# DEFLECTION CHANGES OF MINI IMPLANTS AT DIFFERENT BONE DENSITIES: AN IN VITRO STUDY

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**THE TAMILNADU Dr. M.G.R. MEDICAL UNIVERSITY**

**In partial fulfilment for the degree of**

**MASTER OF DENTAL SURGERY**

**BRANCH V**

**DEPARTMENT OF ORTHODONTICS**

**2015 - 2018**

# CERTIFICATE

This is to certify that this dissertation titled "**DEFLECTION CHANGES OF MINI IMPLANTS AT DIFFERENT BONE DENSITIES: AN IN VITRO STUDY"** is a bonafide work done by **Dr. INDRA . AN** under my guidance during her post graduate study period between 2015 – 2018.

This dissertation is submitted to **THE TAMIL NADU Dr. M.G.R. MEDICAL UNIVERSITY** in partial fulfilment for the degree of **Masters in Dental Surgery, in Branch V – Orthodontics and** Dentofacial Orthopaedics. It has not been submitted either partially or fully for the award of any other degree or diploma.

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# CONTENTS

# *INTRODUCTION*

Orthodontic treatment involves the application of optimal force systems to teeth, with the intention of inducing a biological response that results in tooth movement.[1] Orthodontists accomplish this by constructing appliances that will produce certain desired tooth movements through precise application of forces using auxiliaries such as elastics, springs, and flexible wires composed of various alloys.

Newton's third law states that for every action, there is an equal and opposite reaction.[2] When forces are applied between groups of teeth, one can expect resultant movements of all groups involved to varying degrees. Since most orthodontic appliances are tooth borne, reactive forces generated by the appliance system can result in undesired tooth movements. Proffit defines the term *anchorage* in orthodontic applications as "resistance to unwanted tooth movements."[1]

Traditionally, anchorage was provided extra-orally by the use of headgears and facemasks or intra-orally by acrylic pads resting on palatal tissues and groups of teeth consolidated as a unit.[1] Ideally, teeth that serve as anchorage units should remain stationary, but in reality, undesirable side effects result from force systems that rely on other teeth within the same or opposing arch for support.

However, even a small reactive force can cause undesirable movements; it is important to have absolute anchorage to avoid them. Absolute or infinite anchorage is defined as no movement of the anchorage unit (zero anchorage loss) as a consequence to the reaction forces applied to move teeth. Such an anchorage can only be obtained by using ankylosed teeth or dental implants as anchors, both relying on bone to inhibit movement. Anchorage provided by devices, such as implants or miniscrew implants fixed to bone, may be obtained by enhancing the support to the reactive unit (indirect anchorage) or by fixing the anchor units (direct anchorage), thus facilitating skeletal anchorage.

Temporary Anchorage Devices (TADs) are routinely used as a means of skeletal anchorage in contemporary orthodontics. Their multifaceted use has revolutionized our specialty as we can use them as means for direct or indirect anchorage for various types of orthodontic tooth movements. Miniscrew implants (MSIs) are a treatment adjunct designed to provide absolute skeletal anchorage in orthodontics. They have gained in popularity due to their simplicity in placement, low cost, patient-acceptance and ability to eliminate patient compliance issues in treatment.[3]

Various materials were used for the manufacturing of mini screws before the use of titanium and its alloys came into existence. Some of the materials previously considered were Cobalt -Chromium alloy (Vitallium) and Stainless steel.

Compared to 316L stainless steel, the Ti alloy implants, made with aluminum (Al) and vanadium (V) [Ti-6Al-4V] alloys, are roughly of equal strength, but it has half the elastic modulus. So titanium implants have the advantage over stainless steel as they have high bioactivity and more flexibility that improve integration and mechanical fixation.[4] Torsional properties of stainless steel screws are different from titanium screws. Stainless steel bone screws are easier to handle because the surgeon can feel the onset of plastic deformation and this provides adequate pre-warning to avoid over-torquing the screw while titanium screws break suddenly.[4]

**Carano et al**[5] evaluated the mechanical properties of three commercially available self-tapping screw systems namely Leone (surgical stainless steel), Dentos (titanium grade IV), M.A.S (titanium grade V). The result showed that all three miniscrews have enough resistance to failure during insertion, application, and removal in orthodontics. Although stainless steel has demonstrated to be more resistant to failure than titanium, its overall performance as material for miniscrew could be inferior to titanium.

Thread designs of orthodontic mini-screws have evolved over the years. Selftapping designs, otherwise known as “non-drill-free” screws, require pilot-hole preparation prior to insertion. Today, most manufacturers are promoting the advancement of self-drilling or “drillfree designs where mini-implants are placed in a one step procedure eliminating the need for pre-drilling.

Mini-screw diameters fall within 1.0-2.3mm, and lengths range from 4mm-20mm. Currently, titanium alloy mini-implants of 1.3-1.8mm in diameter and 6-10mm in length, are most popular in everyday clinical orthodontics.[6]

Bone quality also plays a major role when deciding on a mini-implant placement site as it is among the most important factors for achieving good primary stability.[7] It is important for a clinician to understand that bone density and cortical bone thickness varies throughout the oral cavity. Bone density in general is higher in all regions of the mandible than in the maxilla. It has been reported that the placement site should have a cortical bone thickness of more than 1.0mm in order to attain adequate primary stability for mini-implant success.[8]

Cortical bone thicknesses vary tremendously throughout the maxilla and mandible. Anterior regions of the maxilla contain significantly higher proportions of cortical bone than the posterior maxilla, while the reverse is true in the mandible.[9, 10] As a general guideline, cortical bone thicknesses reach approximately 1.0-2.2mm in the anterior alveolar process of the maxilla and hard palate. The cortical bone becomes significantly thinner in the posterior maxilla and tuberosity region, often reaching thicknesses of less than 1mm. Cortical bone thickness is on average 1.0-1.5mm in the anterior interradicular sites of the mandible, increases to 1.5-2.5mm in the canine and premolar interradicular areas, and can reach thicknesses greater than 3.0mm in the mandibular molar and retromolar region.[11]

Bone density is classified into 4 groups based on microscopic cortical and trabecular bone characteristics:

- D1 - Primarily dense cortical bone
- D2 - Dense to thick porous cortical bone on the crest and coarse trabecular bone
- D3 - Thin porous cortical crest and fine trabecular bone
- D4 - Minimal to no crestal cortical bone

Regions of D1- D3 bone have been found to be adequate for temporary anchorage device (TAD) insertion. TADs placed in D1 and D2 bone exhibit lower stress at the screw-bone interface and may provide greater stationary anchorage during loading. Placement in D4 bone is not recommended owing to the high failure rate associated with it (35-50 percent).[12]

Insertion torque is the result of frictional resistance between screw threads and bone and is reported to determine primary stability. Insertion and removal should be done at a slow steady rate with a continuous force so that the load on both the screw and bone will be low. All mini screws are susceptible to breakage upon reaching a certain torque level. However there is a range of safety between recommended insertion torque and maximum insertion torque. **McManus et al[13]** reported that the mean maximum placement torque in the maxilla was 4.6 Newton centimetres [Ncm] and in the mandible it was 8.64 Ncm. **Friberg et al[14]** described a positive correlation between mini-implant insertion torque and bone density values.

When an implant is inserted into bone, due to the resistance offered by the bone, the implant is liable to undergo deviation from its original path. This interaction between the implant and bone is dependent on both the dimensions of the implant and bone density. This deflection or deformation can ultimately lead to fracture or failure of the mini implant.

Mini implant failure can involve factors related to the clinician, the patient and the screw itself. According to **Kuroda et al** [15], root proximity is one of the major risk factors for failure of mini implants. Placement of a mini screw too close to a root can also result in insufficient bone remodelling around the screw and transmission of occlusal forces through the teeth to the screws leading to implant failure. Considering that majority of the mini implants for orthodontic usage are placed in inter-dental areas, a slight deflection from the intended path can thus affect their success.

Hence this study has focussed on evaluating the deflection of titanium alloy self-drilling mini implants from the intended path that occurs during placement.

# *AIMS AND OBJECTIVES*

AIM :

The aim of this in vitro study is to radiographically evaluate the deflection of titanium alloy self-drilling mini implants from the intended path that occurs during placement.

OBJECTIVES:

(1) To determine the deflection changes of the mini implants from its intended path of insertion.

(2) To evaluate the role of bone densities on deflection.

(3) To evaluate the role of implant lengths on deflection

# *REVIEW OF LITERATURE*

**Gainsforth and Highley (1945)**[16] introduced the concept of skeletal anchorage with their animal study. In each of five dogs in their experiment, a screw of vitallium was placed in the anterior border of the ramus of the mandible, one side on each dog. Traction was applied to the screw by means of orthodontic elastics connected to a maxillary appliance. Examination of the bones from sacrificed animals showed a wide destructive process after implantation of either the screws or rings in the rami. Tooth movement was accomplished using basal bone anchorage, but an effective force could not be maintained for more than thirty one days in any case. All of the screws came out in sixteen to thirty one days.

**Misch et al (1988)**[17] proposed the following four bone density groups based on microscopic cortical and trabecular bone characteristics: D1, primarily dense cortical bone; D2, dense to thick porous cortical bone on the crest and coarse trabecular bone; D3, thin porous cortical crest and fine trabecular bone; and D4, minimal to no crestal cortical bone. Suggested implant designs, surgical protocols, healing processes, treatment plans, and progressive loading time spans should be modified for the individual bone density types.

**Melsen et al (1998)**[18] investigated the Aarhus Mini-implant by inserting them in the infra-zygomatic crest and the mandibular symphysis of Macaca monkeys and immediately loading the implants with a force ranging between 0.25-0.50 N in 1 to 6 months period of time. Histologically the screws exhibited a degree of osseointegration varying from 10 to 50 % which was time dependent, but independent of the type of bone and the amount of applied force.

**Tehemar et al (1999)**[19] evaluated factors affecting heat generation during implant site preparation and stated that heat generation increases during drilling in dense bone. Therefore, when placing the mini implants into high density areas such as retromolar and posterior areas in the mandible, clinicians must be careful not to generate heat. Heat generation can be prevented by irrigating abundantly with saline solution, not applying too much pressure on the bone and not using a worn drill. Also, large diameter drill can be used instead of a small diameter drill.

**Masumoto et al (2001)**[20] experimented using 31 dry skulls in a group of Japanese males, and measured buccal cortical bone thickness at the mandibular first molar. An observed range of 2.27 mm to 3.82 mm was found for bone thickness at the mandibular first molar. Each skull was categorized into three groups: short, average, and long facial type. These categorizes were based upon Frankfort-mandibular-plane angle and correlated to buccal cortical bone thickness. The short facial type and small mandibular plane angle had significantly increased buccal cortical bone thickness.

**Miyawaki et al (2003)** [21] compared the success rates of various diameter orthodontic mini screws with mini-plates in the maxilla and mandible of fifty-one patients that were subsequently loaded with an applied orthodontic force of less than 2N. All ten orthodontic mini screws with a 1.0mm diameter and 6mm length failed in this study, despite the relatively high success rates for the other test groups. The second group, consisting of one hundred and one orthodontic mini screws (1.5mm diameter; 11m length), had an 83.9% success rate over the one-year study period. This was comparable to the largest diameter orthodontic mini screws (2.3mm diameter; 14mm length) utilized, reporting a success rate of 85.0%.

**Tadas et al (2003)**[22] performed a 3- dimensional finite element analysis to evaluate the influence of implant length as well as that of bone quality, on the stress/strain in bone and implant. The results of this study suggest that bone of higher rather than lower density might ensure a better biomechanical environment for implants. Moreover, longer screw-type implants could be a better choice in a jaw with bone of low density.

**Kim et al (2005)**[23] evaluated the effects of drilling procedure on the stability of the screws under early orthodontic loading. 32 screws were inserted into the jaw of 2 beagles. Screws in drilling group were inserted into the site that had been drilled with a pilot drilling bur, and those in the drill free group were inserted without drilling. A force of 200 -300g was applied using nickel- titanium coil springs 1 week after insertion. Twelve weeks after insertion, mobility was tested and the screws with the surrounding bone were prepared for histomorphic evaluation. Less mobility and more bone - to – metal contact was seen in drill free group.

