implants remained rigid. Titanium implants with as little as 10% bone contact resisted 3N loads over a period of 20 weeks.

Linder –Aronson[17] *et al* (1990) placed titanium endosseous implants into two monkeys and determined that they successfully resisted 60gms of force over 8 weeks.

Higuchi and Slack[18](1991) studied the use of implants to facilitate orthodontic tooth movement. Using the osseointegration method, a prospective study was conducted involving seven adult patients who were treated with titanium implants used as rigid anchorage units. Orthodontic forces were directed off the implants to correct a variety of malocclusions all 14 implants placed remained stable during the course of treatment with loading forces of 150 to 400gms. No significant complications occurred.

Wehrbein and **Diedrich**[19] (1993) in an experimental study of endosseous titanium implants during and after orthodontic load, placed eight implants in two dogs to determine whether endosseous titanium implants (Brånemark) retain their clinical stability throughout a 26-week period of continuously applied force (2N) and what kind of marginal peri implant bone changes occur in the process. The implants were used for distalization of second premolar over a 26-week period. There were no clinical or histological signs of implant dislocation after the load application period. In the presence of mild peri-implant gingivitis, no increase was found in the incidence of marginal bone resorption adjacent to the loaded implants compared with the non-loaded implants. In the absence of marginal resorption, subperiosteal bone apposition was detected, especially around the test implants in the mandible. The study suggested that endosseous titanium implants were suitable as anchoring units for long term orthodontic tooth movements and applied force may induce marginal bone appositions adjacent to the implants.

Wehrbein[20] *et al* (1997) experimentally studied the effect of long term orthodontic loading on the stability as well as on the peri-implant bone findings of short titanium screw implants inserted in regions with reduced vertical bone height. For this purpose, 6 maxillary premolars ($_1P_1$, $_2P_2$, $_3P_3$) were extracted from each of 2 foxhounds and reduction of alveolar bone height was performed by osteotomy. After a l6-week healing period, 8 implants (4 per dog) were inserted in the edentulous areas. Simultaneously, 2 implants (1 per dog) were positioned in the palatal suture. After an

5-week implant healing period, the fixtures in the Pl/P2 areas and the palate were loaded (test implants) by means of transpalatal bars running anteriorly, fixed on the implants in the Pl/P2 areas, and Sentalloy traction springs (-2 N continuous force) inserted midsagittally between palatal implants and bars (force application period: 26 weeks). The fixtures in the P2/P3 areas served as controls. Clinical and histological evaluation revealed no implant dislocation of the loaded fixtures. The results suggested that short titanium screw implants inserted in the alveolar bone and palatal suture region retain their stability during long-term orthodontic loading, even following a relatively short unloaded implant healing period. Furthermore, long-term orthodontic loading may induce marginal bone apposition adjacent to the implants.

In 1997, **Kanomi**[21] was the first to report the use of mini implants for orthodontic purpose. He used the mini implants for intrusion of mandibular anterior and buccal teeth.

In 1998, **Costa**[22] *et al* published a preliminary report on using mini screws for anchorage in orthodontics. They discussed application of the mini implants as anchorage for various types of tooth movement and possible locations for placement of these implants. These applications were based on studies conducted on dry skulls. They stated that mini screws can be easily placed and removed and can be loaded immediately following insertion. They also stated that, stability is limited when loaded with torsion.

Akin-Nergiz[23] (1998) studied Functional and morphologic reactions of peri-implant bone surrounding screw implants in three dogs by loading the implants with continuous forces of 2N and 5 N. Eight implants were inserted to an endosseous length of 12 mm and placed about 10 mm apart in the region of the lower premolars. The fixtures healed in a closed environment for 12 weeks, after which they were uncovered and loaded with abutments and orthodontic devices to produce horizontal distraction with a force of 2 N for 12 weeks. Subsequently they were loaded with 5 N for another 24 weeks. The distance between and the mobility of the implants were determined before and after each phase of experimental loading. Fixtures of the same type that were osseointegrated and not exposed, or osseointegrated and loaded by mastication, were used as a control. Animals were euthanized and specimens sectioned. The result was, continuously loaded implants showed no significant displacement for any force level. The mobility of the fixtures increased slightly by about 1 Periotest-value at the end of the experiment. No significant peri-implant pocket was seen in implants loaded by continuous or masticatory forces. Histologic and morphometric evaluation indicated compaction of bone as a result of loading. Osseointegrated implants have potential as a firm osseous anchorage for orthodontic treatment and can resist continuous horizontal forces of at least 5 N (about 510gm) during a period of several months.

Wehrbein[24] *et al* (1999) studied the mid-sagittal area of palate as an alternative insertion site for placement of implants for orthodontic anchorage. The results suggest that vertical bone support is at least 2 mm higher than apparent on the cephalogram. In none of 12 patients was a perforation to the nasal cavity found. However, in five subjects the implant projected into the nasal cavity on the post-operative cephalogram. These results were supported by the study of the projections of palate and wires in wire-marked skulls where the wires were placed bilaterally on the nasal floor and on the nasal crest. It is therefore concluded that the mid-sagittal area of the palate lends sufficient bony support for the implantation of small implants (4-6 mm endosseous length, diameter 3.3 mm).

Celenza and **Hochman**[25] in their review article published in the year 2000, have stated that orthodontic mini implant can be either used as direct anchorage or as indirect anchorage.In direct anchorage, the force is directly applied on the implant whereas in indirect anchorage, the implant is connected to a tooth or segment of teeth to stabilize them via an archwire or ligature wire, thereby incorporating them into the anchorage unit.

Park[26] *et al* (2001) introduced the use of titanium micro screws and mini screws for orthodontic anchorage. Microscrews are small enough to be placed in any area of the alveolar bone, easy to implant and remove, and inexpensive. In addition, orthodontic force application can begin immediately after implantation. Treatment of skeletal class I bialveolar protrusion , with micro implant anchorage was successful and they were used for retracting the maxillary anterior teeth and uprighting the mandibular molars

Janssens[27] *et al* (2002) reported the use of an onplant for palatal anchorage to extrude the unerupted horizontal maxillary first molars in a 12-year-old white girl with tooth aplasia and secondary cleft palate. After a healing period of 5 months, the onplant remained stable under indirect elastic tension of approximately 160 g applied for 17 weeks, and the maxillary first molars were successfully extruded.

Deguchi[28] *et al* (2003) studied 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 in 8 dogs and 100% of the elastomeric chain-loaded implants. All of the loaded implants remained integrated. 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 labelling 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 a minimal (under 3 weeks) healing period.

Park *et al*[29] (2004) stated that skeletal anchorage using dental implants, mini plates miniscrews and microscrews provides an absolute anchorage for tooth movement. Miniscrew or microscrew implants have many benefits such as ease of placement and removal,because of their small size, they can be placed in the intra arch alveolar bone without discernable damage to roots. Orthodontic loading can begin immediately after placement, in contrast to dental implants. Characteristics of sliding mechanics with mini implant anchorage are independent from patient compliance, an early improvement of the facial profile, and shortened treatment time by retracting six anterior teeth simultaneously. In sliding mechanics with miniscrew anchorage, chairside can be reduced by the decreased number of archwire changes. By using microscrew implants, clinicians can correct class II and class III dental relationships efficiently in a shorter treatment time. Microscew implanrs might make changes in treatment planning. For instance, patients who have mild class III skeletal problems who resist orthognathic surgery. can be treated with microscrew implant aided mechanics.

Kim *et* al[30] (2006) carried out a study to obtain sufficient stability of implants, the thickness of the soft tissue and the cortical bone in the placement site. To measure soft-tissue and cortical-bone thicknesses, maxillae from 23 Korean cadavers were decalcified, and buccopalatal cross-sectional specimens were obtained. These specimens were made at 3 maxillary midpalatal suture areas: the interdental area between the first and second premolars (group 1), the interdental area between the second premolar and the first molar (group 2), and the interdental area between the first and second molars (group 3). In all groups, buccal soft tissues were thickest closest to and farthest from the cementoenamel junction (CEJ) and thinnest in the middle. Palatal soft-tissue thickness increased gradually from the CEJ toward the apical region in all groups. Buccal cortical-bone was thickest closest to and farthest from the CEJ and thinnest in the middle in groups 1 and 2. Palatal cortical-bone thickness was greatest 6 mm apical to the CEJ in groups 1 and 3, and 2 mm apical to the CEJ in group 2. Along the midpalatal suture, palatal mucosa remained uniformly 1 mm thick posterior to the incisive papilla. Their conclusion was, surgical placement of miniscrew implants for orthodontic anchorage in the maxillary molar region requires consideration of the placement site and angle based on anatomical characteristics.

Papadopoulos and Tarawneh[31] (2007) presented and discussed the development, clinical use, benefits, and drawbacks of the miniscrew implants used to obtain a temporary but absolute/skeletal anchorage for orthodontic applications. Topics discussed included classification, types and properties (e.g., biocompatibility, osseointegration, types of anchorage, screw head, and thread design), clinical applications, site and placement method selection, clinical procedures for implant insertion, and loading and removal processes. Also, the potential complications and the advantages and disadvantages were presented.

Elias Nelson[32] (2012) *et al* have stated that, orthodontic implants have become a reliable method in orthodontic practice for providing temporary additional anchorage. These devices are useful to control skeletal anchorage in less compliant patients or in cases where absolute anchorage is necessary. There are a great number of advantages in this new approach which include easy insertion, decreased patient discomfort, low price, immediate loading, reduced diameter, versatility in the forces to be used, ease of cleaning, and ease of removal. However, a proper management of the screws by the practitioner is necessary in order to increase the success rate of the technique. The purpose of this paper is to update practitioners on the current concepts of orthodontic implants and orthodontic mechanics.

Mute, Peter, Daokar[33] (2013) conservation of anchorage in totality has been perennial problem to traditional orthodontist. Traditionally, orthodontists have used teeth, intraoral appliances, and extraoral appliances, to control anchorage minimizing the movement of certain teeth, while completing the desired movement of other teeth. However, according to Newton's third law of motion, for every action there is an equal and opposite reaction. Thus, there are limitations in our ability to completely control all aspects of tooth movement. For example, we often have inadequate mechanical systems to control anchorage, which leads to anchorage loss of reactive units and often incomplete correction of intra and interarch alignment problems. For long time, orthodontist have struggled to achieve efficient control of anchorage and always dreamt of a device, which could provide absolute anchorage .This dream came true with the advent of implant. Implant has burst onto the clinical orthodontic scene in order to assist the orthodontist in controlling tooth movement. Orthodontic implants or temporary anchorage device (TAD) are temporarily fixed to bone for the purpose of enhancing orthodontic anchorage either by supporting the teeth of the reactive unit or by obviating the need for the reactive unit all together, and which is subsequently removed after use.

Vibhute and Shenoy[34] (2013) have stated that, miniscrews have become the regular components as an anchorage source in orthodontics. Stability and failure of miniscrew is materializing to be multifactorial with no consensus on causative factors. Factors that influence the load transfer at the bone-implant interface and miniscrew stability include host factors, biomechanical factors, sterilization protocol and hygiene. Biomechanical influences on bone structure play an important role in the longevity of bone. Incorrect loading or overloading as a result of ineffective implant geometries may lead to implant loss. Miniscrews are not an exception for following the mechanical behavior or engineering principles. More certainly, they are based on the simple machines, which is a mechanical device that changes the direction or magnitude of a force. Their article produced an insight for rationalization of mechanical factors controlling the stability with principles of simple machines, namely (i) lever, (ii) wheel and axle, (iii) inclined plane, (iv) wedge and (v) screw.

2. SUCCESS RATE OF MINI IMPLANTS

With the increasing use of mini implants, came forward the phenomenon of assessment of success rates of these implants. **Miyawaki**[35] *et al* in 2003, conducted a study to assess success rates and to find the factors associated with the stability of titanium screws placed into the buccal alveolar bone of the posterior region. Fifty-one patients with malocclusions, 134 titanium screws of 3 types, and 17 miniplates were retrospectively examined in relation to clinical characteristics. The 1-year success rate of screws with 1.0-mm diameter was significantly less than that of other screws with 1.5-mm or 2.3-mm diameter or than that of miniplates. They concluded that the success rate was 78.4%. They also enlisted certain factors for implant failure these were, implant diameter of less than 1 mm, inflammation of of the peri implant tissue, high mandibular plane angle.

Tseng[36] *et al* (2006) studied the use of mini-implants for skeletal anchorage, and to assess their stability and the causes of failure. Forty-five mini-implants were used in orthodontic treatment. The diameter of the implants was 2 mm, and their lengths were 8, 10, 12 and 14 mm. The drill procedure was directly through the cortical bone without any incision or flap operation. Two weeks later, a force of 100–200 g was applied by an elastometric chain or NiTi coil spring. Four mini-implants loosened after orthodontic force loading. The overall success rate was 91.1%. The location of the implant was the significant factor related to failure.

Motoyoshi[37] *et al* in 2006 studied to determine an adequate placement torque for obtaining a better success rate of mini-implants. Implant placement torque was measured. 124 implants were placed in 41patients, with an average age of 24.9 years. The peak value of placement torque was measured using a torque screwdriver. The success rate of the mini-implant anchor for 124 implants was 85.5%. The mean implant placement torque ranged from 7.2 to 13.5 N cm, depending on the location of the implants. There was a significant difference in the implant placement torque between maxilla and mandible. The implant placement torque in the mandible was, unexpectedly, significantly higher in the failure group than in the success group. Therefore, a large torque should not be used always. According to their calculations of the risk ratio for failure, to raise the success rate of 1.6-mm diameter mini-implants, the recommended implant placement torque is within the range from 5 to 10 N cm.

Park[38] *et al* (2006) conducted a study to examine the success rates and find factors affecting the clinical success of screw implants used as orthodontic anchorage. Eighty-seven consecutive patients with a total of 227 screw implants of 4 types were examined. Success rates during a 15-month period of force application were determined according to 18 clinical variables. The overall success rate was 91.6%. The clinical variables of screw-implant factors (type, diameter, and length), local onset and method of force application, ligature wire extension, exposure of screw head, and oral hygiene) did not show any statistical differences in success rates. Mobility, jaw (maxilla or mandible), and side of placement (right or left), and inflammation showed significant differences in success rates. They also stated that, to minimize the failure of screw implants, inflammation around the implant must be controlled.

In 2007, **Kuroda**[39] *et al* evaluated the clinical usefulness of miniscrews as orthodontic anchorage. They examined success rates, factors associated with their stability, and patient's postoperative pain and discomfort with a questionnaire. Seventy-five patients, 116 titanium screws of 2 types, and 38 miniplates were retrospectively examined. The success rate for each type of implant was greater than 80%. The analysis of 79 miniscrews with a 1.3-mm diameter showed no significant correlations between success rate and these variables: age, sex, mandibular plane angle, anteroposterior jaw-base relationship, control of periodontitis, temporomandibular disorder symptoms, loading, and screw length. Most patients receiving titanium screws or miniplates with mucoperiosteal-flap surgery reported pain, but half of the patients receiving miniscrews without flap surgery did not report feeling pain at any time after placement. They concluded that miniscrews placed without flap surgery have high success rates with less pain and discomfort after surgery than miniscrews placed with flap surgery or miniplates placed with either procedure.

Garfinkle JS[40] *et al* (2008) conducted a study to determine the success rate, positional stability, and patient evaluation of orthodontic mini-implants. Thirteen patients were treated with 82 mini implants measuring 1.6 mm in diameter and 6 mm in length placed in the buccal alveoli (1 unloaded mini implant and 1 loaded mini implant per quadrant). The right or left side of each arch was randomly selected for immediate loading with up to 250 g of direct force; the contralateral side was loaded 3 to 5 weeks later. Serial impressions, clinical observations, and orthodontic maintenance were performed until adequate space closure was achieved. The success rate of mini implants was 70.3%. Neither the timing of force application nor the force itself precipitated failure of the implants. Orthodontic forces can be applied immediately to mini implants.

3. RISK FACTORS FOR MINI IMPLANT FAILURE

Chen YJ[41] *et al* (2008) evaluated the potential factors that influence failure rates of temporary anchorage devices (TAD's) used for orthodontic anchorage using a retrospective study. Data on 492 TADs (miniplates, pre-drilling miniscrews, and self-drilling miniscrews) in 194 patients were collected. There were no significant differences in failure rates among the TADs for the following variables: gender, type of malocclusion, facial divergency, implantation site (buccal, lingual, or crestal / midpalatal), location (anterior or posterior), method of force application (power chain or Ni-Ti coil spring), arch (upper or lower), type of soft tissue (attached gingiva or removable mucosa), and most of the cephalometric measurements that reflect dento-cranio-facial characteristics. An increased failure rate was noted for the self-drilling TAD, TADs used for tooth uprighting, those inserted on bone with lower density, those associated with local inflammation of the surrounding soft tissue, those loaded within 3 weeks after insertion, and those placed in patients with greater mandibular retrusion. They concluded that Inflammation of soft tissue surrounding a TAD and early loading within 3 weeks after insertion were the most significant factors predicting TAD failure.

In 2004, **Cheng SJ**[42] *et al* in their prospective clinical study, assessed the risk factors associated with failure of mini-implants used for orthodontic anchorage. A total of 140 mini-implants in 44 patients, including 48 miniplates and 92 freestanding miniscrews, were examined in the study. A variety of orthodontic loads were applied. The majority of implants were placed in the posterior maxilla (104/140), and the next most common location was the posterior mandible (34/140). A cumulative survival rate of 89% (125/140) was found. There was no significant difference in the survival rate between miniplates and freestanding miniscrews, but miniplates were used in more hazardous situations. The estimated relative risk of implant failure in the posterior mandible was 1.101%. The risk ratio of failure for implants surrounded by non-keratinized mucosa was 1.117%.

Mah J and **Bergstrand F**[43] (2005) in their status report on temporary anchorage devices, have stated that mini implants with smaller diameter (<1mm) are more prone to failure.

Ohashi E[44] *et al* in their systematic review presented the loading protocols applied when using implants in orthodontic treatments. Electronic databases were searched with the help of a senior Health Sciences librarian. Abstracts which appeared to fulfil the selection criteria were selected by consensus. The original articles were then retrieved and evaluated with a methodological checklist. Based on their results, they concluded that loading protocols for implants involve a minimum waiting period of 2 months before applying orthodontic forces while loading protocols for screws involve immediate loading or a waiting period of 2 weeks to apply forces. Success rates for implants were on average higher than for screws.

Kuroda[45] *et al* (2007) retrospectively evaluated root proximity as a risk factor for the failure of miniscrews used as orthodontic anchorage. They used dental radiographs and 3-dimensional computed tomography images to examine 216 titanium screws in 110 patients. Each screw was classified according to its proximity to the adjacent root. Category I, the screw was separate from the root; category II, the apex of the screw touched the lamina dura; and category III, the body of the screw overlaid on the lamina dura. If the orthodontic force could be applied to the screw for 1 year (or until completion of orthodontic treatment), they recorded the screw anchorage as a success.

Their results showed a success rate--above 80%. Screws placed in the maxilla had higher success rate than those in the mandible. There was a significant correlation between success rate and root proximity. There were significant differences in the success rates between categories I and II, I and III, and II and III. Although screws in all 3 categories in the maxilla and categories I and II in the mandible showed high success rates above 75%, screws in category III in the mandible had a low success rate of 35%.

They concluded that the proximity of a miniscrew to the root is a major risk factor for the failure of screw anchorage. This tendency is more obvious in the mandible.

In 2007, **Kravitz** and **Kusnoto**[46] in their review article have stated that over insertion is also an important factor in implant failure.

4. STUDIES ON IMPLANT ANGULATION

Pollei JK stated that the most important factor which must be considered while placing the implant is that, it has a potential to generate stresses in the bone and altering the angulation of mini implant may help in reduction of such stresses. Measurement of stress generated at the implant bone interface or inside the implant and bone individually is not both possible and feasible. To measure such parameters, Finite Element Analysis (FEA) is a powerful tool where, we can simulate the surroundings of the mini implant when it is placed in the prescribed location , and accurate analysis of the said parameters can be studied, and that too on a computer based design and analysis software. Finite element analysis is a reliable tool and it is being used in the engineering faculty frequently.[3]

Chen[47] *et al* (1995) created three finite element models to study the potential of rigid osseous fixation (osseointegration) for orthodontic anchorage.; a mandible without an implant; a mandible with an implant; and a mandible and implant with a superimposed orthodontic load. Force was applied to different locations and the stresses were computed.

It was observed that mechanical stress distribution adjacent to the implant was not affected by different biting forces, thus only one case was analysed. The stress adjacent to the bone-implant interface changed drastically due to implantation, with major changes occurring on the buccal and mesiobuccal sides.

A strong gradient for intraosseous stress and bone remodelling rate was observed that reflected a mismatch in the moduli of elasticity, between the implant and the supporting bone. Findings of the study had important clinical implications. The study concluded that it was unlikely for a rigidly fixed (osseointegrated) implant to lose integration due to an orthodontic load superimposed in normal function.

Jones[48] (2001) *et al* carried out a study, the aim which was to develop a 3D computer model of the movement of a maxillary incisor tooth when subjected to an orthodontic load. A novel method was to be developed to directly and accurately measure orthodontic tooth movement in a group of human volunteers. This was to be used to validate the finite element-based computer model.

The design took the form of a prospective experiment at a laboratory at the University of Wales in 1996/7. A laser apparatus was used to sample tooth movement every 0.01 seconds over a 1-minute cycle for 10 healthy volunteers, whilst a constant 0.39 N load was applied. This process was repeated on eight separate occasions and the most consistent five readings taken for each subject.

Data was used to calculate the physical properties of the periodontal ligament (PDL An appropriate elastic modulus of 1 N/mm(2) and Poisson's Ratio of 0.45 was derived for the PDL. Strain analysis, using the model, suggested that a maximum PDL strain of 4.77 x 10(-3) was recorded at the alveolar crest, while the largest apical strain recorded was 1.55 x 10(-3). The maximum strains recorded in the surrounding alveolar bone were 35 times less than for the PDL.

A novel method for direct measurement of PDL physical properties in the human subject has been developed. The validated FEM model lends further evidence that the PDL is the main mediator of orthodontic tooth movement. Displacements ranged from 0.012 to 0.133 mm. An appropriate elastic modulus of 1 N/mm(2) and Poisson's Ratio of 0.45 was derived for the PDL.

Kyung HM[49] (2003), **Carano** *et al* [50] (2005) and **Melsen**[51] (2005) stated that most frequently used location for the placement of mini implant is the posterior buccal aspect of maxilla, although various other sites also serve this purpose. Various authors have given various angulations of implant insertion in this region, ranging from 30^0 to 90^0, relative to the long axis of the alveolus.

In 2006, **Byoun**[52] *et al* studied the stress distribution on the diameter of the mini-implant and insertion angle to the bone surface. To perform three dimensional finite element analysis, a hexadron of 15 x 15 x 20 mm^3 was used, with a 1.0 mm width of cortical bone. Mini-implants of 8 mm length and 1.2 mm, 1.6 mm, and 2.0 mm in diameter were inserted at 90^0, 75^0, 60^0, 45^0, and 30^0 to the bone surface. 200gms of horizontal force was applied to the centre of the mini-implant head and stress distribution and its magnitude were analyzed by ANSYS, a three dimensional finite element analysis program.

The findings of this study showed that maximum von Mises stresses in the mini-implant and cortical and cancellous bone were decreased as the diameter increased from 1.2 mm to 2.0 mm with no relation to the insertion angle. Analysis of the stress distribution in the cortical and cancellous bone showed that the stress was absorbed mostly in the cortical bone, and little was transmitted to the cancellous bone.

The contact area increased according to the increased diameter and decreased insertion angle to the bone surface, but maximum von Mises stress in cortical bone was more significantly related with the contact point of the mini-implant into the cortical bone surface than the insertion angle to the bone surface.

Kravitz, Kusnoto[46] (2007)and **Liou**[53] (2007)*et al* had postulated that the insertion angle of 30^0 to 70^0 from the occlusal plane (i.e. 20^0 to 60^0 to the cortical bone long axis) is optimal and it will allow sufficient bone engagement. This will increase the potential for maximal anchorage, while preventing miniscrew slippage along the surface of the bone during insertion, but theoretically.

Huang[54] *et al* (2007) studied stress analysis for various implant designs using 3-dimensional finite element analysis approaches. Six implant designs were included: 3 parallel-sided implants (no thread, triangular thread, and squared thread), 2 stepped configurations (non-thread and triangular thread), and a tapered body of implant with squared thread. All threads had spiral characteristics. The mandibular model was constructed from computed tomographic (CT) images of a human mandible, and the material properties were anisotropic. A 100-N oblique force was applied at a 45^0 angle to the long axis of the implants at the buccal cusp as the loading condition.