**Melsen et al (2005)**[24] stated that self drilling miniscrews should be inserted slowly, with minimal pressure, to assume maximum miniscrew bone contact. A pilot hole is recommended in regions of dense cortical bone, even for self drilling mini screws. During mini screw placement in dense cortical bone, the clinician should consider periodically derotating the miniscrew 1or 2 turns to reduce the stresses on the mini screw and the bone. The clinician should stop inserting the miniscrew as soon as the smooth neck of its shaft has reached the periosteum. Overinsertion can add torsional stress to the mini screw neck, leading to screw loosening and soft tissue overgrowth.

**Deguchi et al (2006)**[25] investigated maxillary and mandibular cortical bone thickness mesial and distal to the first molars, distal of the second molars, and in the premaxillary region of ten patients. Cone beam CT scans with slice thickness of 0.5mm were taken in high-resolution mode and measurements of cortical bone thickness were made at various angles (30°, 45°, and 90°) relative to a line parallel to the long axis of the adjacent teeth in the maxilla and mandible. A significant difference between maxillary and mandibular measurements mesial and distal to the first molar and distal to the second molar was observed. Reported measurements of lingual cortical bone thickness were similar to those at the corresponding buccal positions, except at the distopalatal aspect of the second molars where significantly thicker cortical bone was present. In the premaxilla, mean cortical bone thickness at A-point was significantly less than at the anterior nasal spine.

**Motoyoshi et al (2006)**[26] determined an adequate placement torque for obtaining a better success rate of mini-implants that are screwed into the buccal alveolar bone of the posterior region as an anchor for orthodontic treatment. The success rate of the mini-implant anchor for 124 implants was 85.5%. The mean implant placement torque ranged from 7.2 to 13.5Ncm, 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 significantly higher in the failure group than in the success group. Therefore, a large implant placement torque should not be used always. According to the calculations of the risk ratio for failure, to raise the success rate of 1.6-mm diameter mini implants, the recommended implant placement torque should be within the range from 5 to 10Ncm.

**Park et al (2006)**[27] examined the success rates and find factors affecting the clinical success of screw implants used as orthodontic anchorage. Mobility, jaw (maxilla or mandible), and side of placement (right or left), and inflammation showed significant differences in success rates. To minimize the failure of screw implants, inflammation around the implant must be controlled, especially for screws placed in the right side of the mandible

**Poggio et al (2006)**[28] provided clinical indications for a safe application of the miniscrews. Volumetric tomographic images of 25 maxillae and 25 mandibles were examined. For each interradicular space, the mesiodistal and the buccolingual distances were measured at two, five, eight, and eleven mm from the alveolar crest. In the maxilla, the greatest amount of mesiodistal bone was on the palatal side between the second premolar and the first molar. The least amount of bone was in the tuberosity. The greatest thickness of bone in the buccopalatal dimension was between the first and second molars, whereas the least was found in the tuberosity. In the mandible, the greatest amount of mesiodistal dimension was between first and second premolar. The least amount of bone was between the first premolar and the canine. In the buccolingual dimension, the greatest thickness was between first and second molars. The least amount of bone was between first premolar and the canine.

**Wilmes et al. (2006)**[7] examined the parameters affecting the primary stability of several orthodontic mini screws. One-thousand mini screw insertions were undertaken with variable pre-drilling in the ilium of country pigs and the insertion and removal torques were measured. The authors found no significant differences in cortical bone thickness based on sex or age. Aside from differences between the jaws, there was little difference observed in cortical bone thickness, especially about the first molars.

**Song et al (2007)**[29] evaluated the effect of cortical bone thickness on the maximum insertion and removal torque of different types of self-drilling mini-screws and to determine if torque depends on the screw design. Differences in the cortical bone thickness had little effect on the maximum insertion and removal torque in cylindrical type. There were significant relationships between cortical bone thickness, maximum insertion and removal torque, and implantation time in each type of self-drilling mini-screw. Since different screw designs showed different insertion torques with increases in cortical bone thickness, the suitable screw design should be selected according to the cortical thickness at the implant site.

**Chaddad et al (2008)**[30] examined the role of surface characteristics on primary stability and survival rates of orthodontic mini screws. Seventeen machined smooth titanium Dual-Top orthodontic mini screws (1.4mm, 1.6mm, and 2.0mm diameters; 6.0mm, 8.0mm, and 10.0mm lengths) and fifteen sandblasted, acid-etched surface treated mini screws with a 2mm polished collar (1.8mm diameter; 8.5mm, 9.5mm, and 10.5mm lengths) were placed in ten patients. Pre-drilling of the cortical bone was done prior to insertion for all mini screws, and a torque ratchet was used in placement to determine insertion torque values. Immediate loading of all mini-screws was performed with a 50- 100g force (NiTi coil-spring or elastic chain), which was increased to 250g of applied force after two weeks. There were no statistically significant differences in primary stability or survival rates over the 150-day study period between those mini screws with and without surface treatment to enhance osseointegration.

**Cheol Hyun Moon et al (2008)**[31] determined the factors related to success rate of orthodontic miniscrew implants placed at the attached gingiva of the posterior buccal region. They concluded placement site could be considered as one of the important factors to get better result as bone quality is known to be one of the major factors in the stability of mini screws.

**Ilser et al(2008)**[32] compared the parameters associated with implant insertion using two different methods of enhancing implant primary stability and to identify any relationship between these parameters at implant insertion. A total of 60 implants were placed in the maxillary posterior regions of 22 patients. The bone densities at the implant sites were recorded using a computerized tomography machine in Hounsfield unit (HU). The maximum insertion torque data were recorded. Strong correlations were observed between the bone density and insertion torque, and implant stability values at implant placement. The results of this study suggest that using thinner drills for implant placement in the maxillary posterior region where bone quality is poor may improve the primary implant stability, which helps clinicians to obtain higher implant survival rates.

**Kim et al (2008)**[33] compared the stability of cylindrical miniscrews, 7 mm in length, with that of tapered mini screws 5 mm in length, using torque values to determine if the healing time before loading affects the stability of the mini screw and if the insertion torque is assosciated with the removal torque measured after a few weeks of healing in tibias of twelve rabbits.

There was no significant difference between tapered and cylindrical screws in terms of the mean insertion or removal torque values within each group. The shorter tapered screw showed similar stability to the cylindrical screw, which strongly suggests that the tapered shape is more advantageous than the cylindrical shape. Removal torque did not increase significantly over time. They recommended immediate loading of miniscrew.

**Ono et al (2008)**[11] investigated cortical bone thickness in the posterior alveolar regions of the maxilla and mandible in forty-three orthodontic patients. Cortical bone thickness was measured at 1.0mm intervals in a plane parallel to the occlusal plane of each tooth from 1mm to 15mm below the level of the alveolar crest. Overall, average cortical bone thickness ranged from 1.09mm to 2.12mm in the maxilla, and from 1.59mm to 3.03mm in the mandible, with maxillary cortical bone thickness significantly thinner than that observed in the mandible. More specifically, mesial to the first molar, average cortical bone thickness ranged from 1.09mm to 1.62mm in the maxilla, and 1.59mm to 2.66mm in the mandible.

**Rubelisa et al (2008)**[34] evaluated the assosciation between trabecular bone density measurements of implant sites. Differences in the bone densities of the 4 anatomical regions in the mouth were significant, with the mandible yielding a higher mean density value, followed by the anterior maxilla, posterior mandible and posterior maxilla. This confirms the importance of a site specific bone tissue evaluation prior to implant installation.

**Seon-A Lim et al (2008)**[35] determined the variation in the insertion torque of orthodontic miniscrews according to the screw length, diameter, and shape. There was a significant increase in torque with increasing screw length and diameter. The insertion torque was affected by the outer diameter, length, and shape in that order. An increase in screw diameter can efficiently reinforce the initial stability of the miniscrew, but the proximity of the root at the implanted site should be considered.

**Turkyilmaz et al (2008)**[36] presented clinical study to determine the local bone density in dental implant recipient sites using computerized tomography (CT) and to investigate the influence of local bone density on implant stability parameters and implant success. Insertion torque and resonance frequency analysis were used as implant stability parameters. The peak insertion torque values were recorded with OsseoCare machine. CT is a useful tool to determine the bone density in the implant recipient sites, and the local bone density has a prevailing influence on primary implant stability, which is an important determinant for implant success.

**Chun et al (2009)**[37] evaluated bone density differences between interradicular sites. Bone densities in most areas were higher than 850 HU. Bone densities in both maxilla and mandible significantly increased from the alveolar crest toward basal bone in posterior areas, while the opposite was observed in anterior areas. Bone densities progressively increased from anterior to posterior areas in the mandible. The results suggest that mini-implants for orthodontic anchorage may be effective when placed in most areas with equivalent bone density up to 6 mm apical to the alveolar crest. Site selection should be adjusted according to bone density assessment.

**Jan D'haese et al (2009)**[38] in their study of prosthetic implants observed the difference in mean apical deviations that was related to implant length, with longer implants showing significantly higher apical deviation compared with shorter ones. This was explained by the fact that drilling deeper into the bone with a similar angle of insertion results in a higher apical deviation for a longer than for a shorter implant.

**Jin Hugh Choi et al (2009)**[39] determined bone density at various orthodontic implant sites and compare them according to depth and area. Bone density tended to decrease with increasing depth, particularly in the posterior area. Mean bone density showed a progressive increase from posterior to anterior region. The mean bone densities between the maxilla and the mandible showed higher values in the mandible, and these differences were more significant on the buccal side of the posterior. The differences in bone densities according to depth and area should be considered when selecting and placing miniscrew implants.

**Motoyoshi et al (2009)**[40] evaluated Cortical bone thickness at mini-implant placement sites in 65 orthodontic patients and was found to be directly proportional to the success rate of the mini-implant. To examine the biomechanical effects of cortical bone thickness, finite element models were made for cortical bone thickness from 0.5 to 1.5 mm, at 0.25-mm intervals. Cortical bone models without cancellous bone were constructed to examine the biomechanical influence on cortical bone after cancellous bone resorption. Cortical bone thickness influenced the stresses in the cancellous bone, but could not directly influence the stresses in the cortical bone. For Cortical bone thickness < 1 mm, the cancellous bone models exhibited von Mises stresses exceeding 6 MPa, and the cortical bone models without cancellous bone showed von Mises stresses exceeding 28 MPa.

**Noble et al (2009)**[41] recommended that as long as root damage can be avoided, mini implants should be placed as perpendicular to the bone as possible (90˚ angulation). Also concluded that placement of mini implants at 90˚ to the cortical plate is the most retentive insertion angle.

**Stahl et al (2009)**[42] evaluated the effect of various Deflections of the implants varied between 2 µm and 20 µm. The deflections of the implant increased as Young's modulus of the cancellous bone dropped with a cortical thickness of 1 mm. When the load direction was tilted in a buccal direction, the stresses and amount of strain were reduced by as much as 35%.parameters in regard to various implant types, sizes, and load directions using the finite element method.

**Zhao et al *(2009)***[43] in his study of different healing times before loading found that 3 weeks is an important time point for implant-bone units to gain biomechanical strength and integration. Osseointegration found after CT scans and maximum force during pullout testing were significantly correlated with healing time.

**Borges et al (2010)**[44] assessed maxillary and mandibular alveolar and basal bone density in Hounsfield units In the maxilla, the greatest bone density was found between the premolars in the buccal cortical bone of the alveolar region. The maxillary tuberosity was the region with the lowest bone density. Bone density in the mandible was higher than in the maxilla, and there was a progressive increase from anterior to posterior and from alveolar to basal bone.

**Crismani et al (2010)**[45] did a systematic review of effects related to patient, screw, surgery, and loading on the stability of miniscrews. Screw diameters of 1 to 1.1mm yielded significantly lower success rates than those of 1.5 to 2.3 mm. Screw placement with or without a surgical flap showed contradictory results between studies. Three studies showed significantly higher success rates for maxillary than for mandibular screws. Loading and healing period were not significant in the miniscrews' success rates. Authors proposed that screws under 8 mm in length and 1.2 mm in diameter should be avoided. Immediate or early loading up to 200 cN was adequate and showed no significant influence on screw stability.