Results showed that, when compared with cylindric implants, threaded implants demonstrated increased peak stress at the crestal bone. The bone stress of stepped implants was decreased in the cortical region but was increased in the trabecular region. Both threaded and stepped designs showed decreased interfacial stresses of bone near the valleys of the threaded and stepped areas. The tapered design decreased stresses by up to 32% in the cortical region and 17% in the trabecular region. They concluded that although threaded implants could not decrease the peak stress at the crestal bone, both threaded and stepped designs show an ability to dissipate the interfacial stresses of bone. The use of tapered implants could reduce peak stress in both cortical and trabecular bone.

Suzuki[55] *et al* conducted a study, where they used finite element analysis to investigate changes in stress distribution at the supporting bone and miniscrew by changing the angle and the shape of the miniscrew and the direction of force.

Three types of miniscrews (cylindrical pin, helical thread, and nonhelical thread) were designed and placed in 2 types of supporting bone (cancellous and cortical). The miniscrews were inclined at 30°, 40°, 45°, 50°, 60°, 70°, 80°, and 90° to the surface of the supporting bone. A force of 2N was applied in 3 directions.

Their results showed that, Significantly lower maximum stress were observed in the cancellous bone compared with the cortical bone. By changing the implantation angle, the ranges of the maximum stress distribution at the supporting bone were 9.46 to 14.8 MPa in the pin type, and 17.8 to 75.2 MPa in the helical thread type. On the other hand, the ranges of the maximum stress distribution at the titanium element were 26.8 to 92.8 MPa in the pin type, and 121 to 382 MPa in the helical thread type. According to the migration length of the threads in the non-helical type, the maximum stresses were 19.9 to 113 MPa at the bone, and 151 to 313 MPa at the titanium element. By changing the angle of rotation in the helical thread type, the maximum stress distributions were 25.4 to 125 MPa at the bone, and 149 to 426 MPa at the titanium element. Furthermore, the maximum stress varied at each angle according to the direction of the applied load.

They concluded that, the maximum stresses observed in all analyzed types and shapes of miniscrews were under the yield stress of pure titanium and cortical bone. This indicates that the miniscrews in this study have enough strength to resist most orthodontic loads.

Woodall[56] (2011) *et al* studied to test the hypothesis that screw angulation affects screw-anchorage resistance. Three-dimensional finite element models were created to represent screw-placement orientations of 30°, 60°, and 90°, while the screw was displaced to 0.6 mm at a distance of 2.0 mm from the bone surface. In a parallel cadaver study, 96 titanium alloy screws were placed into 24 hemi-sected maxillary and 24 hemi-sected mandibular specimens between the first and second premolars.

The specimens were randomly and evenly divided into 3 groups according to screw angulation (relative to the bone surface): 90° vs 30° screw pairs, 90° vs 60° screw pairs, and 30° vs 60° screw pairs. All screws were subjected to increasing forces parallel to the occlusal plane, pulling mesially until the miniscrews were displaced by 0.6 mm.

Their results showed that 90° screw placement provided greater anchorage resistance than 60° and 30° placements. In the cadaver study, although the maximum anchorage resistance provided by screws placed at 90° to the cadaver bone surface exceeded, on average, the anchorage resistance of the screws placed at 60°, which likewise exceeded the anchorage resista nce of screws placed at 30°, these differences were not statistically significant. They concluded that placing orthodontic miniscrews at angles less than 90° to the alveolar process bone surface does not offer force anchorage resistance advantages.

Jasmine[2] *et al* in 2012 studied stress in bone and microimplants during en-masse retraction of maxillary and mandibular anterior teeth with different insertion angulations using the finite element method. Finite element models of a maxilla and a mandible with types D3 and D2 bone quality, and of microimplants with a diameter of 1.3 mm and lengths of 8 and 7 mm were generated.

The microimplants were inserted at 30°, 45°, 60°, and 90° to the bone surface. A simulated horizontal orthodontic force of 200 g was applied to the centre of the microimplant head, and stress distribution and its magnitude were analysed with a 3D finite element analysis program.

Their results showed that the maximum von Mises stresses in the microimplant and the cortical bone decreased as the insertion angle increased. Analysis of the stress distribution in the cortical and cancellous bones showed that the stress was absorbed mostly in the cortical bone, and little was transmitted to the cancellous bone. The maximum von Mises stress was higher in type D3 bone quality than type D2 bone quality. They also concluded that placement of microimplants at a 90° angulation in the bone reduces the stress concentration, thereby increasing the likelihood of implant stabilization. Perpendicular insertion offers more stability to orthodontic loading.

In 2013, **Lin**[57] *et al* studied biomechanical effects of exposure length of the mini-implant, the insertion angle, and the direction of orthodontic force. Twenty-seven finite element models were constructed to simulate the biomechanical response of the alveolar bone adjacent to the mini-implant. Factorial analysis was performed to investigate the comparative influence of each factor.

The results showed that the exposure length of the mini-implant had a statistically significant influence on bone stress, with a contribution of 82.35%. Increased exposure length resulted in higher bone stress adjacent to the mini-implant. Whereas all factors investigated had a statistically significant influence on cancellous bone stress, the stress values associated with cancellous bone were much less than those of cortical bone.

They concluded that increased exposure lengths resulted in higher bone stresses adjacent to the mini-implant. The percentage of contribution of the insertion angle of the mini-implant (6.03%) was also statistically significant but much less than that of the exposure length (82.35%). The direction of orthodontic force had no significant effect on cortical bone stress.

In 2014, **Perillo**[1] *et al* studied the influence of placement angle and direction of force on the stability of miniscrews. Finite element analysis was performed using miniscrews inserted into 1mm of cortical bone and 10mm of trabecular bone at angles of 30, 60, 90, 120, and 150 degrees to the alveolar bone. Force of 2N was applied to the heads of the miniscrews in two directions of 0 and 30 degrees.

The finite element analysis showed that inserting miniscrews at 90 degree angle would provide better anchorage than 30, 60, 120, and 150 degree angles at either direction of force. The least trabecular bone von Mises stress was 5.6MPa at 90^0 at both directions of force and the least cortical bone stress was 31.2MPa at 90^0 both directions of force.

They concluded that, insertion of miniscrews at angles less than or greater than 90^0 to the alveolar process bone might decrease the anchorage stability of the miniscrew.

Later in 2015, **Marimuthu**[58] *et al* studied the biomechanical effects of implant insertion angle and direction of orthodontic force on maxilla and mandible by finite element approach and factorial analysis. A three-dimensional finite element bone block models of maxilla and mandible with type D3 and D2 bone quality were constructed. Mini-implants were inserted at 30°, 60°, and 90° and orthodontic force was applied to the center of the mini-implant head at 60°, 90°, and 120° angulation. ANSYS software was used to evaluate the stress on implant, stress on bone and displacement of bone.

Maximum von Mises stress was observed at 30° insertion angle. The stress on implant, stress on bone and displacement of bone increased as the insertion angle decreased from 90° to 30° and was statistically significant in both maxilla and mandible. The direction of orthodontic force had no statistically significant effect on stress and displacement around mini-implant in both maxilla and mandible. The stress on bone and displacement of bone was greater in maxilla compared to that of mandible and was statistically significant.

They concluded that, Placement of mini-implant perpendicular to the long axis of the tooth reduces the stress concentration around the mini-implant and its interface, thereby increasing the likelihood of implant stability. The direction of orthodontic force has no significant effect on implant stability.

5. INTRODUCTION TO FINITE ELEMENT METHOD

Logan[59] in 2011 in his course on finite element methods stated that, **finite element method** (**FEM**) is a numerical method for solving problems of engineering and mathematical physics. It is also referred to as **finite element analysis** (**FEA**). Typical problem areas of interest include structural analysis, heat transfer, fluid flow, mass transport, and electromagnetic potential. The analytical solution of these problems generally require the solution to boundary value problems for partial differential equations. The finite element method formulation of the problem results in a system of algebraic equations. The method yields approximate values of the unknowns at discrete number of points over the domain. To solve the problem, it subdivides a large problem into smaller, simpler parts that are called finite elements. The simple equations that model these finite elements are then assembled into a larger system of equations that models the entire problem. FEM then uses variational methods from the calculus of variations to approximate a solution by minimizing an associated error function.

6. HISTORY OF FINITE ELEMENT METHOD

While it is difficult to quote a date of the invention of the finite element method, the method originated from the need to solve complex elasticity and structural analysis problems in civil and aeronautical engineering. Its development can be traced back to the work by A **Hrennikoff**[60] and **R. Courant**[61] in the early 1940s.

The finite element method obtained its real impetus in the 1960s and 1970s by the developments of **J. H. Argyris** with co-workers at the University of Stuttgart, R. W. Clough with co-workers at UC Berkeley, **O. C. Zienkiewicz** with co-workers Ernest Hinton, Bruce **Irons**[62] and others at the University of Swansea, **Philippe G. Ciarlet** at the University of Paris and **Richard Gallagher** with co-workers at Cornell University. Further impetus was provided in these years by available open source finite element software programs. NASA sponsored the original version of NASTRAN, and UC Berkeley made the finite element program SAP IV[63] widely available. A rigorous mathematical basis to the finite element method was provided in 1973 with the publication by **Strang** and **Fix.**[64] The method has since been generalized for the numerical modeling of physical systems in a wide variety of engineering disciplines, e.g., electromagnetism, heat transfer, and fluid dynamics.[65,66]

Van Staden[67] *et al* in 2006, published a review of the achievements and advancements in dental technology brought about by computer-aided design and the finite element method (FEM) of analysis. The scope of the review covered dental implants, jawbone surrounding the implant and the biomechanical implant and jawbone interaction. They carried out FEM analysis for various parameters such as implant geometry(diameter and length), taperage, material properties of implant, loading of the implant, implant surface structure, material properties of jawbone etc, and concluded that FEA has been used extensively to predict the biomechanical performance of various dental implant designs as well as the effect of clinical factors on the success of implantation. The principal difficulty in simulating the mechanical behaviour of dental implants is the modelling of the living human bone tissue and its response to applied mechanical forces. Research has been conducted on the design philosophy, length, diameter and shape of implants as well as the biomechanical bond formed between the implant and the jawbone. Such work has been mainly directed towards finding the most biocompatible materials with which the dental implants are constructed, in an attempt to reduce the potential risks of clinical failure.

In 2014, **Trivedi**[68] stated that the finite element analysis (FEA) is an upcoming and significant research tool for biomechanical analyses in biological research. It is an ultimate method for modelling complex structures and analyzing their mechanical properties. FEA has been used to study the stress patterns in various implant components and also in the peri-implant bone. It is also useful for studying the biomechanical properties of implants as well as for predicting the success of implants in clinical condition. FEA of simulated traumatic loads can be used to understand the biomechanics of fracture. FEA has various advantages compared with studies on real models. The experiments are repeatable, there are no ethical considerations and the study designs may be modified as per the requirement.

Materials and Methods

1. Modelling of maxilla

A CBCT scan of whole maxilla was obtained using Kodak 9000 C 3D scanner. A total of 587 slices of the maxilla were obtained. These images were then later sent to Osteo 3D^{TM} for converting these images into .stl format. From the sent images, a 3D model of maxilla was created and also, the model was converted into a 3D printable format (.stl)(Standard Triangulation Language) as suggested by **Bardeswaran**[69] . This 3D printable model was converted to Initial Graphic Exchange Specification format. This model was then imported in computer aided design software program(CATIA P3 V5-6 R2015 B26 / 2016; Dassault Systèmes). Model boundaries were set after 3-D models of the posterior right maxilla and corresponding maxillary dentition(second premolar and first molar) were generated. These were established at: the interproximal region between the maxillary right second premolar as the mesial boundary; the distal aspect of the maxillary first molar as the distal boundary; the complete coronal anatomy of second premolar and first molar teeth as the inferior boundary; and all maxillary structures up to 5mm superior to tooth apices as the superior boundary .The cortical, trabecular bone interface was sequentially created by modelling a second surface offset 1.5 mm internal to the external cortical surface in order to define the cortical and trabecular bone boundary. Now, the modelling of teeth (maxillary first molar and second premolar of upper right side) was done. The model of teeth included lamina dura , PDL , dentine and enamel as suggested by **Pollei**[3] .