**Florvaag et al (2010)**[46] examined five self-drilling and self-tapping mini-screw types with variable diameters ranging from 1.6mm to 2.0mm, and minimum lengths of 8mm. Overall, one hundred and ninety six mini screws were placed, with and without pilot hole preparation in thirty bovine femoral heads, utilized for the striking similarity in cortical bone thickness relative to human maxillary and mandibular alveolar cortices. All mini screws were inserted perpendicular to the bony surface, but pull-out testing was performed at three inclinations relative to the long axis of the mini screw: axially, 20°, and 40°. The three cylindrical mini screw designs placed with drill-free insertion achieved the highest axial pull-out values. The cylindrical mini screws also exhibited the greatest mean values for pull-out tests performed at 20° angulations. However, it was the cylindrical mini screws that showed the most significant decrease in pull-out resistance.

**Li et al (2010)**[47] studied the prosthetic implants under finite element analysis. The results indicated that in the posterior mandible, implant diameter plays more significant roles than length in reducing cortical bone stress and enhancing implant stability. However, implant length is more effective than diameter in reducing cancellous bone stress under loading.

**Okumura et al (2010)**[48] performed a finite element analysis to investigate the effect of maxillary cortical bone thickness, implant design and diameter on stress around implants. Regardless of load direction, implant design and diameter, cortical and cancellous bone stresses increased with the decrease of crestal cortical bone thickness. To improve implant success in the posterior maxilla, rather than implant selection, careful preoperative evaluation of the cortical bone at the planned implant site is recommended. If this cortical bone is very thin or even lacking, implant treatment should be carried on with caution by progressive loading in the range of functional loads.

**Yan Chen et al (2010)**[49] measured insertion and removal torque of 360 self drilling micro implants inserted in three types of artificial bone. They concluded that IT is an important indicator for insertion resistance and holding power. The ideal mechanical IT is dependent on the diameter of the micro-implants. Using a self-drilling technique, micro-implants with a diameter of less than 1.3 mm are unsuitable for insertion into a bone with a density greater than 40 pcf mechanically.

**Barros et al (2011)**[50] evaluated the effect of mini-implant diameter on fracture risk and selfdrilling efficacy. 405 mini-implants with 9 diameters from 1.2 to 2.0 mm were used. The fracture resistance index was remarkably greater for each 0.1 mm added in diameter. The placement torque increased significantly, whereas the axial placement load was progressively reduced during placement. Increases in mini-implant diameters significantly influenced the increases of placement torque and fracture torque on quantities that progressively reduced the fracture risk. The self-drilling efficacy was not strongly influenced by diameter.

**Isoda et al (2011)**[51] assessed bone quality with density values obtained by cone-beam computed tomography and to determine the correlations between bone density and primary stability of dental implants. Statistically significant correlations were found between the density values and insertion torque, density values and implant stability quotient, and insertion torque and implant stability quotient. The bone quality evaluated by specific CBCT showed a high correlation with the primary stability of the implants. Hence, preoperative density value estimations by CBCT may allow clinicians to predict implant stability.

**Marquezan et al (2011)**[52] evaluated bone density in two bovine pelvic regions and verify the primary stability of miniscrews inserted into them. However, the miniscrew primary stability was not different when varying the bone type. Insertion torque and pull out strength were not influenced by these differences in bone density when cortical thickness was about 1 mm thick.

**Oguz Ozan et al (2011)**[53] evaluated the correlation between the density of bone where implants were placed and the angular deviations that occur between the virtually planned and actually placed implants. They concluded that the lower bone density values have resulted in the greater angular deviations in the group. This deviation might have been derived from the free hand placement of implants and poor quality of bone.

**Suzuki et al (2011)**[54] analyzed the maximum insertion torque and maximum removal torque values of orthodontic miniscrews. Maximum insertion torque values were significantly higher for the self-drilling miniscrews (14.5 Ncm) than for the predrilling miniscrews (9.2 Ncm) in all implant sites. For both predrilling and self-drilling miniscrews, the highest maximum insertion torque values were observed at the midpalatal suture site followed by the dentoalveolar bones of the mandible and maxilla, respectively. In contrast, Maximum removal torque values were significantly higher for the predrilling miniscrews (22.6 Ncm) than for the self-drilling miniscrews (17.6 Ncm)

**Wilmes et al (2011)**[6] quantitatively analysed the impact of bone quality and predrilling diameter on the insertion torque of five different mini implants. Twenty pig iliac bone segments were discussed and embedded in resin. The insertion torques of mini implant of sixes 1.6x8mm, 1.6x10mm, and 2.0x10mm of two different manufacturers were measured. The pilot drilling was performed using a bench drilling machine at 915 rpm with pilot drills 1.1, 1.2 and 1.3 mm. During rotation, insertion and removal torques are measured. Insertion torques increased with smaller pre-drilling diameters in all mini-implant types. The results clearly showed that bone quality, the design and size of the mini- implants and the preparation of the implantation site influence insertion torques, and therefore on primary stability.

**Woodall et al (2011)**[55] found that the anchorage resistance of an implant placed at 90˚ to the alveolar bone was dramatically greater than that of an implant placed at either 60˚ or 30˚. The cortical bone stress created by loading 90˚ placed implants was less than the bone stress created by loading screws at either 60˚ or 30˚.

**Abhishek et al (2012)**[56] evaluated maximum equivalent stress distribution and maximum deflection assosciated with mini implants placed in 2 different cortical bone thickness. Greater stress and deflection were observed with 1.5 rather than 2mm cortical bone thickness. Greater cortical bone thickness gives better initial stability.

**Ankit H. Shah et al (2012)**[57] experimentally studied the effects of altering implant length, outer diameter, cortical bone thickness, and cortical bone density on the primary stability of orthodontic miniscrew implants. The 6-mm mini-implant displayed significantly higher insertion torque than the 3-mm mini-implant did. The 3-mm mini-implant with 2.0-

mm outer diameters showed significantly higher insertion torque than the 3-mm MSIs with 1.75-mm outer diameters. The IT was significantly greater for the mini-implant placed in thicker and denser cortical bone. Both outer diameter and length affect the stability of mini-implants. Increases in cortical bone thickness and cortical bone density increase the primary stability of the mini-implants.

**Cho et al (2012)**[58] investigated the effects of orthodontic mini-implant shape and predrilling depth on the mechanical properties of mini-implant during the insertion procedure. In the same predrilling depth, no differences were observed in maximum insertion torque between cylindrical and tapered groups. In cases of thick cortical bone, predrilling might be an effective tool for reducing microdamage without compromising mini-implant stability.

**Lindsy Holm et al (2012)**[59] evaluated the effects of mini-implant features (length, design, core diameter), insertion technique (insertion angle, cortical punch), and cortical bone depth and density on mini implant primary stability. Mini-implants achieved greater primary stability in higher-density cortical bone, and the 1.5 mm diameter tapered and 2.0 mm cylindrical designs offered greater primary stability than the 1.5 mm cylindrical design.

**Pan et al (2012)**[60] evaluated the influence of different implant materials on the primary stability of orthodontic mini-implants by measuring the resonance frequency. Twenty-five orthodontic mini-implants with a diameter of 2 mm were used. The first group contained stainless steel mini-implants with two different lengths (10 and 12 mm). The second group included titanium alloy mini-implants with two different lengths (10 and 12 mm). The mini-implants were inserted into artificial bones with a 2-mm-thick cortical layer and 40 or 20 lb/ft$^3$ trabecular bone density at insertion depths of 2, 4, and 6 mm. The resonance frequency of the mini-implants in the artificial bone was detected. Resonance frequency was not influenced by the implant materials titanium alloy or stainless steel. Therefore, the primary stability of a mini-implant is influenced by insertion depth and not by implant material. Insertion depth is extremely important for primary implant stability and is critical for treatment success.

**Singh et al (2012)**[61] analyzed the stress distribution and displacement patterns that develop in a mini implant and its surrounding osseous structures under loading with finite element analysis. Increased stress values were located at the necks of the implants and the surrounding cortical bone.

**Te-Chun Liu et al (2012)**[62] investigated the roles of bone quality, loading conditions, screw effects, and implanted depth on the biomechanics of an orthodontic miniscrew system by using finite element analysis. Both stress and displacement increased with decreasing cortex thickness, whereas cancellous bone density played a minor role in the mechanical response. The screw diameter was the dominant factor for minscrew mechanical responses. Bone stress and screw displacement decreased with increasing screw diameter and cortex thickness, and decreasing exposed length of the screw, force magnitude, and oblique loading direction.

**Cassetta et al (2013)**[63] evaluated alveolar cortical bone thickness and density differences between interradicular sites at different levels from the alveolar crest, and assessed the differences between adolescents (12-18 years of age) and adults (19-50 years of age), males and females, upper and lower arch, anterior and posterior region of jaws and buccal and oral side. Statistically significant differences in alveolar cortical bone thickness and density between age, gender, sites and sides were found. Adults show a thicker alveolar cortical bone than adolescents. Alveolar cortical bone thickness and density were greater in males than in females, in mandible than in maxilla, in the posterior region than the anterior, in oral than buccal side. There is an increase of thickness and density from crest to base of alveolar crest.

**Cho et al (2013)[64]** determined the effects of insertion angle and thread type on the fracture properties of orthodontic mini-implants during insertion. When mini implants contacts artificial root at a critical contact angle, the deformation or fracture of mini-implants can occur at lower maximum insertion torque values than those of penetration.

**Chugh et al (2013)[65]** quantitatively evaluated the bone density at the interradicular areas of the alveolar and basal bones of maxilla and mandible by computed tomography. The highest cortical bone density was observed between the second premolar and first molar at the alveolar bone level and between the first and second molars at the basal bone level in the maxilla. Maxillary tuberosity showed the least bone density. The density of the cortical bone was greater in the mandible than in the maxilla and showed a progressive increase from the incisor to the retromolar area. The basal bone showed a higher density than the alveolar bone. Different qualities of the bone were found in the anatomic regions studied, which confirms the importance of knowledge of site-specific bone tissue density to correlate with various clinical findings.

**Lin et al (2013)**[66] determined the biomechanical effects of exposure lengths of mini implants, the insertion angle and the direction of orthodontic force. Increased exposure lengths resulted in higher bone stresses adjacent to mini implant. The direction of orthodontic force had no effect on cortical bone stress.

**Pithon et al (2013)**[67] evaluated the influence of the length of the mini-implant on its mechanical properties. The insertion torque increased with increasing screw length and increasing cortical bone thickness. Increasing the length of the screw does not increase its mechanical strength, but can efficiently reinforce the initial stability of mini-implants.

**Serra et al (2013)**[68] compared the fracture surface characteristics commercially pure titanium, Ti-6Al-4V alloy, and nano structured, plastically deformed titanium mini-implants by torque test. Torque test results showed significant increase in the maximum torque resistance of nano titanium mini-implants when compared to commercially pure titanium mini-implants, and no statistical difference between Ti-6Al-4V alloy and nano titanium mini-implants. Since nanostructured titanium mini-implants have mechanical properties comparable to titanium alloy mini-implants, and biocompatibility comparable to commercially pure titanium mini-implants, it was suggestive that nano structured titanium could replace Ti-6Al-4V alloy as the material base for mini-implants.

**Tewfiq et al (2013)**[69] evaluated side, gender, age, and regional differences in bone density of the alveolar bone at various orthodontic implant sites. The mean bone density of the alveolar cortical bone was greater in the mandible than in the maxilla and showed a progressive increase from the anterior to the posterior area, while in the maxilla the highest bone density was at the premolars region. The maxillary tuberosity was the region with lowest bone density. Cancellous bone had almost comparable densities between the mandible and the maxilla and its density was less than those of cortical sites. When mini implants are indicated, no gender and side differences affect the success rate regarding bone density; while age and area should be considered when selecting and placing mini implants for orthodontic anchorage.

**Tina et al (2013)**[12] reviewed endeavours to compile the research of bone density in maxilla and mandible. They concluded that Knowledge of low density sites prior to implant placement allows clinician to use longer implant in these areas to improve retention. In areas of high bone density, use of pre-drilling method avoids the breakage of implant. Sufficient irrigation should be done to prevent overheating of bone in that area. Immediate loading of mini-implants is possible because of higher bone density in all the areas of cortical bone.