2. Modelling of implant

The implant used in this study was 7 mm in length and the diameter was 1.3 mm. The surface anatomy of implant was assessed using a contracer machine. The contracer machine accurately measures the anatomy of any symmetrical object. These measurements are accurate to the level of one thousandth part of a millimetre. The machine which was used in this study is manufactured by MAHR Metrology, Germany. Using the report generated by the contracer machine, the modelling of the implant was done in the CATIA software(fig a1).

All the parts were assembled in the software , and the implant was inserted 6mm apical to the interdental crest as this is the most favoured spot for insertion of implant in this region of oral cavity. Five different models were created with the angulation of the implant at 30^0,45^0, 60^0, 75^0, 90^0 individually, with respect to the long axis of the cortical bone in the CATIA software(fig a2-a6).

ANSYS software requires certain mechanical properties of the materials to be tested. In our study, D3 type of bone was simulated as it is the most common type of bone that is found in this area as suggested by **Chugh**[70] *et al*. The mechanical properties of the biological components were taken as given by **Jasmine**[2] *et al* except for stainless steel, which were suggested by **Singh**[76] *et al.*

These models were then analysed in the ANSYS software. A 20 N of torsional force and a 150gms of linear force was applied onto the implant as used by **Arantes** *et al*[71]. In this study, an extensive 393208 nodes and 157138 elements were meshed.

The torsional force was applied considering the implant in its place and not while inserting. The linear force was applied on the centre of implant head to simulate the situation when en-masse retraction of anterior teeth is done using coil spring.

After the application of force, both the torsional and linear maximum stresses generated on the implant and in the bone (cancellous and cortical) at different angles were recorded.

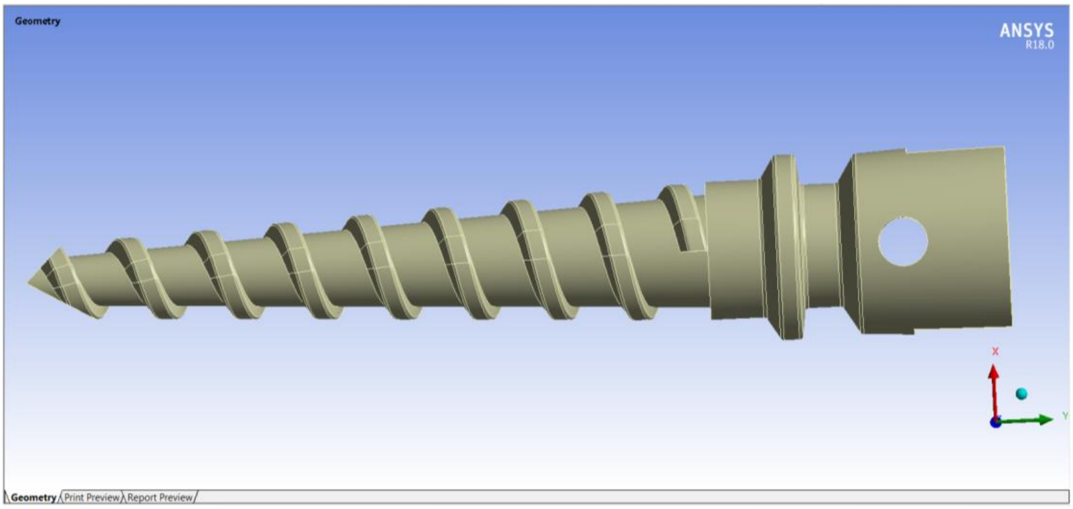
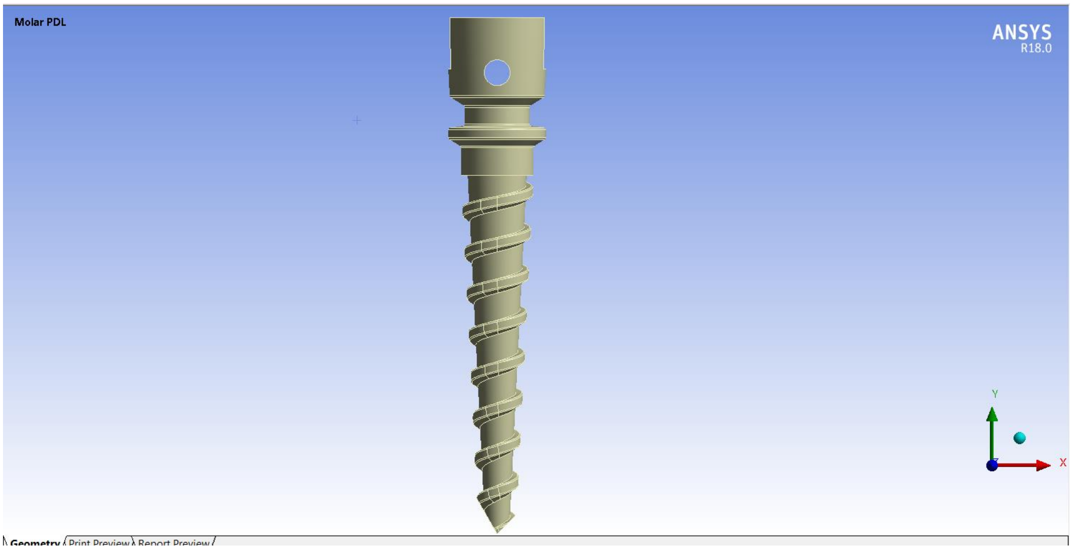


fig a1(i)



Figures a1(ii)

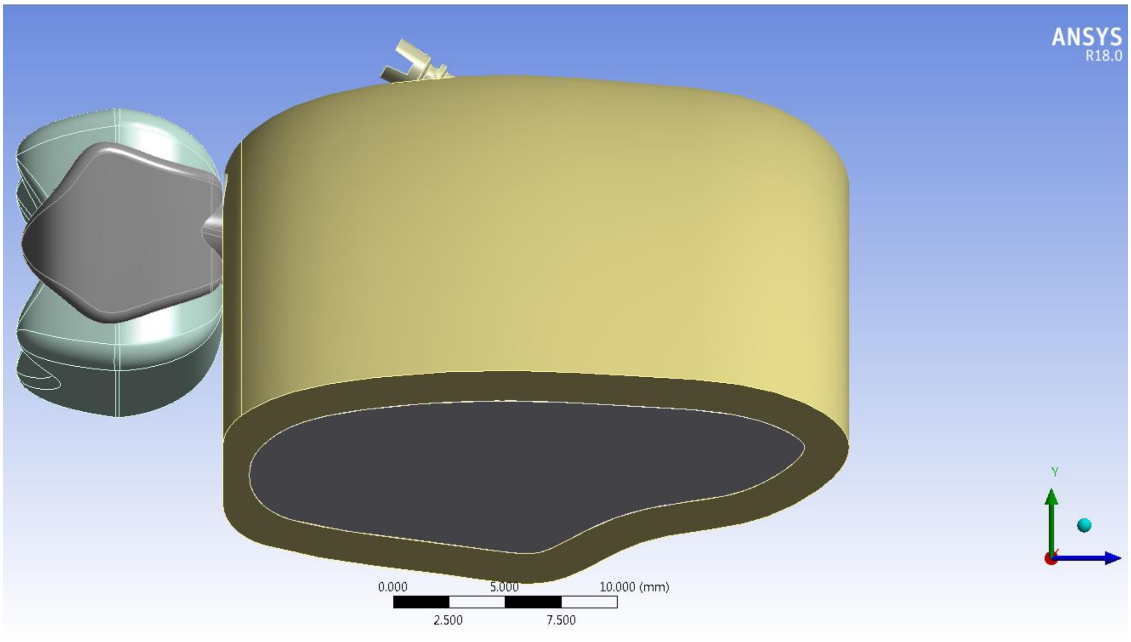


Fig a2. Implant angulation at 30^{0}

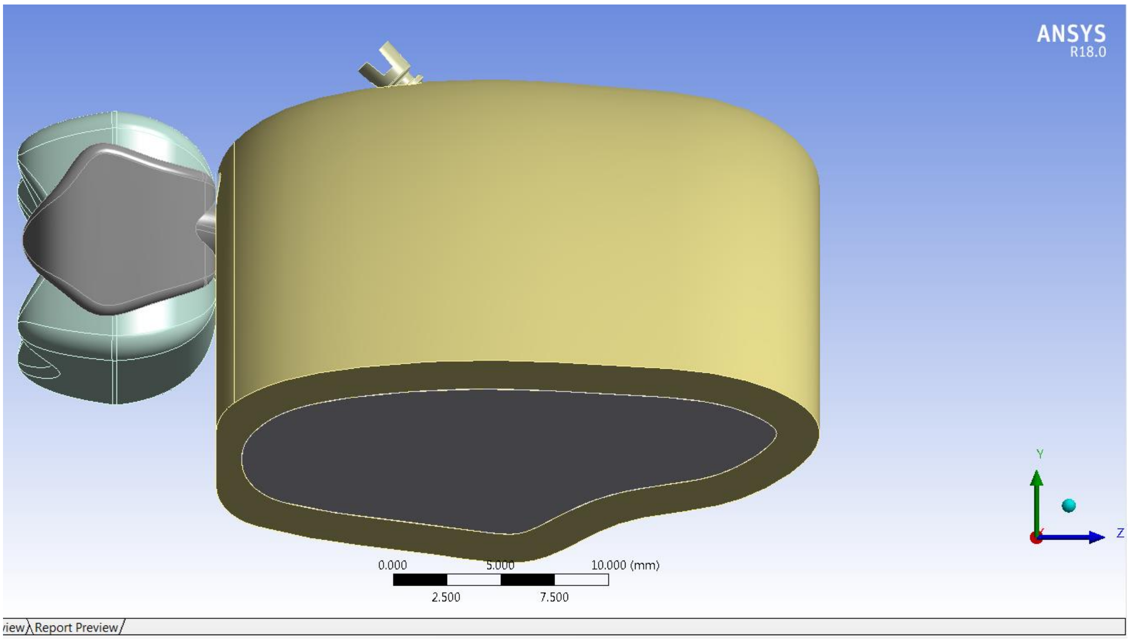


Fig a3. Implant angulation at 45^{0}

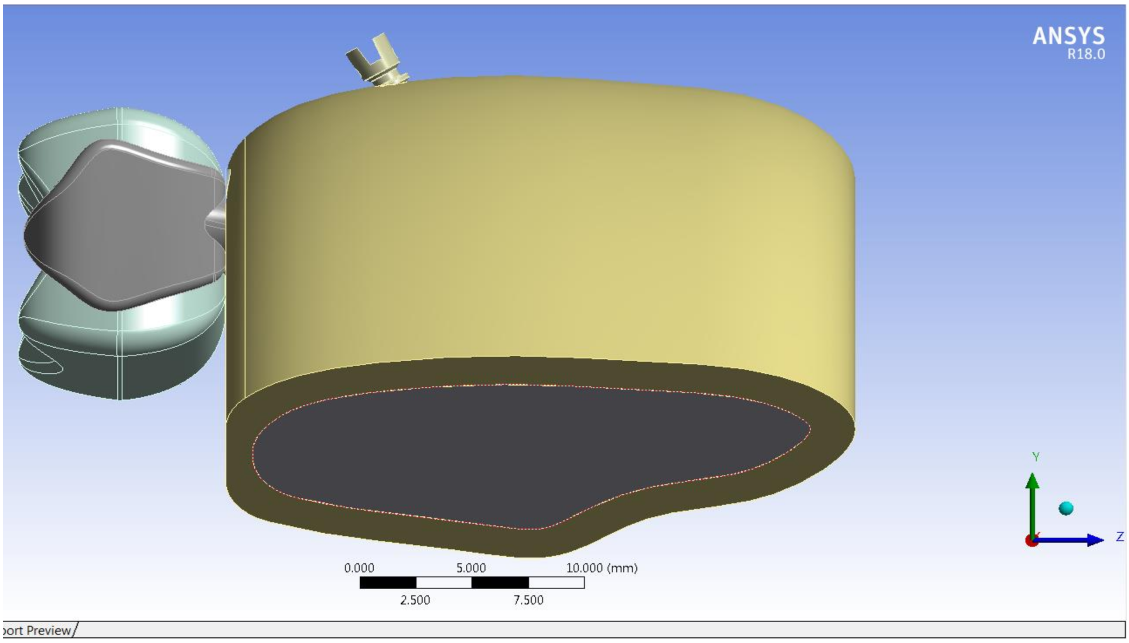


Fig a4. Implant angulation at 60^{0}

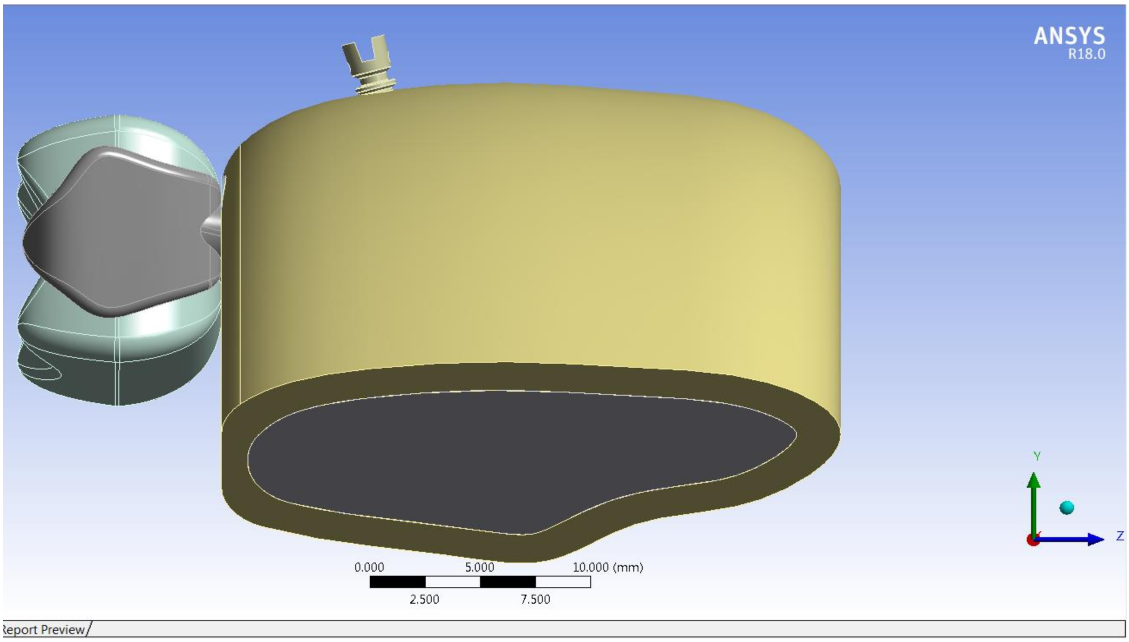


Fig a5. Implant angulation at 75^{0}

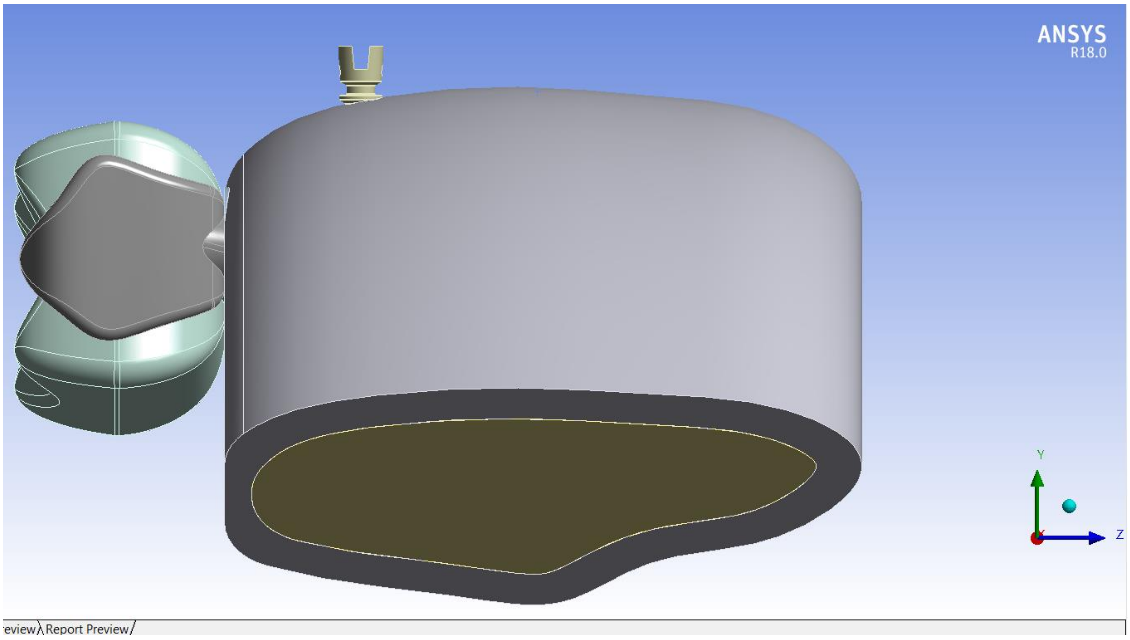


Fig a6. Implant angulation at 90^0

RESULTS

The statistical analysis was done using the Statistical Package for the Social Science (SPSS version 22, Armonk, NY: IBM Corp). Depending upon the nature of data, statistical tests were chosen. Descriptive statistics of mean and standard deviation for the stress generated in implant and bone on application of torsional and linear forces were calculated. Inferential statistics were applied to compare the stress generated between Stainless Steel and Titanium implants using Independent samples t-test.

Tab.1 depicts the stress generated in implants placed at different angulations (30^0,45^0,60^0,75^0 and 90^0) on application of torsional and linear forces. (fig 1,2)

As the angle of implant placement increases from 30^0 to 90^0, the stress generated in both the types of implants increases on application of torsional as well as linear force. The stress generated in stainless steel implant under **torsional force** at 30 degrees was 15.3MPa while in Titanium implant it was 16.009 MPa. At 90 degrees, it increased to

16.249 MPa in Stainless steel implant, whereas in Titanium implant it increased to 16.416 MPa. On application of **linear force** at 30 degrees, the stress generated in Stainless steel implant was 13.781 MPa, while in Titanium implant it was 13.217 MPa. As the angle increased to 90 degrees, the stress generated in stainless steel implant was 18.924 MPa and in Titanium was 19.395 MPa. Thus, it was seen that the stress generated in Stainless steel implant is less as compared to Titanium implant.

Table 2 shows the comparison of stress generated in both the types of implants on application of torsional and linear forces. This difference was not statistically significant. The p-value for comparison of stress under torsional force was p=0.10, while for linear force it was p=0.66.

Table 3 shows the stress generated in bone (cortical and cancellous) loaded with stainless steel and titanium implants on application of torsional force. At 30 degrees implant insertion angle, the stress generated in cortical bone loaded with stainless steel implant was 1.72MPa whereas for Titanium it was 1.97MPa. the stress increased at 45 degrees to 2.12MPa and 2.55MPa in Stainless steel and Titanium implants respectively. It reduced at 60 degrees while it increased at 75 degrees. At 90 degrees, the stress generated was same i.e 1.9MPa in cortical bone loaded with both the type of implants. The stress generated in cancellous bone loaded with stainless steel and titanium implants increased as the implant insertion angle increased but there was no significant difference between the implants at different angles of insertion. (fig 3,4)

Table 4 depicts the stress generated in bone (cortical and cancellous) loaded with stainless steel and titanium implants on application of linear force. At 30 degrees implant insertion angle, the stress generated in cortical bone loaded with stainless steel implant was 1.44MPa while in bone with titanium implant it was 1.62MPa. At 45 degrees angle, it increased to 1.98MPa and 2.32MPa in Stainless steel and titanium respectively. At 60 degrees angle, there was no significant change in the stress generated in cortical bone with both the types of implants. At 75 degrees, it increased. At 90 degrees angle, it increased to 2.26MPa and 2.42MPa in bone loaded with stainless steel and titanium implants respectively. The amount of stress generated in the cancellous bone loaded with stainless steel as well as titanium implant increased as the implant insertion angle increased from 30 degrees to 90 degrees. (fig 5,6)

Table 5 shows the comparison of stress generated in bone loaded with stainless steel and titanium implants on application torsional and linear forces. Under both the forces, the amount of stress generated in cortical bone loaded with stainless steel implant was less as compared to titanium, although this difference was not statistically significant. the amount of stress generated in cancellous bone on application of both the types of forces was almost equivalent in stainless steel and titanium implants.

Table 6 depicts the elastic deformation in both types of implants on application of torsional and linear forces at different angulations of implant placement. The elastic deformation also increased in both, stainless steel and titanium implant as the implant insertion angle increased from 30 degrees to 90 degrees on application of torsional and linear forces. (fig 7,8)

Table 7 shows the comparison of elastic deformation between the stainless steel and titanium implant on application of torsional and linear forces. The difference in amount of elastic deformation between stainless steel and titanium implant was not statistically significant under torsional force as well as under linear force at different angles of implant insertion.

Tab. 1. Stress generated in implant on application of torsional and linear force at different angulations

	Under Torsional force		**Under Linear force**	
Angulations	Type of implant		Type of implant	
	STAINLESS STEEL	**TITANIUM**	**STAINLESS STEEL**	**TITANIUM**
30^0	15.3	16.009	13.781	13.217
45^0	14.551	16.627	15.494	16.954
60^0	15.888	15.88	18.744	18.974
75^0	15.98	15.988	17.736	19.63
90^0	16.249	16.416	18.924	19.395

Tab. 2. Comparison of stress generated between both the types of implants on application of torsional and linear forces.

	STAINLESS STEEL Mean ± SD	**TITANIUM Mean ± SD**	**p-value**
Torsional force	15.6 ± **0.6**	16.2 ± **0.3**	0.10*
Linear force	16.9 ± **2.2**	17.6 ± **2.6**	0.66*

*not statistically significant

Tab. 3. Stress generated in bone on application of Torsional force at various angles

	Cortical bone		**Cancellous bone**	
Angulations	Type of implant		Type of implant	
	STAINLESS STEEL	**TITANIUM**	**STAINLESS STEEL**	**TITANIUM**
30^0	1.7217	1.9792	0.018028	0.018751
45^0	2.1209	2.5571	0.031957	0.031004
60^0	1.4506	2.0399	0.026539	0.020908
75^0	2.5998	3.3395	0.024416	0.014097
90^0	1.9063	1.9651	0.046758	0.026501

Tab.4. **Stress generated in bone on application of Linear force at various angulations**

	Cortical bone		**Cancellous bone**	
Angulations	Type of implant		Type of implant	
	STAINLESS STEEL	**TITANIUM**	**STAINLESS STEEL**	**TITANIUM**
30^0	1.4495	1.6197	0.014771	0.015595
45^0	1.9833	2.3261	0.050846	0.050867
60^0	1.923	2.0454	0.065224	0.070689
75^0	2.3592	2.6874	0.055716	0.058204
90^0	2.2664	2.4215	0.13143	0.11372

Type of force	**Type of bone**	**Type of implant**		**p-value**
		STAINLESS STEEL Mean ± SD	**TITANIUM Mean ± SD**	
Torsional force	Cortical bone	1.95 ± **0.4**	2.36 ± **0.6**	0.24*
	Cancellous bone	0.03 ± **0.01**	0.02 ± **0.006**	0.23*
Linear force	Cortical bone	2.0 ± **0.3**	2.21 ± **0.4**	0.38*
	Cancellous bone	0.06 ± **0.04**	0.06 ± **0.03**	0.94*

Tab. 5. Comparison of stress generated in bone in both types of implants

*not statistically significant

Tab.6. **Elastic deformation in both types of implants on application of torsional and linear forces at different angulations.**

	Torsional force		**Linear force**	
Angulations	Type of implant		Type of implant	
	STAINLESS STEEL	**TITANIUM**	**STAINLESS STEEL**	**TITANIUM**
30^0	0.00020865	0.00018171	0.00022348	0.00020371
45^0	0.0002108	0.00017112	0.00028816	0.00026026
60^0	0.00023329	0.00019564	0.0004295	0.00036897
75^0	0.00023162	0.00020541	0.00065216	0.0004583
90^0	0.00020242	0.00020117	0.00050316	0.0004579

Tab. 7. Comparison of elastic deformation on application of torsional and linear forces.