**Alrbata et al (2014)**[70] determined the appropriate range of cortical bone thickness for supporting an orthodontic mini implant using finite element model. Titanium alloy implant 1.4mm x 7mm was used in cylindrical one models of varying cortical bone thickness and a 2N horizontal force was applied to the mini implant. It was seen that the highest stress occurred near fulcrum where the implant tips and presses into the cortical bone in the direction of the force. Increase in cortical bone thickness resulted in decrease in peak stress but only till a maximum thickness of 2mm.

**Brown et al (2014)**[71] compared detailed mechanical and histologic properties of stainless steel miniscrew implants with identically sized titanium alloy miniscrew implants. All implants were stable at insertion and after 6 weeks. The only significant difference was the higher (9%) insertion torque for stainless steel. No significant differences were found between stainless steel and titanium alloy miniscrew implants in microdamage burden and bone-to-implant contact regardless of loading status. Stainless steel and titanium alloy miniscrew implants provide the same mechanical stability and similar histologic responses, suggesting that both are suitable for immediate orthodontic clinical loads.

**Di Lello et al (2014)**[72] evaluated the insertion and removal torque for mini implants inserted in different inclinations. Insertion torque was lower than the removal one in both insertion degrees. They concluded that 60˚ angulation does not offer any advantages to the primary stability for orthodontic mini implants.

**Fulya Ozdemir et al (2014)**[73] quantitatively evaluated the cortical bone densities of the maxillary and mandibular alveolar processes in adults with different vertical facial types using cone-beam computed tomography. They concluded that patients with the hyperdivergent facial type tend to have less-dense buccal cortical bone in the maxillary and mandibular alveolar processes than those patients with other facial types. Women tend to have denser palatal cortical bone in the alveolar process than men. Clinicians should be aware of the variability in the cortical bone density at mini-implant placement sites and take this into consideration to avoid loss of mini-implants due to insufficient initial stability or breakage during placement.

**Genevive et al (2014)**[74] conducted the study to evaluate the effects of orthodontic mini screw placement angle and structure in terms of length and diameter on stress distribution at the bone mini screw interface. Based on the stress patterns, biomechanical stability of the mini screw is enhanced by a placement angle of 90˚ to the long axis of the tooth.

**Raghavendra et al (2014)**[75] in their review article studied safe zones for miniscrews in orthodontics. The safe zone for mini-implant placement in the anterior region is between the central and lateral incisors in the maxilla and between the lateral incisor and the canine in the mandible at the 6-mm level from the CEJ. At the buccal aspect of the posterior region for all skeletal patterns, the safest zone in the interradicular space of the posterior maxilla was the space between the second premolar and the first molar. In the posterior mandible, the safer zones were located between the first and second premolars and between the first and second molars. Palatally, the optimal site is between the first and second premolars as it has the advantage of the highest cortical thickness.

**Renata de Faria Santos et al (2014)**[76] measured insertion torque, tip mechanical resistance to fracture and transmucosal neck of mini-implants, as well as to analyze surface morphology. Mechanical tests were carried out to measure the insertion torque of MIs in different cortical thicknesses, and tip mechanical resistance to fracture as well as transmucosal neck of mini implants. Surface morphology was assessed by scanning electron microscopy before and after the mechanical tests. All mini-implants tested presented adequate surface morphology. The resistance of mini-implants to fracture safely allows placement in 1 and 2-mm cortical thickness. However, in 3-mm cortical thickness and dense bones, pre-drilling with a bur is recommended before insertion.

**Giselle Lemes Vilani et al (2015)**[77] assessed the influence of cortical thickness and bone density on the insertion torque of a mini implant. Mini implants with lengths of 6mm and 8mm were inserted into synthetic bone blocks. Based on the results of the study they concluded that shorter mini implants have lower primary stability as measured by insertion torque. Greater primary stability is obtained when cortical bone thickness increases. In addition, to minimize fracture risk it is proposed that of the mini implant size should be selected according to the insertion site.

**Kang et al (2015)**[78] investigated the biomechanical properties and bone-implant intersurface response of machined and laser surface-treated stainless steel mini-screw implants. There were no significant differences in fracture resistance and bone-implant contact between the two groups. Laser treatment increased surface roughness without compromising fracture resistance. Despite increasing surface roughness, laser treatment did not improve bone-implant contact. Overall, it appears that medical grade SS has the potential to be substituted for titanium alloy mini-screw implants.

**Gautham et al (2016)**[79] evaluated the stress patterns produced in mini implants and alveolar bone, for various implant dimensions using three dimensional finite element method. The results showed that 1mm diameter mini implants are not safe to be used clinically for orthodontic anchorage. The 1.3 X 6 mm mini implants are recommended for use during anterior segment intrusion and retraction and 1.3 X 8 mm mini implants are recommended for use during molar intrusion.

**Rafael Ribeiro Maya et al (2016)**[80] conducted the ex vivo study to evaluate the effect of vertical placement angle of mini implants on primary stability by analyzing maximum insertion torque. The maximum insertion torque was higher for both mini implant types when they were placed at a 90˚angle (14.40Ncm) compared with those placed at a 60˚ angle. Regardless of the type of mini implants (cylindrical and conical) used, placement at a 90˚ angle resulted in a higher maximum insertion torque.

**Corina et al (2017)**[81] in their study with prosthetic implants showed that the maximum inaccuracy was measured for 11.5mm length implant inserted in the posterior maxilla. The length of the implant, the softer bone in maxilla allowed this deviation during insertion.

# *MATERIALS AND METHODS*

**MATERIALS USED IN THE STUDY:**

1. Sixty three Absoanchor Ti-6Al-4V alloy mini implants by Dentos®, Korea
2. Long handle implant driver, Dentos®, Korea
3. Sixty three Solid rigid polyurethane foam(saw bones)with homogenous density
4. Spirit level
5. Customized stand for implant placement
6. Discovery XR656 digital radiographic machine by G.E.®
7. G.E. Media Viewer software for image analysis
8. Nikon DS 300 DSLR camera

**METHODOLOGY:**

The present study was undertaken at the Department of Orthodontics and Dentofacial Orthopedics, Sri Ramakrishna Dental College and Hospital, Coimbatore

Sixty three Absoanchor self-drilling, mini implants made of Titanium- 6Aluminium-4Vanadium [Ti-6Al-4V] alloy implants from Dentos® Korea, of varying lengths were used for the experiment. Mini implants were conical in shape and the head of the implant was hexagonal with a small hole for passing threads and ligature wires through it. [fig 1]

FIGURE 1: TITANIUM MINI IMPLANT LENGTH 6mm, 8mm, 10mm

Mini implants used in this study are categorized as shown in table below:

| | | |
|---|---|---|
| Titanium mini implants | Length 6mm with diameter 1.3mm | 21 nos |
| | Length 8mm with diameter 1.3mm | 21 nos |
| | Length 10mm with diameter 1.3mm | 21 nos |

Sixty three homogenous Solid rigid polyurethane foam (saw bone) with varying bone density [fig 2] were used in this study to simulate anatomic sites for clinical insertion of mini implants in maxilla and mandible. Following densities were used in the study

FIGURE 2: ARTIFICIAL BONE BLOCKS 20pcf, 30pcf, 40pcf

Artificial bone blocks used in this study are categorized as shown in table below:

| | | |
|---|---|---|
| Homogenous Solid rigid polyurethane foam (saw bone) 2" X 2" X 2" | 20 pcf | 21 nos |
| | 30 pcf | 21 nos |
| | 40 pcf | 21 nos |

Bone blocks were segregated for implant insertion such that one block had one mini screw. Saw bones have the biological properties similar to those of natural bone. Artificial bone, which is composed of synthetic, homogeneous materials, has been shown to be a good substitute for jaw bone.[49]

**Insertion of mini implants:**

A long handle implant driver is used for insertion [fig 3]. A stand was custom fabricated for the study [fig 4, 5]. The implant, implant driver and the bone block were held perpendicular to each other in the custom made stand [fig 6]. The stand was made in such a way to enable adjustment of the bone block and driver in vertical plane. To confirm that the point of insertion of the implant was truly horizontal, a spirit level was placed on the surface of the block before insertion [fig 7]. The mini implant was inserted into the bone block by slow continuous manual insertion. Similarly, all the remaining implants were also inserted one mini implant per bone block. Torque resisting force of the mini implants used in this study were between 1-2 Kgf.cm.

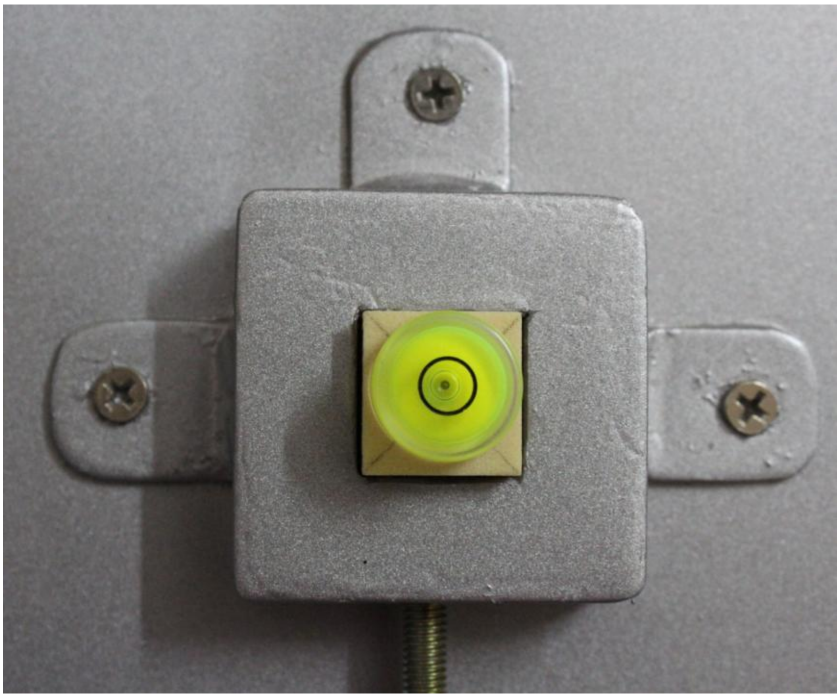
FIGURE 3: LONG HANDLE IMPLANT DRIVER

FIGURE 4: CUSTOM MADE STAND – FRONTAL VIEW

FIGURE 5: CUSTOM MADE STAND - LATERAL VIEW

FIGURE 6: INSERTION OF IMPLANT

FIGURE 7: SPIRIT LEVEL TO CHECK THE BONE SURFACE



**Radiographic imaging of the bone block:**

Once the mini implants were inserted, a digital radiograph was taken of each of the blocks individually. A G.E Discovery XR656 digital radiographic machine [fig 8] with the X-ray source 100cm from the object set at 80kV and 292mAs was used with radiographic exposure time of 1milli second [fig 9]. The bone blocks were placed at the centre of the X-ray beam path. A spirit level was used to ensure that the blocks were not inclined.