	STAINLESS STEEL **Mean ± SD**	**TITANIUM** **Mean ± SD**	**p-value**
Torsional force	0.0002 ± **0.00001**	0.00015 ± **0.00003**	0.08
Linear force	0.00042 ± **0.0001**	0.00054 ± **0.0002**	0.365

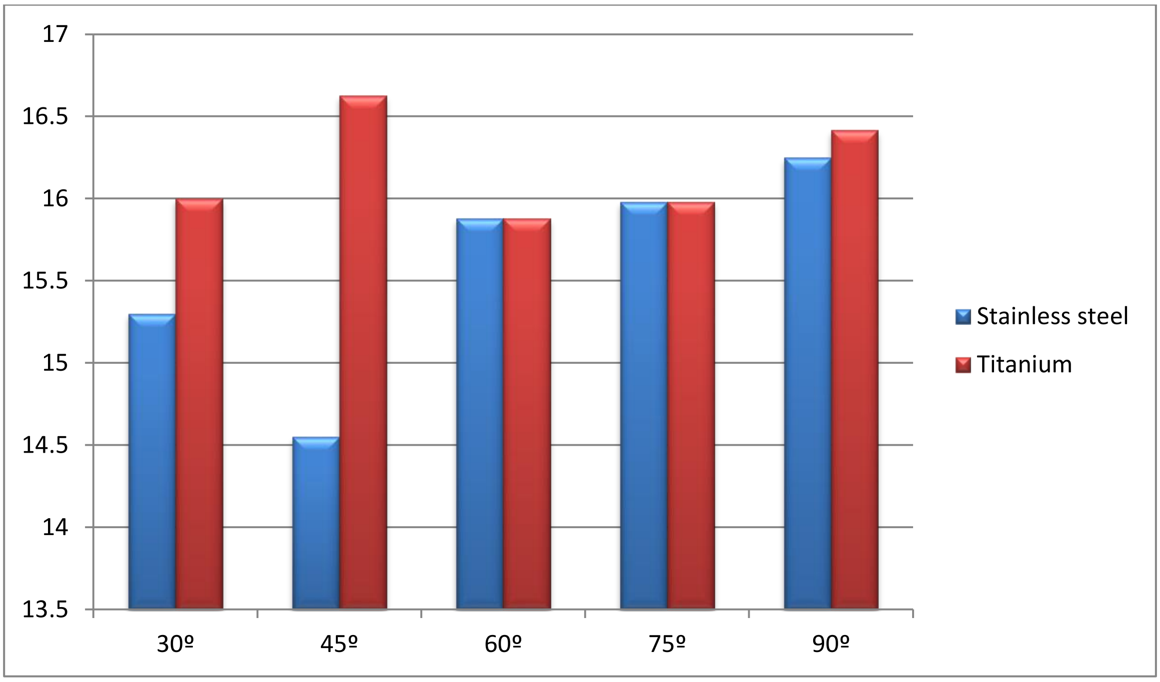


Fig.1. **Stress generated in implant on application of torsional force at different angulations.**

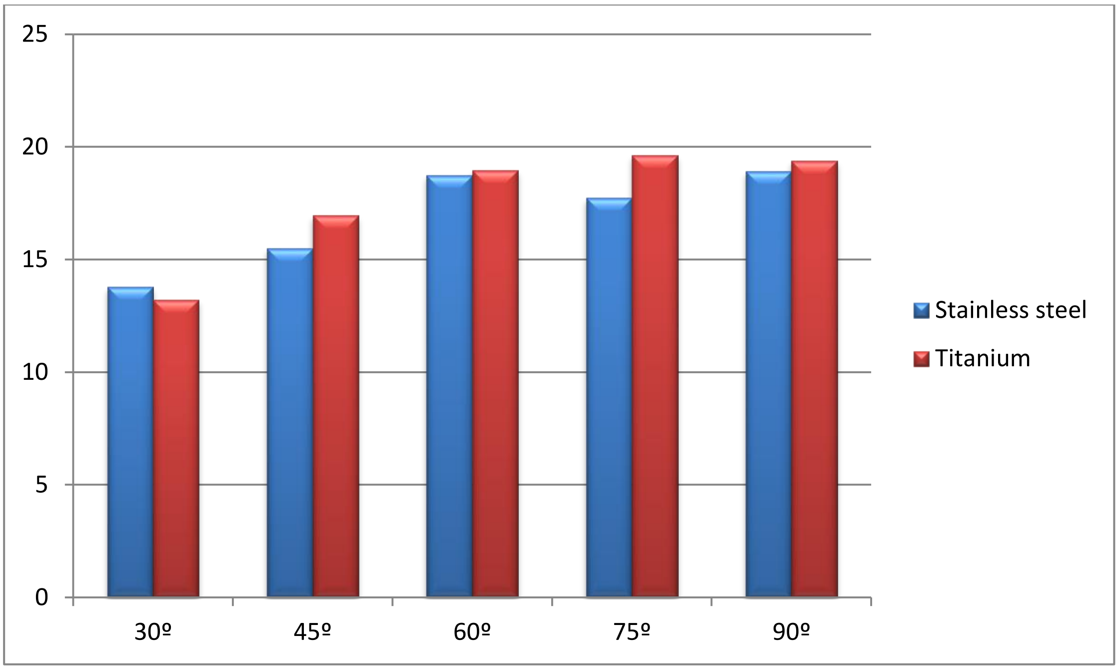


Fig.2. **Stress generated in implant on application of linear force at different angulations**

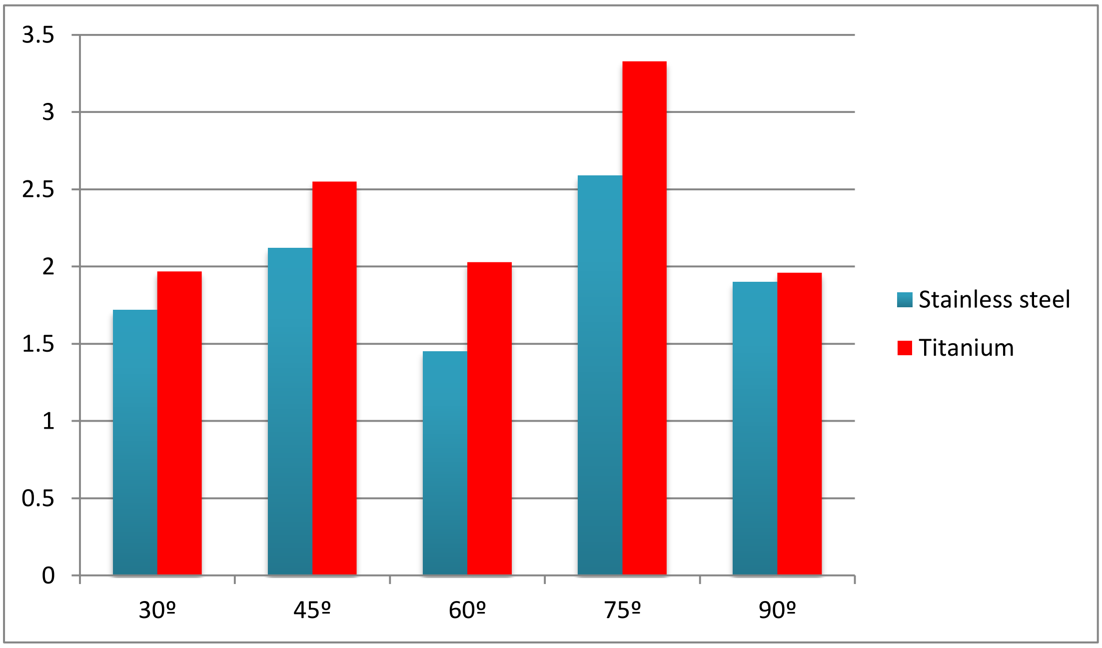


fig. 3 Stress generated in cortical bone on application of Torsional force at various angulations of implant placement.

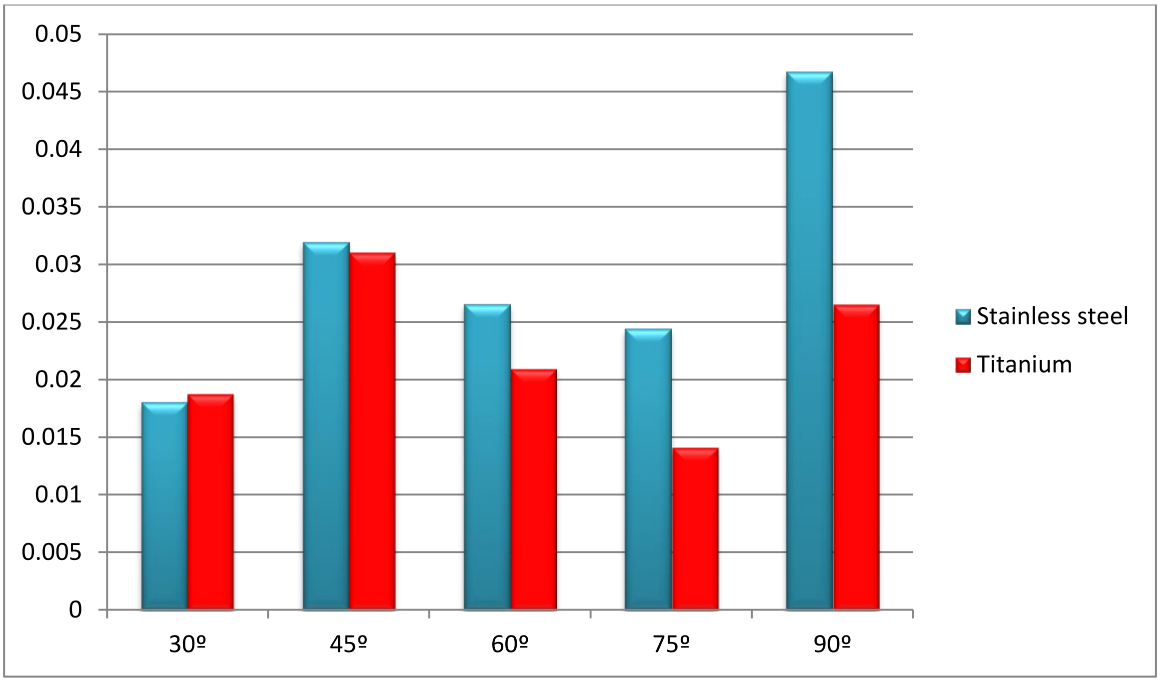


fig. 4 Stress generated in cancellous bone on application of Torsional force at various angulations of implant placement

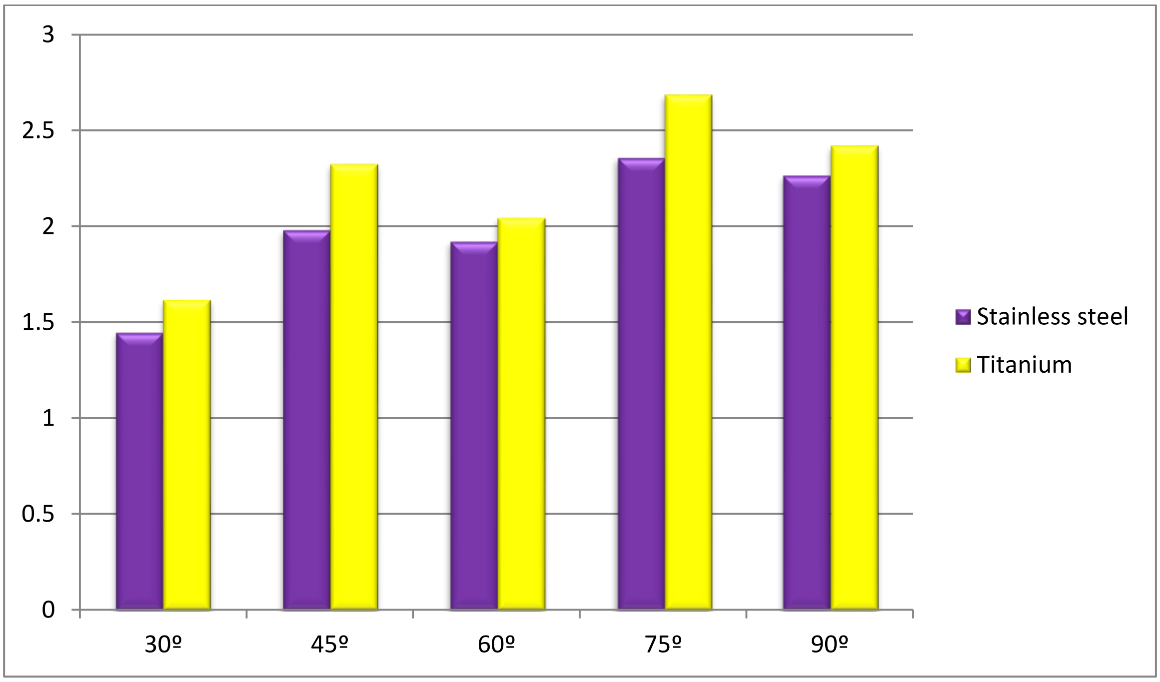


fig. 5 Stress generated in cortical bone on application of linear force at various angulations of implant placement.