FIGURE 8: G.E DISCOVERY XR656 DIGITAL RADIOGRAPHIC MACHINE

FIGURE 9: RADIOGRAPHIC SETTINGS FOR EXPOSURE

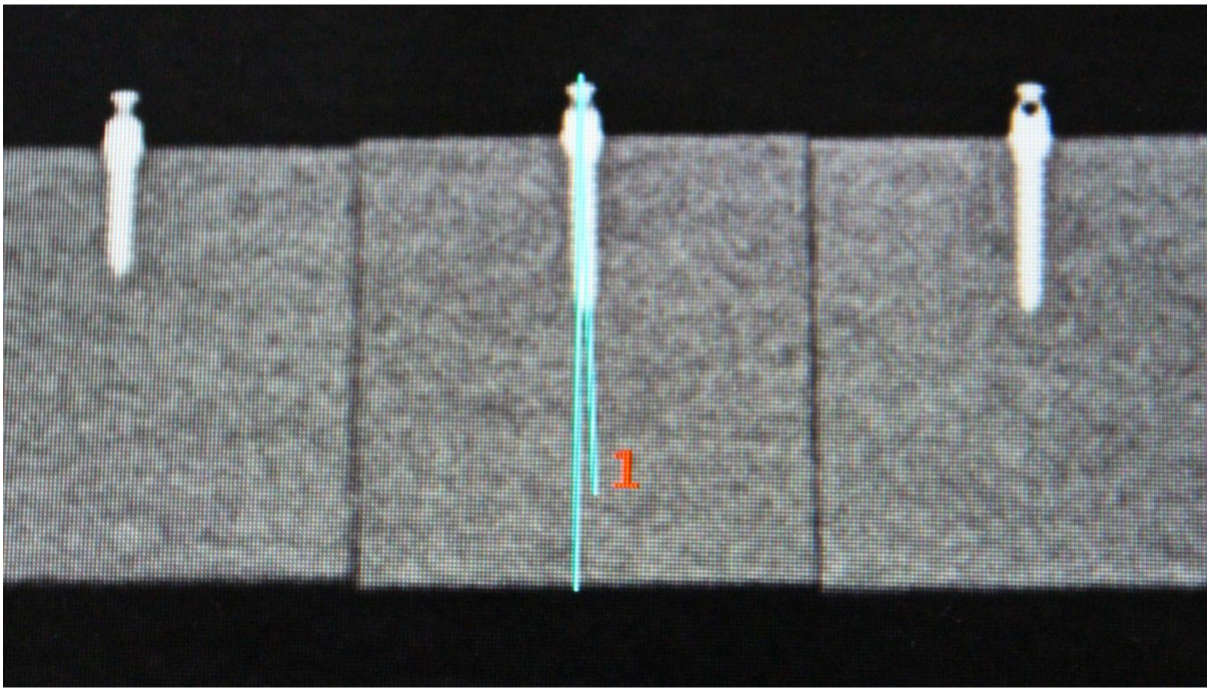
**Image analysis for deflection measurement:**

The radiographic image obtained was adjusted for optimum contrast and magnification prior to obtaining the mini implant deflection values. A pictorial representation of the image analysis is shown in Figure 10. In the image, the red line represents the true horizontal line passing through the centre of point of insertion of the implant. The black line represents the long axis of the mini implant passing through its apex and tip. The angle between the two lines represents the degree of deflection of the mini implant.

FIGURE 10: PICTORIAL REPRESENTATION

Image analysis was done using the G.E. Media Viewer software as the tool for measuring the implant deflection. The long axis of the mini implant was considered as a line joining the apex and the tip of the implant. A true vertical line passing through the centre of point of insertion of the mini implant was used to obtain the degree of deviation of its long axis upon insertion into the bone [fig 11]. The procedure was thus repeated for all the mini implants.

FIGURE 11: ANALYSIS OF RADIOGRAPHIC IMAGE



# *RESULTS*

The study involved the placement of 63 mini implants of 3 different lengths (6mm, 8mm, and 10mm) into three bone densities (20 pcf, 30 pcf, and 40 pcf). Among the 63 mini implants, 60 were placed successfully without mini implant breakage and artificial bone fracture, except for 3 mini implants of dimension 10mm X 1.3mm which fractured at the neck of the implant in the 40pcf block.

## STATISTICAL ANALYSIS

The sample size of 63 was decided for the study using power analysis by GPower3.0.5 software. Descriptive statistics, including the mean value and standard deviation of the deflection value for different implant lengths and bone densities were calculated. This is shown in Table1. Initially the dependent variable is tested (Table 2, Graph 1) for Gaussian (normal) distribution and proved to be normality which is the basic assumption of applying parametric tests (ANOVA family). For significant differences, the data were evaluated using a one-way analysis of variance (ANOVA) test, followed by the post hoc test. SPSS 17.0 was used to find estimates and significance. The mean difference is significant at 0.05 level. Response surface methodology (RSM) explores the relationships between several explanatory variables and one or more response variables. RSM use a sequence of designed experiments to obtain an optimal response. Statistical approaches such as RSM can be employed to maximize the production of a special substance by optimization of operational factors. MiniTab 17.0 was used to fit quadratic regression and to draw RSM from which optimality has been identified.

TABLE 1: DESCRIPTIVE STATISTICS OF OBSERVED DEFLECTION

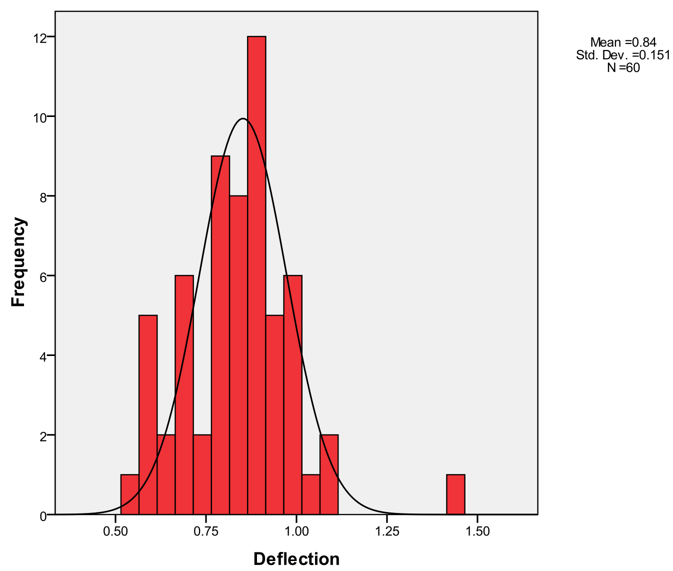
| Implant length | Bone Density | Mean | Std. Deviation | Number of subjects |
|---|---|---|---|---|
| 6mm | 20pcf | .8186 | .03934 | 7 |
| | 30pcf | .8000 | .04509 | 7 |
| | 40pcf | .6143 | .05442 | 7 |
| | Total | .7443 | .10438 | 21 |
| 8mm | 20pcf | .9186 | .04220 | 7 |
| | 30pcf | .8829 | .02628 | 7 |
| | 40pcf | .6671 | .03988 | 7 |
| | Total | .8229 | .11904 | 21 |
| 10mm | 20pcf | 1.0714 | .17468 | 7 |
| | 30pcf | .9257 | .05740 | 7 |
| | 40pcf | 1.2300 | .48111 | 7 |
| | Total | 1.0757 | .30951 | 21 |
| Total | 20pcf | .9362 | .14665 | 21 |
| | 30pcf | .8695 | .06830 | 21 |
| | 40pcf | .8371 | .39028 | 21 |
| | Total | .8810 | .24353 | 63 |

TABLE 2: ONE-SAMPLE KOLMOGOROV-SMIRNOV TEST

| | | Deflection |
|---|---|---|
| N | | 60 |
| Normal Parameters[a,,b] | Mean | .8385 |
| | Std. Deviation | .15117 |
| Most Extreme Differences | Absolute | .095 |
| | Positive | .095 |
| | Negative | -.091 |
| Kolmogorov-Smirnov Z | | .735 |
| Asymp. Sig. (2-tailed) | | .652 |

a. Test distribution is Normal.
b. b. Calculated from data.
c. The KS Test result shows that normality assumption is retained and suggests to apply parametric tests.

GRAPH 1




Histogram of Deflection for identifying the pattern of data and is found to be normal.

All mini implants underwent deflection upon insertion with a maximum mean deflection of 1.1 degrees and a minimum of 0.6 degrees. 'A test of between subjects' effects was done to assess the influence of length and density and also the combined effects of length and density on deflection. The influence of length and density was found to be statistically significant. The influence of combined effects of length and density was found to be non significant (Table 3).

TABLE 3: TWO-WAY ANOVA TEST ON EFFECTS OF LENGHT and DENSITY on DEFLECTION

| Source | Type III Sum of Squares | Df | Mean Square | F | Sig. |
|---|---|---|---|---|---|
| Corrected Model | 1.063[a] | 8 | .133 | 23.787 | .000 |
| Intercept | 40.964 | 1 | 40.964 | 7330.696 | .000 |
| Length | .400 | 2 | .200 | 35.797 | .000 |
| Density | .483 | 2 | .242 | 43.230 | .000 |
| Length * Density | .048 | 4 | .012 | 2.168 | .086 |
| Error | .285 | 51 | .006 | | |
| Total | 43.533 | 60 | | | |
| Corrected Total | 1.348 | 59 | | | |

Mean values of deflection of the implants with varying densities were calculated with their respective standard deviation. The values are shown in Table 4. Mean values of deflection of the implants with varying lengths were calculated with their respective standard deviation. The values are shown in Table 5.

TABLE 4: Descriptive statistics for EFFECT OF DENSITY ON DEFLECTION Implant lengthwise

| Implant Length | | N | Mean | Std. Deviation | Std. Error | 95% Confidence Interval for Mean | | Minimum | Maximum |
|---|---|---|---|---|---|---|---|---|---|
| | | | | | | Lower Bound | Upper Bound | | |
| 6mm | 20pcf | 7 | .8186 | .03934 | .01487 | .7822 | .8550 | .77 | .88 |
| | 30pcf | 7 | .8000 | .04509 | .01704 | .7583 | .8417 | .74 | .88 |
| | 40pcf | 7 | .6143 | .05442 | .02057 | .5640 | .6646 | .54 | .70 |
| | Total | 21 | .7443 | .10438 | .02278 | .6968 | .7918 | .54 | .88 |
| 8mm | 20pcf | 7 | .9186 | .04220 | .01595 | .8795 | .9576 | .86 | .99 |
| | 30pcf | 7 | .8829 | .02628 | .00993 | .8586 | .9072 | .85 | .91 |
| | 40pcf | 7 | .6671 | .03988 | .01507 | .6303 | .7040 | .60 | .71 |
| | Total | 21 | .8229 | .11904 | .02598 | .7687 | .8770 | .60 | .99 |
| 10mm | 20pcf | 7 | 1.0714 | .17468 | .06602 | .9099 | 1.2330 | .91 | 1.44 |
| | 30pcf | 7 | .9257 | .05740 | .02170 | .8726 | .9788 | .84 | .99 |
| | 40pcf | 4 | .8550 | .07853 | .03926 | .7300 | .9800 | .80 | .97 |
| | Total | 18 | .9667 | .14548 | .03429 | .8943 | 1.0390 | .80 | 1.44 |

TABLE 5: Descriptive statistics for EFFECT OF LENGHT ON DEFLECTION bone density wise

| Bone Density | | N | Mean | Std. Deviation | Std. Error | 95% Confidence Interval for Mean | | Minimum | Maximum |
|---|---|---|---|---|---|---|---|---|---|
| | | | | | | Lower Bound | Upper Bound | | |
| 20pcf | 6mm | 7 | .8186 | .03934 | .01487 | .7822 | .8550 | .77 | .88 |
| | 8mm | 7 | .9186 | .04220 | .01595 | .8795 | .9576 | .86 | .99 |
| | 10mm | 7 | 1.0714 | .17468 | .06602 | .9099 | 1.2330 | .91 | 1.44 |
| | Total | 21 | .9362 | .14665 | .03200 | .8694 | 1.0029 | .77 | 1.44 |
| 30pcf | 6mm | 7 | .8000 | .04509 | .01704 | .7583 | .8417 | .74 | .88 |
| | 8mm | 7 | .8829 | .02628 | .00993 | .8586 | .9072 | .85 | .91 |
| | 10mm | 7 | .9257 | .05740 | .02170 | .8726 | .9788 | .84 | .99 |
| | Total | 21 | .8695 | .06830 | .01490 | .8384 | .9006 | .74 | .99 |
| 40pcf | 6mm | 7 | .6143 | .05442 | .02057 | .5640 | .6646 | .54 | .70 |
| | 8mm | 7 | .6671 | .03988 | .01507 | .6303 | .7040 | .60 | .71 |
| | 10mm | 4 | .8550 | .07853 | .03926 | .7300 | .9800 | .80 | .97 |
| | Total | 18 | .6883 | .10804 | .02547 | .6346 | .7421 | .54 | .97 |

PARAMETERS ASSESSED

1) Deflection of mini implant with varying bone density
    a) Deflection of mini implant of dimension 6mmX 1.3mm in 20pcf, 30pcf, 40pcf.
    b) Deflection of mini implant of dimension 8mmX 1.3mm in 20pcf, 30pcf, 40pcf.
    c) Deflection of mini implant of dimension 10mmX 1.3mm in 20pcf, 30pcf, 40pcf.
2) Deflection of mini implant with varying lengths
    a) Deflection of mini implant of lengths 6mm, 8mm, 10mm in 20pcf.
    b) Deflection of mini implant of lengths 6mm, 8mm, 10mm in 30pcf.
    c) Deflection of mini implant of lengths 6mm, 8mm, 10mm in 40pcf.