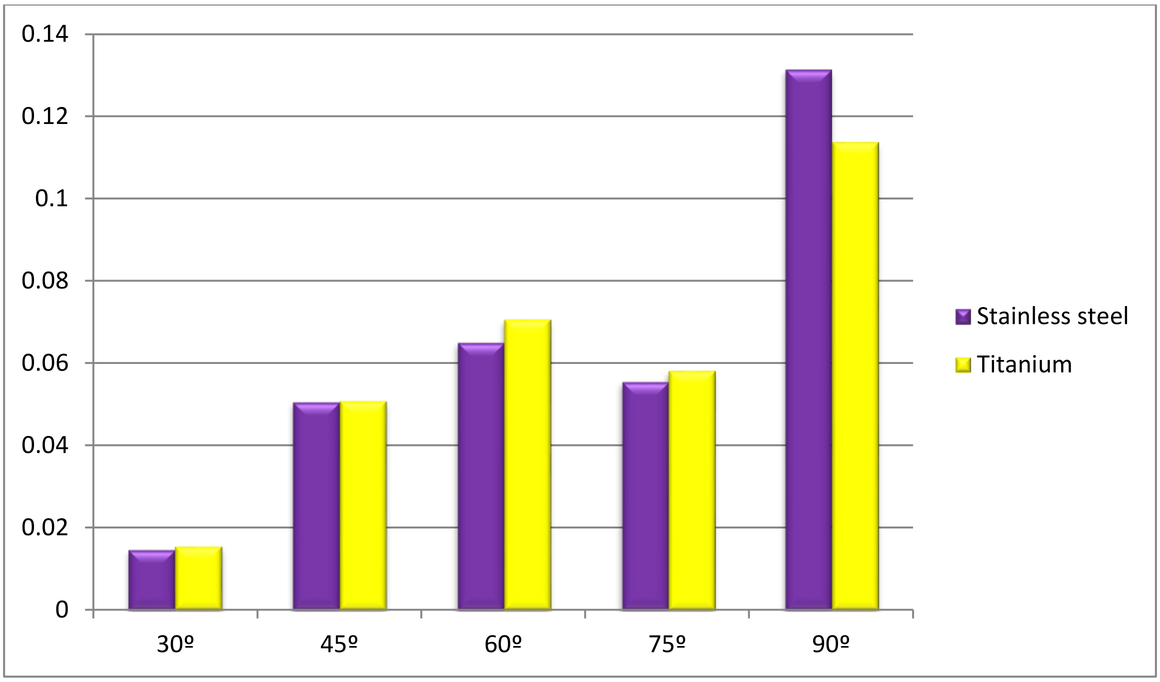


fig. 6. Stress generated in cancellous bone on application of linear force at various angulations of implant placement.

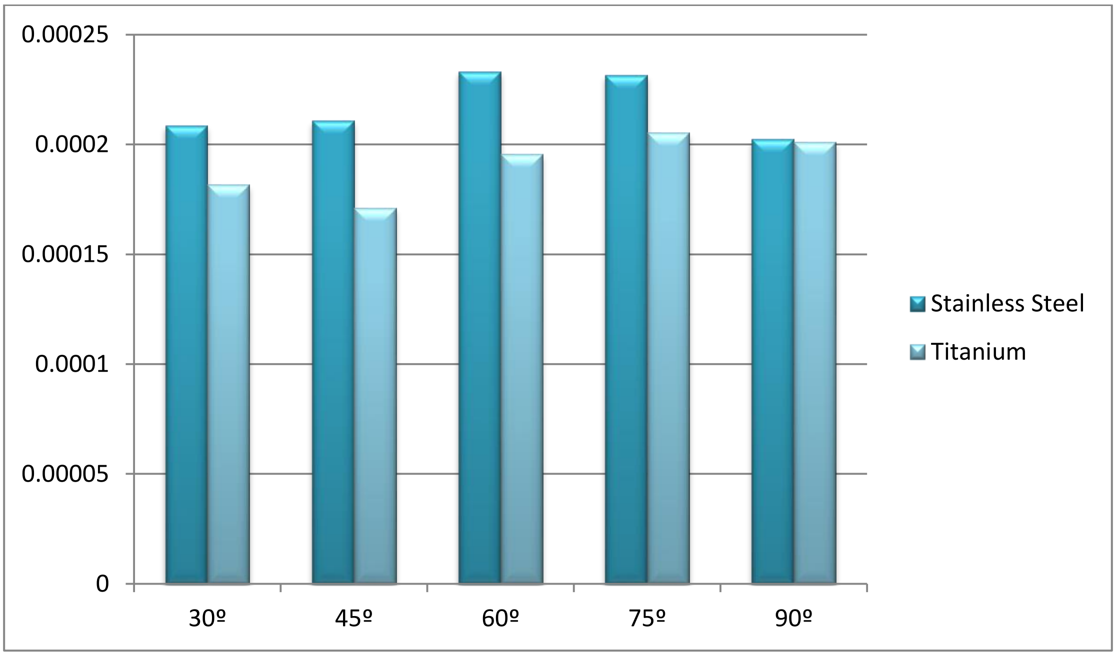


Fig. 7. Elastic deformation in both types of implants on application of torsional forces at different angulations

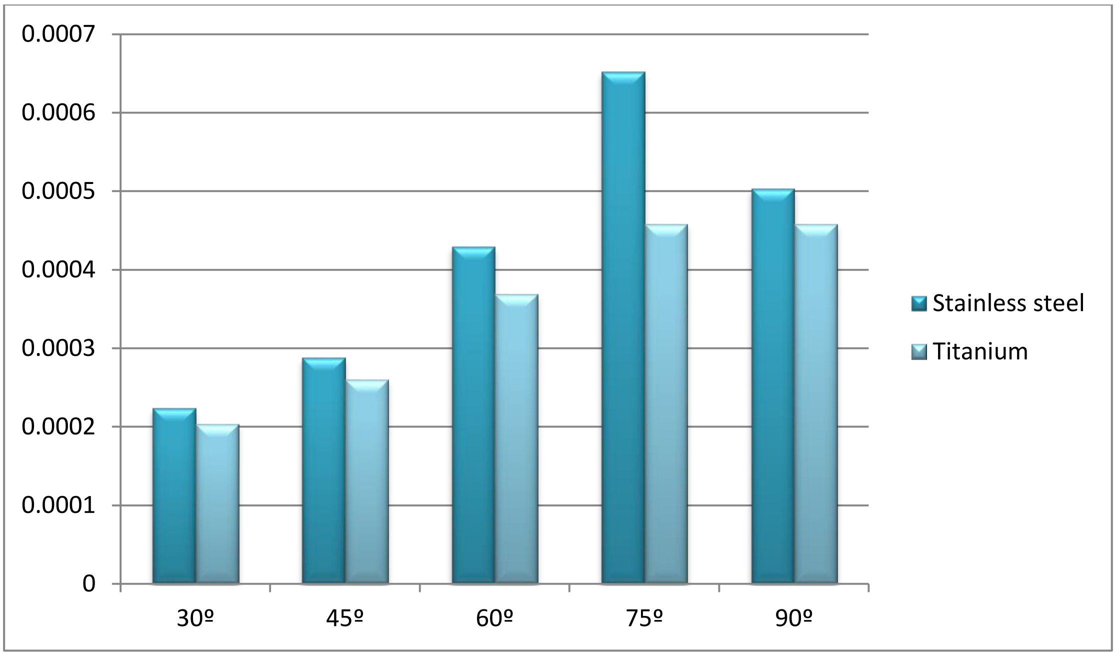


Fig. 8. Elastic deformation in both types of implants on application of linear forces at different angulations

Discussion

In the present study, finite element method has been used as it is a paramount tool for research in orthodontics as stated by **Knop**[72] *et al* , where they had highlighted certain parameters such as the ideal position of orthodontic appliances during specific mechanics, direction of tooth displacement, stress distribution in the alveolar bone, periodontal ligament, enamel, dentin pulp and auxiliaries such as mini implants, areas most likely to present root resorption and stress distribution on archwires can be studied effectively and accurately. It has got certain advantages such as it is accurate as it is done by a software, noninvasive, can control the variables in the study and it provides quantitative data regarding the structures which are located even deep inside the nasomaxillary complex.

In the present study, the site for implant placement was posterior buccal aspect of maxilla between second premolar and first molar as this site has the greatest mesiodistal buccal/palatal distance as stated given by **Fayed**[73] *et al*. **Kyung HM**[49] (2003), **Carano** *et al* [50] (2005) and **Melsen**[51] (2005) stated that most frequently used location for the placement of mini implant is the posterior buccal aspect of maxilla, although various other sites also serve this purpose.

The vertical height of insertion of implant used in this study was at 8mm from the alveolar crest in between second premolar and first molar as suggested by **Poggio**[74] *et al* as this site can utilized effectively without causing any harm to the maxillary sinus.

The angulations of implant insertion used in the present study ranged from 30^0 to 90^0. **Kravitz, Kusnoto**[46] (2007)and **Liou**[53] (2007)*et al* have postulated that insertion angle of 30^0 to 70^0 from the occlusal plane (i.e. 20^0 to 60^0 to the cortical bone long axis) is optimal and it will allow sufficient bone engagement. This will increase the potential for maximal anchorage, while preventing miniscrew slippage along the surface of the bone during insertion. Also, **Kravitz** and **Kusnoto**[46] (2007) stated that, though the acutely angled implant may have more cortical bone contact, the chances of slippage of mini implant are more as the acuteness of implant increases, especially if pilot hole is not used as a guide.

The biomechanical performance of the implant on the underlying bone under different angulations was also studied. The implant angulation has a significant effect on the bone contact. The less is the angulation of implant to the long axis of buccal alveoli, the more is the contact between the implant and the cortical bone, as stated by **Deguchi**[75] *et al.* They quantitatively evaluated cortical bone thickness in various locations in the maxilla and the mandible. Three-dimensional computed tomographic images were reconstructed for 10 patients. Cortical bone thicknesses were measured in the buccal and lingual regions mesial and distal to the first molar, distal to the second molar and in the premaxillary region at 2 different levels. Differences in cortical bone thickness at 3 angles (30 degrees, 45 degrees, and 90 degrees) were also assessed. Distances of the intercortical bone surface to the root surface and the root proximity were also measured at the above areas. They found that significantly less cortical bone thickness was observed at the buccal region distal to the second molar compared with other areas in the maxilla. Cortical bone thickness was found to be increased at 30 degrees than 45 degrees and it was more at 45 degrees when compared to 90 degrees.

According to literature, the placement angle for the mini implant may vary between 30^0 to 90^0, depending on the site of placement. In the present study, biomechanical performance of titanium and stainless steel implants at different angulations were studied, so that more evidence can be gathered regarding the material of mini implant in the bone, so that least amount of stresses are generated in both implant and bone and thus, minimising the chances of failure.

In the present study, when comparative evaluation of stainless steel and titanium mini implants at different angulations and two different forces was done, it was found that for both the titanium mini implants, when linear force was applied, the stress generated in the implant increased as the angle increased from 30^0 to 90^0. The maximum stress generated in the implant during application of linear force was in the range of 13.217-19.395 MPa. These findings are in accordance with the findings of **Brar**[76] in their finite element method, where it was seen that, as the angle of mini implant increases from 45^0 to 90^0, the maximum stress generated in the implant also increases.

The findings of the present study are also in accordance with those findings of **Sivamurthy**[77] *et al,* which show that, as the angle of implant insertion increases from 30^0 to 60^0 stress generated in the implant also increased. In their study, they used a force of 2N to simulate the force of anterior en-masse retraction.