**DEFLECTION OF MINI IMPLANT WITH VARYING BONE DENSITY**

There is a constant decrease in deflection with increase in density. 20pcf showed maximum deflection followed by 30pcf and the least was seen in 40 pcf . Similar results were obtained for implants of all dimensions. The P values have been showed in Table 8.

For significant differences, the data were evaluated using a one-way analysis of variance (ANOVA) test as shown in Table 6 and Table 7. After evaluating an overall statistically significant difference in group means using one way – ANOVA (Table 6 and 7) POST HOC TESTS are carried out to determine the difference between groups.

TABLE 6:ANOVA TEST FOR VARYING BONE DENSITIES

| Bone Density | | Sum of Squares | Df | Mean Square | F | Sig. |
|---|---|---|---|---|---|---|
| 20pcf | Between Groups | .227 | 2 | .114 | 10.063 | .001 |
| | Within Groups | .203 | 18 | .011 | | |
| | Total | .430 | 20 | | | |
| 30pcf | Between Groups | .057 | 2 | .029 | 14.250 | .000 |
| | Within Groups | .036 | 18 | .002 | | |
| | Total | .093 | 20 | | | |
| 40pcf | Between Groups | .153 | 2 | .076 | 24.987 | .000 |
| | Within Groups | .046 | 15 | .003 | | |
| | Total | .198 | 17 | | | |

TABLE 7: ANOVA TEST FOR VARYING IMPLANT LENGTHS

| Implant Length | | Sum of Squares | Df | Mean Square | F | Sig. |
|---|---|---|---|---|---|---|
| 6mm | Between Groups | .179 | 2 | .089 | 40.959 | .000 |
| | Within Groups | .039 | 18 | .002 | | |
| | Total | .218 | 20 | | | |
| 8mm | Between Groups | .259 | 2 | .130 | 95.666 | .000 |
| | Within Groups | .024 | 18 | .001 | | |
| | Total | .283 | 20 | | | |
| 10mm | Between Groups | .138 | 2 | .069 | 4.691 | .026 |
| | Within Groups | .221 | 15 | .015 | | |
| | Total | .360 | 17 | | | |

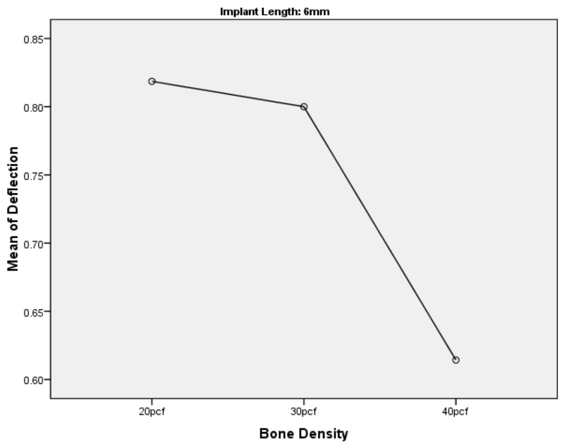
TABLE 8: POST HOC TESTS

| Implant Length | (I) Bone Density | (J) Bone Density | Mean Difference (I-J) | Std. Error | Sig. | 95% Confidence Interval | |
|---|---|---|---|---|---|---|---|
| | | | | | | Lower Bound | Upper Bound |
| 6mm | 20pcf | 30pcf | .01857 | .02496 | 1.000 | -.0473 | .0845 |
| | | 40pcf | .20429* | .02496 | .000 | .1384 | .2702 |
| | 30pcf | 20pcf | -.01857 | .02496 | 1.000 | -.0845 | .0473 |
| | | 40pcf | .18571* | .02496 | .000 | .1198 | .2516 |
| | 40pcf | 20pcf | -.20429* | .02496 | .000 | -.2702 | -.1384 |
| | | 30pcf | -.18571* | .02496 | .000 | -.2516 | -.1198 |
| 8mm | 20pcf | 30pcf | .03571 | .01967 | .258 | -.0162 | .0876 |
| | | 40pcf | .25143* | .01967 | .000 | .1995 | .3033 |
| | 30pcf | 20pcf | -.03571 | .01967 | .258 | -.0876 | .0162 |
| | | 40pcf | .21571* | .01967 | .000 | .1638 | .2676 |
| | 40pcf | 20pcf | -.25143* | .01967 | .000 | -.3033 | -.1995 |
| | | 30pcf | -.21571* | .01967 | .000 | -.2676 | -.1638 |
| 10mm | 20pcf | 30pcf | .14571 | .06493 | .121 | -.0292 | .3206 |
| | | 40pcf | .21643* | .07614 | .037 | .0113 | .4215 |
| | 30pcf | 20pcf | -.14571 | .06493 | .121 | -.3206 | .0292 |
| | | 40pcf | .07071 | .07614 | 1.000 | -.1344 | .2758 |
| | 40pcf | 20pcf | -.21643* | .07614 | .037 | -.4215 | -.0113 |
| | | 30pcf | -.07071 | .07614 | 1.000 | -.2758 | .1344 |

a) Deflection of mini implant of dimension 6mm X 1.3mm in 20pcf, 30pcf, 40pcf.

Statistically significant difference was seen between 20pcf and 40pcf , 30pcf and 40pcf. The mean deflection is represented in Graph 2.

GRAPH 2: DEFLECTION OF MINI IMPLANT OF LENGTH 6mm




b) Deflection of mini implant of dimension 8mmX 1.3mm in 20pcf, 30pcf, 40pcf.

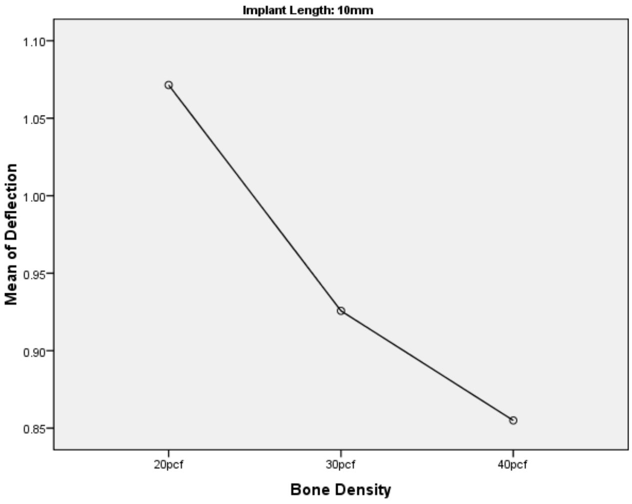
Statistically significant difference was seen between 20pcf and 40pcf , 30pcf and 40pcf. The mean deflection is represented in Graph 3

GRAPH 3: DEFLECTION OF MINI IMPLANT OF LENGTH 8mm

c) Deflection of mini implant of dimension 10mmX 1.3mm in 20pcf, 30pcf, 40pcf.

Statistically significant difference was seen between 20pcf and 40pcf. The mean deflection is represented in Graph 4.

GRAPH 4: DEFLECTION OF MINI IMPLANT OF LENGTH 10mm




**DEFLECTION OF MINI IMPLANT WITH VARYING LENGTHS**

There is a constant increase in deflection with increase in length. 10mm mini implant showed maximum deflection followed by 8mm and the least was seen in 6mm. Similar results were obtained in all the bone densities. The P values have been showed in Table 9.

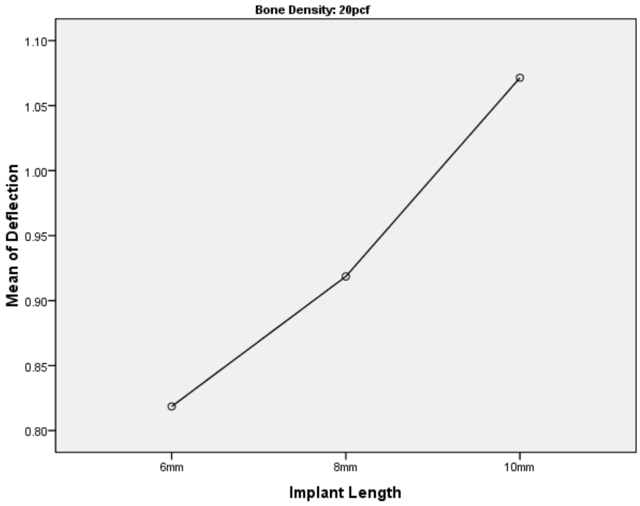
TABLE 9: POST HOC TESTS

| Bone Density | (I) Implant Length | (J) Implant Length | Mean Difference (I-J) | Std. Error | Sig. | 95% Confidence Interval | |
|---|---|---|---|---|---|---|---|
| | | | | | | Lower Bound | Upper Bound |
| 20pcf | 6mm | 8mm | -.10000 | .05677 | .285 | -.2498 | .0498 |
| | | 10mm | -.25286* | .05677 | .001 | -.4027 | -.1030 |
| | 8mm | 6mm | .10000 | .05677 | .285 | -.0498 | .2498 |
| | | 10mm | -.15286* | .05677 | .045 | -.3027 | -.0030 |
| | 10mm | 6mm | .25286* | .05677 | .001 | .1030 | .4027 |
| | | 8mm | .15286* | .05677 | .045 | .0030 | .3027 |
| 30pcf | 6mm | 8mm | -.08286* | .02394 | .008 | -.1460 | -.0197 |
| | | 10mm | -.12571* | .02394 | .000 | -.1889 | -.0625 |
| | 8mm | 6mm | .08286* | .02394 | .008 | .0197 | .1460 |
| | | 10mm | -.04286 | .02394 | .271 | -.1060 | .0203 |
| | 10mm | 6mm | .12571* | .02394 | .000 | .0625 | .1889 |
| | | 8mm | .04286 | .02394 | .271 | -.0203 | .1060 |
| 40pcf | 6mm | 8mm | -.05286 | .02954 | .281 | -.1324 | .0267 |
| | | 10mm | -.24071* | .03464 | .000 | -.3340 | -.1474 |
| | 8mm | 6mm | .05286 | .02954 | .281 | -.0267 | .1324 |
| | | 10mm | -.18786* | .03464 | .000 | -.2812 | -.0945 |
| | 10mm | 6mm | .24071* | .03464 | .000 | .1474 | .3340 |
| | | 8mm | .18786* | .03464 | .000 | .0945 | .2812 |

a) DEFLECTION OF MINI IMPLANT OF LENGTHS 6mm, 8mm, 10mm in 20pcf.

Statistically significant difference was seen between 10mm and 6mm , 10mm and 8mm. The mean deflection is represented in Graph 5.

GRAPH 5: DEFLECTION OF MINI IMPLANT ON 20pcf BONE DENSITY




b) DEFLECTION OF MINI IMPLANT OF LENGTHS 6mm, 8mm, 10mm in 30pcf

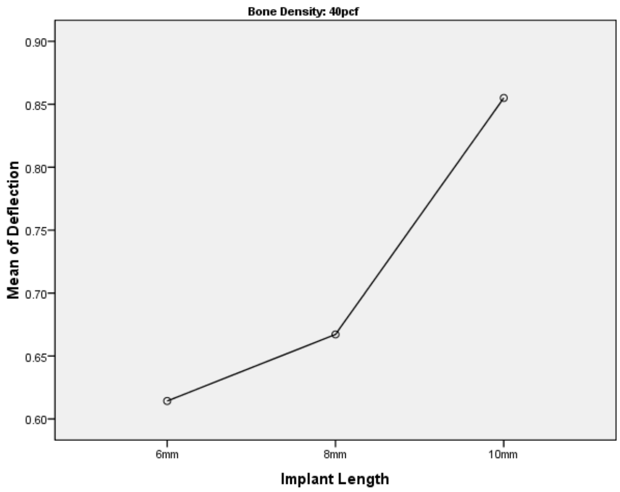
Statistically significant difference was seen between 10mm and 6mm , 8mm and 6mm. The mean deflection is represented in Graph 6.

GRAPH 6: DEFLECTION OF MINI IMPLANT ON 30pcf BONE DENSITY

c) DEFLECTION OF MINI IMPLANT OF LENGTHS 6mm, 8mm, 10mm in 40pcf

Statistically significant difference was seen between 10mm and 6mm , 10mm and 8mm. The mean deflection is represented in Graph 7.

GRAPH 7: DEFLECTION OF MINI IMPLANT ON 40pcf BONE DENSITY




OVERALL DEFLECTION OF MINI IMPLANT ON VARYING DENSITY

The mean deflection of a mini implant that can occur in each bone density irrespective of length of the mini implant are as follows:

- Minimum deflection of 0.8˚ and maximum of 1.0˚ was seen in 20pcf
- Minimum deflection of 0.7˚ and maximum of 0.9˚ was seen in 30pcf
- Minimum deflection of 0.6˚ and maximum of 0.8˚ was seen in 40pcf

The mean values have been showed in Table 10.

TABLE 10: ESTIMATED MARGINAL MEANS OF DEFLECTION ON DENSITY

| Bone Density | Mean | Std. Error | 95% Confidence Interval | |
|---|---|---|---|---|
| | | | Lower Bound | Upper Bound |
| 20pcf | .936 | .038 | .860 | 1.013 |
| 30pcf | .870 | .038 | .793 | .946 |
| 40pcf | .712 | .038 | .676 | .814 |

OVERALL DEFLECTION OF MINI IMPLANTS ON VARYING LENGTH

The mean deflection of mini implants of varying lengths irrespective of the bone density it is inserted are as follows:

- 6mm mini implant deflected to a maximum of 0.8˚ and minimum of 0.6˚
- 8mm mini implant deflected to a maximum of 0.9˚ and minimum of 0.7˚
- 10mm mini implant deflected to a maximum of 1.0˚ and minimum of 0.9˚

The mean values have been showed in Table 11.

TABLE 11: ESTIMATED MARGINAL MEANS OF DEFLECTION ON LENGTH

| Implant length | Mean | Std. Error | 95% Confidence Interval | |
|---|---|---|---|---|
| | | | Lower Bound | Upper Bound |
| 6mm | .744 | .038 | .668 | .821 |
| 8mm | .823 | .038 | .746 | .899 |
| 10mm | 1.076 | .038 | .999 | 1.152 |

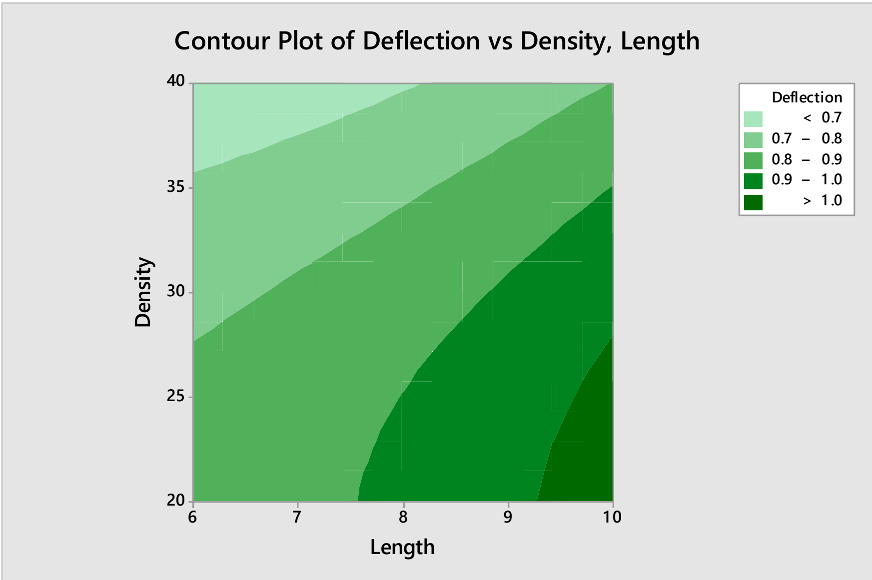
Correlating the lengths and densities maximum and minimum deflection was determined using Response Surface Method analysis. This is shown in Graph 8.

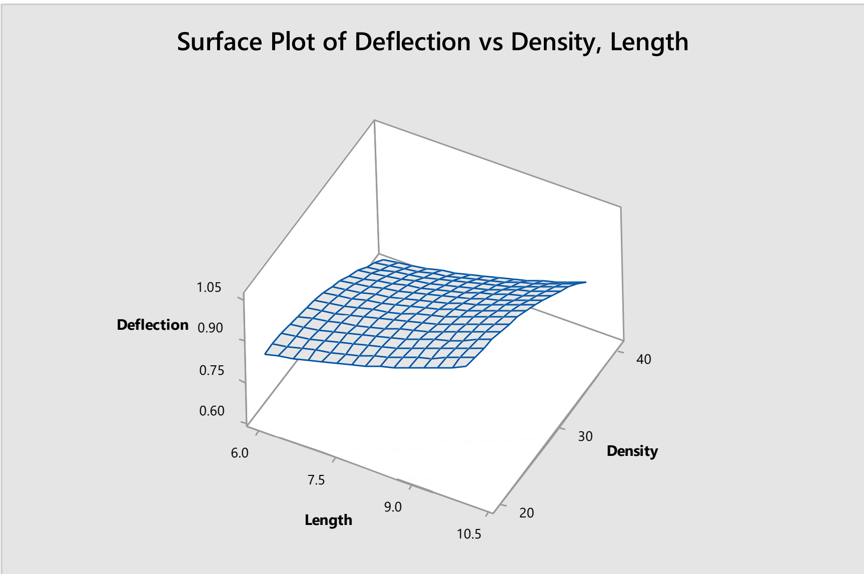
GRAPH 8:

Response Surface Method analysis provided the following quadratic equation to find optimum solution.

The following graphs are generated for the optimization:

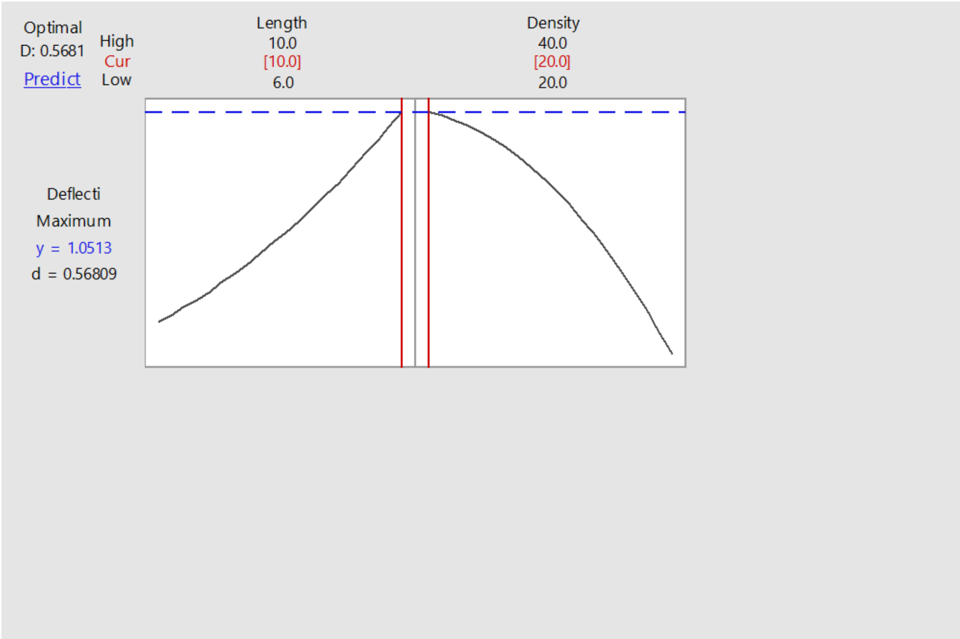
Deflection = 0.593 - 0.0208 Length + 0.0214 Density + 0.00522 Length*Length
- 0.000491 Density*Density - 0.000434 Length*Density



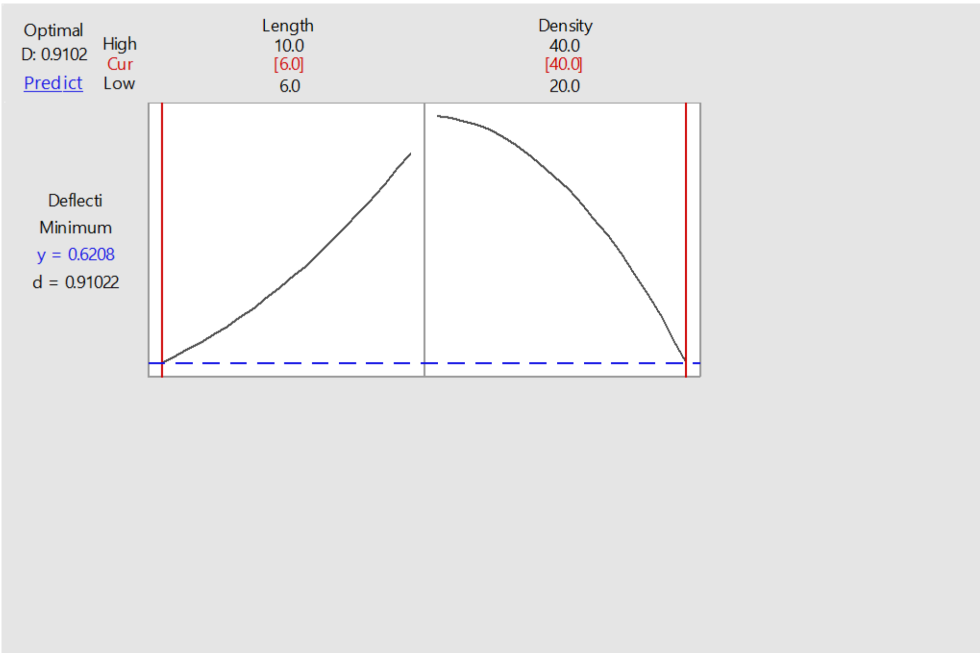
Correlating the lengths and densities the maximum deflection was seen in 10mm mini implant in 20pcf was about 1.05˚. This is represented in Graph 9.

GRAPH 9: MAXIMIZATION OF DEFLECTION




Correlating the lengths and densities the minimum deflection was seen in 6mm mini implant in 40pcf was about 0.6˚. This is represented in Graph 10.

GRAPH 10: MINIMIZATION OF DEFLECTION



# *DISCUSSION*

Temporary anchorage devices have added a whole new dimension to orthodontic treatment, allowing tooth movements to be carried out which were previously thought difficult or impossible[82]. Mini implants have influenced orthodontic treatment plans by providing possible management of complicated discrepancies than those treatable by conventional biomechanics. By the help of mini implants, force can be applied directly to the bone-borne unit. Therefore, mini implants not only eliminated concerns about anchorage – demanding cases, but they also have enabled clinicians to overcome tooth movement in 3 dimensions. Furthermore, adjunctive orthodontic treatments in adults, and treatment for impacted teeth are the other indications of mini implant treatment[83].

Most commonly mini screws are made of stainless steel and commercially pure titanium and its alloys. Titanium screws have the advantage over the stainless steel as they have high bioactivity and more flexibility that improve integration and mechanical fixation. The titanium alloy [Ti-6Al-4V] is used instead of pure titanium because of its superior strength, which allows it to overcome problems such as fractures or distorsions[84]. **Roberts et al** in their study have shown titanium implants developed osseous contact, and continuously loaded implants remained stable. The results indicated that titanium implants provided firm osseous anchorage for orthodontics. Hence Grade 5 titanium (Ti-6Al-4V) implant material was chosen for the present study.

Mini implants are available in different lengths (5 - 12mm) and diameters (1.2 – 2mm) to accommodate placement at different sites in both jaws. Studies have shown in the mandible where the bone is generally denser, a 6 – 8mm length is optimal while in maxilla 8 – 10mm length is preferred. **Deguchi et al**[31] recommended that mini screws less than 1.5mm in diameter could reduce the failure rate in cases where the roots of the adjacent teeth are too close. **Poggio et al**[28] in his study showed that 1.2 – 1.3 mm diameter mini implants were placed safely when less than 3.5mm of interradicular space is available. Thinner implants lead to risks of fracture while thicker implants makes root contact more probable[85]. Hence in this study commonly used dimensions of implants have been used for evaluation and comparison of deflection.