Machado[78] had also found that, as the angle of implant insertion increased from 30^0 to 90^0, stress induced in the implant also increased. They used a force of 200 gms perpendicular to the long axis of miniscrew to simulate the reactionary force during en-masse retraction. They had constructed 10 different finite element models with implants of varying diameter and concluded that as the diameter of implant increases, the stress generated on the corresponding bone decreases.

On the contrary, **Motoyoshi**[79] *et al* in their study, found that the maximum stress generated in the implant ranged from 9.43-15.13 MPa. They applied a 2N force to the head of the mini-implant at 45 degrees to the bone surface. In their study six types of finite element models were designed to show various thread pitches from 0.5 to 1.5 mm. The difference in the maximum stresses generated in the implant may be attributed to the different pitch size and also to the different designs of the implant. The implants used in their study were cylindrical where as in our study, conical implants were used. Also, the force used in our study was 1.47N whereas in their study, force was used was 2N.

In the present study, when linear force of 150 gms was applied onto the stainless steel and titanium mini implants, the stress generated in the implant increased as the angle increased from 30^0 to 90^0. On the contrary, **Pollei**[3] in their study have found that as the angle increased from 45^0 to 90^0, the stress generated in the implant was found to be decreased. Our study showed that the greatest stress developed in stainless steel implant was at 90^0 which was 18.924 MPa and at 45^0 in titanium which was 19.395 MPa. Whereas in their study, the maximum stresses in implant were greatest at 45^0 for both stainless steel and titanium, which were 89.3 and 82.75 MPa respectively.

Contrary to our findings, **Singh**[80] *et al* in a finite element set up, found that under horizontal loading force of 350 gms, the maximum stress generated in stainless steel implant was 19.56MPa whereas in titanium implant was 11.35 MPa. Under torsional loading, the maximum stress in stainless steel and titanium miniscrew implants was 17.2 and 8.7 MPa respectively, which was slightly lower than those obtained under horizontal loading.

Also, contrary to our study, **Suzuki**[55] *et al* had found that greatest stress generated in their study was at 75^0 angle (75.2 MPa) for titanium implant. This great difference in the stress value of the mini implant at that angulation may be attributed to increased value of force (2N) and material properties. In their study, they used pure titanium and not the surgical grade titanium.

Jasmine[2] *et al* had found the results which are contrary to our findings. Maximum stress generated in implant was at 30^0 angle(77.49 MPa) and as the angle inceased to 45^0, 60^0 and 90^0 the stress decreased to 60.51 MPa, 60.12 MPa and 16.8 MPa. This difference may be attributed to the difference in the 3D modelling of the implant. In their 3D model, they had also constructed all the various other components such as anterior teeth, archwire, coil spring and crimpable hook.

Contrary to our findings, **Marimuthu**[58] in their study found that, the stress generated was decreasing as the angle increased from 30^0 to 90^0. The stress generated implant at 30^0, 60^0 and 90^0 was 14.07, 10.78 and 6.64 MPa respectively.

The stress generated in stainless steel implant under torsional force at 30 degrees was 15.3MPa while in Titanium implant it was 16.009 MPa. At 90 degrees, it increased to 16.249 MPa in Stainless steel implant, whereas in Titanium implant it increased to 16.416 MPa. these findings are consisitent with the finding of **Maya**[81] *et al* who in their study, found that, as the angle of insertion of mini implant increases, the torque required to insert the implant also increases. Though theirs was a cadaver based study, it can be postulated that as the insertion torque is increasing with the increase in the insertion angle, stress generated in the implant is also increasing.

Contrary to the findings of our study, **Arantes**[71] *et al* in their finite element setup, had found that, as the angle of mini implant insertion increase, the stress generated in the mini implant decreased. It must be noted that, in their study, they had compared the mini implants of two different brands, to find out that which brand is superior.

Stress generated in the bone (cortical and cancellous) was evaluated for both stainless steel and titanium mini implants at all the five angulations for both type of forces. Stress generated in bone (cortical and cancellous) loaded with stainless steel and titanium implants on application of torsional force. At 30 degrees implant insertion angle, the stress generated in cortical bone loaded with stainless steel implant was 1.72MPa whereas for Titanium it was 1.97MPa. the stress increased at 45 degrees to 2.12MPa and 2.55MPa in Stainless steel and Titanium implants respectively. It reduced at 60 degrees while it increased at 75 degrees. At 90 degrees, the stress generated was same i.e 1.9MPa in cortical bone loaded with both the type of implants. The stress generated in cancellous bone loaded with stainless steel and titanium implants increased as the implant insertion angle increased but there was no significant difference between the implants at different angles of insertion. These findings are consistent with the finding of **Arantes**[71] *et al* who compared the biomechanical performance of mini implants of two different brands (SIN and RMO), when a torsional force of 20N was applied when the implant was inserted at two angulations (45^0 and 90^0). Both the implants were manufactured using the surgical grade titanium. They selected an implant of one SIN brand as superior over the other and the stress generated by that implant at 45^0 in cortical bone, cancellous bone and in the implant itself was 245.65, 4.69 and 542.44 MPa ($x10^{-3}$ mm/mm) respectively and at 90^0 , stresses generated in cortical bone, cancellous bone and in the implant were 93.69, 2.92 and 186.68 MPa ($x10^{-3}$ mm/mm) respectively.

These findings are also in accordance with the findings of **Maya**[81] *et al* who used two types of mini implants (cylindrical and conical), the maximum insertion torque for cylindrical implants at 60^0 was 14.13 and at 90^0 was 17.27 MPa. On the other hand, the maximum insertion torque values for the conical mini-implants were 11.40 Ncm for the 60° angle and 14.40 Ncm for the 90° angle. It must be noted that, their study was a split mouth study and it was done on human cadavers.

In our study, stress generated in bone (cortical and cancellous) loaded with stainless steel and titanium implants on application of linear force was evaluated. At 30 degrees implant insertion angle, the stress generated in cortical bone loaded with stainless steel implant was 1.44MPa while in bone with titanium implant it was 1.62MPa. At

45 degrees angle, it increased to 1.98MPa and 2.32MPa in Stainless steel and titanium respectively. At 60 degrees angle, there was no significant change in the stress generated in cortical bone with both the types of implants. At 75 degrees, it increased. At 90 degrees angle, it increased to 2.26MPa and 2.42MPa in bone loaded with stainless steel and titanium implants respectively. The amount of stress generated in the cancellous bone loaded with stainless steel as well as titanium implant increased as the implant insertion angle increased from 30 degrees to 90 degrees. These findings are in accordance with values obtained by **Perillo**[1] *et al when* only 30^0 and 60^0 angles are considered. The implant was inserted at five angles (30^0, 60^0, 90^0, 120^0. 150^0) and a linear force of 2N was applied at two angles (0^0, 30^0). Considering only first three implant angulations, the maximum stress generated was at 60^0 implant angulation, which was 58 and 35 MPa for cortical and cancellous bone respectively. When direction of force was at 30^0, the maximum stress generated was also at 60^0 implant angulation which was 62.3 and 33.4 MPa respectively. The differences in the values obtained by them may be attributed to the increased length of screw (8mm) and an increased force of 2N.

Our findings are in accordance with the findings of **Lin**[57] *et al* (2013) who used stainless steel mini implants of three different lengths (8mm, 10mm, 12mm) at three different angulations (60^0, 90^0, 120^0) and orthodontic force of 2N was applied in three different angles (30^0, 45^0, 60^0). Also, the implants were exposed, i.e, the implants were kept outside and were not inserted completely. When direction of force was at 30^0 with minimum exposure was considered (3mm), and angle of insertion was at 60^0 the stress generated in cortical and cancellous bone was 1.72 and 0.25 MPa respectively and at 90^0, it was 1.74 and 0.21 MPa for cortical and cancellous bone respectively.

From the obtained results, it is clear that, as the angle of placement of implant increases, the stress generated in the implant also increases when both linear and torsional forces are applied. Also, stress generated in the bone (cortical and cancellous) and in the mini implant was also less for stainless steel implant when both linear and torsional forces were applied. These findings are in accordance with the findings of **Yao**[83] *et al*, who had stated that stainless steel exhibits better tensile properties than titanium.

In our study, only one value was obtained at each angle for one particular material of implant when one type of force was exerted. For eg : five values were obtained for stainless steel implant angulated at five different angles, when a linear force was exerted. For this reason, descriptive statistical methods were applied.

Analysis was also performed where stress generated in stainless steel implant (both linear and torsional) was compared with stress generated in titanium implant (both linear and torsional) separately, irrespective of the angulations. It was found that, on application of both linear and torsional forces, both the implants performed the same, i.e, the values were not significant.

It was also found that, when similar statistical tests were applied to quantify the difference of stress generated by both the implants material types in bone (cortical and cancellous) irrespective of angles, the results were not significant.

When elastic deformation between both the implants was compared, it was seen that there is no statistically significant difference between the elastic deformation of the implants made stainless steel and titanium.

Limitation

- The most crucial step in performing a finite element analysis is the modelling of the all the structures involved in the FEM setup. Hence, a good operator is required who can model the structures involved as they are.
- The tooth is treated as pinned to the bone via periodontal ligament. This rigidity in the elemental nodal complex can result in errors in calculations.
- To simulate physical environment certain assumptions are made which can result in errors in maximum stress calculations.

Summary

The present study was carried out to compare and evaluate the biomechanical performance of titanium and stainless steel mini implants in the maxilla at different angulations in a finite element setup. Linear force of 150 gms from the mesial direction and a torsional force of 20N/cm in the clockwise direction was applied to the head of the implant for both material types.

3D modelling of all the structures involved was done in dedicated computer aided design software named CATIA. The model consisted of tooth, periodontal ligament, lamina dura, cortical bone and cancellous bone. Five models were made, each with a different angulation of implant ranging from 30^0, 45^0, 60^0, 75^0 and 90^0.

A contracer machine was used for accurately defining the dimensions of the implant. The contracer machine is an industrial equipment which can measure the minutest details and generate a report regarding its outer anatomy. Using the dimensions as generated by contracer machine, the modelling of implant was done and the models were exported to software named ANSYS.

ANSYS is a dedicated mechanical analysis of software. Meshing of the models was done in this software. An extensive 393208 nodes and 157138 elements were meshed. In this software, we need to assign certain physical properties (Young's modulus and Poisson's ratio) of the materials to be tested. All the materials were assigned the properties and the model was fixed at upper boundary. All materials were considered as homogeneous and isotropic, linearly elastic

Analysis was done by applying 150 gms of linear force and 20N of torsional force on the implants with both material types. Based on the results, following observations were made:-

- Stress generated in stainless steel implants was less for both linear and torsional forces for at majority of the angles, when compared to titanium implants.
- Stress generated in both cortical and cancellous bone with both the materials under torsional force increased as the angle of insertion increased but there was no significant difference between the implants at different angles of insertion.
- The amount of stress generated in cortical and cancellous bone loaded with stainless steel as well as titanium mini implant on application of linear force increased as the angle increased from 30^0 to 90^0.

- The difference of stress generated in both stainless steel and titanium implants on application of both linear and torsional forces , irrespective of angles was not statistically significant.

- The difference between stress generated in cortical bone with application of both linear and torsional forces irrespective of angle was less with stainless steel mini implant when compared to titanium mini implant.

- The difference between stress generated in cancellous bone with application of both linear and torsional forces irrespective of angles was not significant between both stainless steel and titanium mini implants.

- The difference between elastic deformation between stainless steel and titanium implants under torsional for significant whereas it was insignificant under linear forces, irrespective of angles.

Conclusion

The findings for the study lead to following conclusion –

- Stress generated in stainless steel implant during linear force and torsional force was less than that generated in the titanium implant.
- Stress generated in stainless steel and titanium implant during linear and torsional force increased progressively from 30^0 to 90^0 for most of the angles.
- Difference in stress generated by stainless steel implant in cortical bone for both linear and torsional forces was less when compared to titanium implant whereas for cancellous bone, the difference was insignificant at all the angles
- Irrespective of angles, difference in stress generated in stainless steel implants and titanium implants for both the forces was not significant.

- Irrespective of angles, the elastic deformation occurring in both stainless steel and titanium mini implants was very less when compared to their respective Young's moduli.

- Stainless steel mini implants can prove as an effective choice over the titanium mini implants as they have better tensile properties when compared to titanium.

- Elastic deformation is similar for implants made up of either stainless steel or titanium when a torsional force of 20N and a linear force of 150gms is applied.

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