As widely known, osseointegration is not assumed for mini-implants as only the mechanical contact between bone and implant interface is necessary to provide stability. This is the reason of immediate loading ability of mini-implants, since no healing period is awaited. However, osseointegration in mini-implants was found to be present in many studies and these investigators recommend a waiting period prior to force application.[86] Complete osseointegration of mini-implants used in orthodontic therapy is not wanted due to the complications during removal, most of them are manufactured with a smooth surface which impairs the development of bone formation. Despite the amount of osseointegration that may occur it is thought that removal is not difficult since coherence is relatively low as active remodelling and less mineralized bone formation takes place in the bone around the loaded screw part.[87]

The initial stability of mini implant is derived from tight contact with bone and not from osseointegration, the properties of surrounding bone are very important[73]. The anatomy of maxilla and mandible comprise different thickness, density, volume and structures. Human maxilla and mandible vary considerably in volume, density and organization of bone structures as a result of adaptation to the specific conditions of each individual[88]. In 1988, **Misch[17]** proposed the following four bone density groups based on microscopic cortical and trabecular bone characteristics. D1, primarily dense cortical bone; D2, dense to thick porous cortical bone on the crest and coarse trabecular bone; D3, thin porous cortical crest and fine trabecular bone; and D4, minimal to no crestal cortical bone.[89] Generally, D1 bone might be located in the lower anterior or posterior regions but is quite rare. D2 bone is common in the mandible at approximately two thirds of the lower anterior, approximately half of the lower posterior. D3 bone is common in the maxilla at approximately half of the upper posterior, approximately 65% of the upper anterior, and almost half of the lower posterior. D4 bone is found in the maxillary posterior[39]. Suggested implant designs, surgical protocols, healing processes, treatment plans, and progressive loading time spans should be modified for the individual bone density types.

**Choi et al** in his study comparing bone density between maxilla and mandible showed that the mandible had higher values. The density in the maxilla and mandible increased progressively from the midline towards the posterior region. Previous studies[34] had shown differences in the bone densities of the 4 anatomical regions in the mouth were significant, with the anterior mandible yielding a higher mean bone density value, followed by the anterior maxilla, the posterior mandible, and the posterior maxilla. Detailed information on bone density will help us to identify suitable implant sites, thereby improving the success rate of the procedure.

In this study artificial bone block (Sawbones; Pacific Research Laboratories Inc, Wash) were used. In numerous previous studies[49], wood, polyvinyl chloride, and porcine bone were used as the test materials in in vitro tests. In the present study, the artificial bone, the biological properties of which are similar to those of natural bone, is more suitable to determine the deflection of micro-implants. Artificial bone, which is composed of synthetic, homogeneous materials, has been shown to be a good substitute for jaw bone, which varies considerably and so presents difficulties in terms of the mechanical characteristics of the metallic implants. Research had shown that certain densities of rigid polyurethane foams possess mechanical properties that are in the range human bone. The densities chosen correspond to the mean bone density in the posterior and anterior regions of the maxilla and mandible[90].

Studies have shown that the placement angle of the screw can have an effect on its anchor value and the stress transmitted. **Woodall et al**[55] through their finite element analysis and parallel cadaver study clearly demonstrated that compared to 30° and 60°, a 90° insertion angle to the bone surface showed the maximum anchorage advantage. **Jasmine et al**[91] through their finite element analysis study showed that perpendicular insertion of mini implant in bone reduces the stress concentration and offers more stability to orthodontic loading. Machado et al through their 3D finite element analysis study showed that to achieve better biomechanical stability of loaded mini screws in the selected site, placement angle of 90˚ is recommended. Hence the insertion angle was chosen as 90° for the present study.

In the present study 63 mini implants of variable lengths were placed in different densities of bone to evaluate:

1) Role of bone densities on deflections of mini implant with constant length and diameter.
2) Role of lengths of mini implant on deflections with constant diameter and bone density

All mini implants had deflected to varying degrees upon insertion into the bone irrespective of its length and density chosen. Correlating the lengths and densities the maximum deflection was seen in 10mm implant in 20pcf artificial bone and the minimum deflectiom was seen in 6mm implant in 40pcf artificial bone. By keeping length and diameter constant there was progressive decrease in deflection with increase in density of the bone (20pcf, 30pcf, 40pcf). This decreasing tendency of deflections is consistent for all the lengths of the mini implants (6mm, 8mm, 10mm).

In our study maximum deflection was seen in 20pcf rather than 40pcf artificial bone. This outcome might be explained as higher the density of bone greater the initial stability of the implant. In an in vitro study **Abhishek Meher et al**[56] described similar outcomes of deflections. Greater stress and deflection was observed with 1.5mm rather than 2mm cortical bone thickness.

Similar to our study, **Oguz Ozan et al**[53] also showed that lesser bone density values have resulted in greater angular deviation. Less angular deviation values observed in high density bone can be explained by the fact that the dense bone cannot affect the angular deviation regardless of the implant placement method.

Furthermore, by keeping the density of the bone and diameter of implant constant, there was progressive increase in deflection of the implant with increasing length (6mm, 8mm, 10mm). This increasing tendency of deflection as length of mini implant increases is consistent for all the bone densities (20pcf, 30pcf, 40pcf). **Corina et al**[81] in his study with prosthetic implants showed that longer implants deviated during placement. Similar outcome was seen in **Jan D'haese et al**[38] study that shorter implants showed lesser deviation compared with longer implants which is explained by the fact that drilling deeper into the bone with a similar angle of insertion results in a higher apical deviation for a longer implant.

It is known that varying length and diameter can change the strength of the material. The strength of the material is directly proportional to the fourth power of its diameter and inversely proportional to the cube of its length. Hence, the stronger the implant, the greater is its ability to resist deflection. The comparison of results of this in vitro study are in agreement with this principle as the result demonstrates that there is a direct relation of the deflection of the implant on its length[92].

Studies by **Miyajima et al**[93] the following elasticity coefficients were observed for cortical bone, spongy bone and titanium alloy implants: 1.4 x 10 $^{4}$ MPa, 7.9 x 10$^{3}$ MPa and 1.1 x 10$^{4}$ MPa respectively. Most of the stress that occurs during insertion is absorbed by the cortical bone with minimal transfer to the cancellous bone. Thus, the difference in mechanical properties between cortical bone and titanium alloy is a factor responsible for deflection of the mini implant which is exhibited in this study.

In our study also the deflection was observed at the point of entry of the mini implant into bone. **Singh et al**[61] in their finite element study observed deformation of titanium alloy screws but not that of stainless steel screws under similar loading conditions and also that the stress pattern was greatest at the neck of mini implant in both screws. Our study is concurrent with **Liu et al**[94] also who stated that the point of entry of the implant into the cortical bone acts as a pivot for its bending.

During clinical application, the effects of bone density and length of mini implants on deflection, should be considered. Before implants are selected, measurements should be taken to determine the amount of bone that is available for placement. Special attention is required during mini implant placement to reduce the chance of injury to delicate anatomic structures such as blood vessels, nerves, sinus and dental roots.[95] This can be done using investigative tools like radiographs or computed tomographic techniques.

The initial stability of mini implant is derived from the tight contact with bone and not from osseointegration, there by the properties of the bone are very important[73]. The bone density influences the amount of the bone in contact with the implant surface. When implant is driven in thin and less dense bone stress is known to be distributed to the cancellous and cortical bone, whereas stress is centred on the cortical bone where it is thick and dense[12]. Therefore to obtain greater implant surface area longer implants are used in less dense bone. Shorter implants are used in high dense bones as the strength of the implant originates from cortical bone itself. Reducing the length of the mini implant in high density bone increases the success rate by decreasing the deflection of the implant as exhibited in this study.

Longer mini implants when placed in high density bone, insertion torque increases there by chances of fracture or breakage of implant is more. **Tehemar et al**[18] stated that predrilling to reduce the insertion torque will lead to heat generation that result in bone necrosis. Longer mini implants in high density bone will increase the failure rate by increasing the deflection of the implant as exhibited in this study.

Longer mini implants in low density bone showed maximum deflection. In order to increase the surface area and reduce the stress in the bone, length or width of the implant is increased. **Tadas et al**[21] performed a 3- dimensional finite element analysis to evaluate the influence of implant length as well as that of bone quality, on the stress/strain in bone and implant. The results of this study suggest that bone of higher rather than lower density might ensure a better biomechanical environment for implants. Moreover, longer screw-type implants could be a better choice in a jaw with bone of low density.

In the present study three mini implants of length 10mm were fractured at the neck of the implant during insertion in the 40pcf artificial bone. This may be explained due to increased in torsional stress during placement which lead to implant bending and fracture. Mini implants fracture may occur when rotating force was applied over 70% of the torque resisting force of mini implant. Torque resisting force of the mini implants used in this study were between 1-2 Kgf.cm.

It is thought that the placement torque of self-drilling mini-implants can easily become excessive in the thick, mandibular cortical bone, which can cause the mini implant to fracture. When mini implants of different diameters produced by the same manufacturer were compared by **Pithon et al**[96], it was found that their torsional strength values increased as their diameters also increased. This means that insertion torques for installing small diameter mini-implants into high density bones is near the fracture torque, thus requiring more careful attention on the part of the orthodontist. Excessive torque also increases microdamage to cortical bone leading to cracks in the cortical bone immediately adjacent to the implant surface.

Studies had shown the proximity of mini implants to the adjacent tooth root is the major risk factor for their failure. A tooth constantly leaves and enters into the socket during mastication, occlusion, swallowing, among other functions. Such intra- alveolar movements are softened and limited by periodontal collagenous and elastic fibers. When a mini implant is placed too near the periodontal ligament, it causes friction during intra alveolar movements. This will lead to break down of blood vessels, cells and fibers stimulating inflammation and as a consequence, peri-implant bone resorption and mechanical interlocking loss is seen[86]. **Ashish Handa et al** showed that stress in the bone decrease as the distance of the orthodontic mini implant relative to tooth increase.

Understanding the biologic and mechanical aspects of mini implants in orthodontics is an essential prerequisite. Bone density and soft tissue health directly affect implant stability. Longer mini implants can be used in less dense bone as in maxilla, whereas shorter mini implants can be used in high dense bone as in mandible to increase the stability and success rate of implants. Bone density and implant length play a role in deflection of mini implant from its intended path of insertion. The relationship of the insertion pathway with the adjacent structures has to be evaluated in order to reduce the iatrogenic damage.

# *SUMMARY AND CONCLUSION*

Mini screws have revolutionized the field of anchorage in orthodontics. Several studies have been put forth by various authors to enlighten the knowledge of mini implants and its behaviour.

This study was conducted:

(1) To determine the deflection changes of the mini implants from its intended path of insertion.

(2) To evaluate the role of bone densities on deflection.

(3) To evaluate the role of implant lengths on deflection.

A set of sixty three mini implants of varying lengths were inserted into artificial bone blocks of three different bone densities corresponding to the mean bone density of anterior and posterior regions of maxilla and mandible. Once the mini implants were inserted, a digital radiograph was taken of each of the blocks individually. Image analysis was done using the G.E. Media Viewer software as the tool for measuring the implant deflection.

The results of the study showed:

(1) All mini implants had undergone deflection of varying degrees on insertion.

(2) Deflection of mini implant decreases as the density of bone increases.

(3) Deflection of mini implant increases as the length of implant increases

Correlating the lengths and densities the maximum deflection was seen in 10mm mini implant in 20pcf bone block was about 1.05˚ and the minimum deflection was seen in 6mm mini implant in 40pcf bone block was about 0.6˚.

In conclusion, the bone density influences the amount of the bone in contact with the implant surface. When implant is driven in thin and less dense bone stress is known to be distributed to the cancellous and cortical bone, whereas stress is centred on the cortical bone where it is thick and dense. Understanding the biologic and mechanical aspects of mini implants in orthodontics is an essential prerequisite. Bone density and soft tissue health directly affect implant stability. Knowledge of bone density in the maxilla and mandible will correlate many of the clinical findings as well as allow the clinician to plan the anchorage strategies and placement of implants with necessary precautions accordingly. Longer mini implants can be used in less dense bone as in maxilla, whereas shorter mini implants can be used in high dense bone as in mandible to increase the stability and success rate of implants. Bone density and implant length play a role in deflection of mini implant from its intended path of insertion. There by evaluation of the relationship of the insertion pathway with the adjacent structures is needed to reduce the iatrogenic damage.

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