EVALUATION OF DEFLECTION CHARACTERISTICS OF MINI IMPLANTS FOLLOWING PLACEMENT- AN EX-VIVO STUDY

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CERTIFICATE

This is to certify that this dissertation titled **"EVALUATION OF DEFLECTION CHARACTERISTICS OF MINI IMPLANTS FOLLWING PLACEMENT- AN EX-VIVO STUDY"** is a bonafide work done by **Dr. ANISHA N. PRASAD** under my guidance during her post graduate study period between 2012-2015.

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

The movement of teeth during orthodontic therapy occurs primarily through the application of forces. In order for these forces to cause changes in tooth position, adequate support must be available from which these forces can be applied. Hence, ever since its origin, the field of orthodontics and dentofacial orthopaedics has focussed on the importance of anchorage and the consequences of its loss. This anchorage can be derived from other teeth, extraoral sources or from skeletal structures. But these forces also act reciprocally on the anchoring structures thereby causing undesirable movement of such structures.

Hence, the concept of skeletal anchorage was introduced to offer capabilities for treatment unavailable previously. Various methods for obtaining skeletal anchorage like endosseous implants, bone screws used for fixation in surgery were tried initially and now mini implants especially manufactured for orthodontic anchorage are readily available. This helps forces to be applied to produce tooth movement in any direction without detrimental reciprocal forces.

The possibility of skeletal anchorage was explored by **Creekmore and Eklund (1983)** [1] by using a Vitallium (Cobalt-Chromium) screw for intruding anterior teeth in the maxilla. **Kanomi (1997)**[2] clinically demonstrated the first successful use of orthodontic mini implants with a diameter of 1.2mm and 6mm in length for mandibular incisor intrusion with no root resorption or periodontal pathologies.

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

Most present mini implants are fabricated from either commercially pure titanium [cpTi / Ti grade 4] or titanium alloy [Ti-6Al-4V / Ti grade 5]. They have excellent corrosion resistance and are highly biocompatible. A protective surface oxide layer develops when it comes into contact with oxygen or tissue fluids and even if it is lost, it is regenerated within milliseconds due to its affinity towards oxygen and nitrogen. Titanium grade 4 has tensile strength of 550 MPa whereas Titanium grade 5 has a tensile strength of 910 MPa.

Both have a similar Young's modulus of 100-110 GPa .Titanium alloy offers greater strength, more favourable surface condition, stress-strain behaviour and wear resistance.

Depending on the method of insertion two types of mini implants are available: self-drilling and self-tapping. Self drilling mini implants have a cutting tip and can be inserted directly into the bone. Self-tapping implants need implant site preparation with the use of a drill to make a pilot hole following which the implant is then inserted.Self drilling implants offer numerous advantages like easy insertion technique, increased tactile sensation and no additional armamentarium is necessary.

Immediately after insertion, the retention of any mini-screw is purely mechanical in nature and is achieved through a combination of displacement and compression of the adjacent bone. This process is known as primary stability. It is independent of the implant material but is highly dependent on the design of the screw, bone thickness and insertion technique. Primary stability relies on the mechanical interlocking of the threads of the screw with mainly the cortical bone, hence greater the bone quantity, the better the primary stability will be. The minimum preferred cortical bone thickness for mini implants to be stable is greater than 1mm[3].

The thickness of the cortical bone in the maxilla is generally lesser than in the mandible.

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**[4] 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**[5] described a positive correlation between mini-implant insertion torque and bone density values, and concluded that methods used to measure torque during mini-implant placement should be used routinely.

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 length and diameter of the implant. Due to its size, despite the use of titanium, the flexural strength of the mini implant is decreased. Consequently, the maximum force required to cause permanent deformation also decreases. 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**[6], 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 the aim of this in vitro study was to radiographically evaluate the deflection of titanium alloy self-drilling mini implants from the intended path that occurs during placement and also to compare effect of various lengths and diameters on the quantum of deflection.

REVIEW OF LITERATURE

Creekmore TD and Eklund MK[1] (1983) attempted to determine if a metal implant could withstand a constant force over a long period of time of adequate magnitude to depress an entire anterior maxillary dentition without becoming loose, infected, painful, or pathologic. A surgical Vitallium bone screw was inserted just below anterior nasal spine and they achieved 6mm intrusion over a one year period.

Park HS[7] (1999) treated a case with skeletal cortical anchorage using titanium microscrew implants. During six months of orthodontic force application from skeletal cortical anchorage, the author could get 4mm bodily retraction and intrusion of upper anterior teeth. The titanium microscrew implants and remained firm and stable throughout treatment. These results indicated that skeletal cortical anchorage might be a very good option.

Favero et al[8] (2002) reviewed implant related studies published between the years 1970-2002 and found that the maximum load that can be applied is influenced by the amount of implant-bone contact which is in turn affected by the screw length, diameter and shape. An inverse relationship existed between diameter and length. Finally the selection of screw is also dependent on the availability of bone in that particular region.

Fanuscu[9] (2003) quantified the elastic properties of maxillary and mandibular bone at the lamellar level and compared these properties among varying sites. Moderately resorbed edentulous maxilla and mandible from a human cadaver were analyzed. The mean overall elastic moduli were 14. 76 GPa for cortical bone and 15.37 Gpa for cancellous bone. They concluded that the overall values for modulus of elasticity and hardness varied mildly with the possibility of site-specific differences.

Miyawaki et al[10] (2003) examined the success rates and factors associated with the stability of titanium screws placed into the buccal alveolar bone of the posterior region. Three kinds of titanium screws 1.0x6mm, 1.5x11mm and 2.3x14mm were evaluated. The screws were placed into the buccal alveolar bone through attached gingiva in the second premolar to second molar region of the maxilla or the mandible. They concluded that the diameter of a screw of 1.0 mm or less, inflammation of the peri-implant tissue, and a high mandibular plane angle (i.e., thin cortical bone), were associated with failure of the titanium screw.

Kanie et al[11] (2004) compared the mechanical properties of two prosthetic mini implants of sizes 1.8mm x 21mm and 1.8mm x 22mm. They determined flexural strength, elemental composition, surface characteristics. They found that maximum strength and proportional limit varied significantly.

Liou et al[12] (2004) inserted miniscrews on the maxillary zygomatic buttress as a direct anchorage for en masse anterior retraction using nickel-titanium closed-coil springs. On radiographic evaluation, they found that the miniscrews tipped forward by 0.4 mm at the screw head.. They concluded that miniscrews are a stable anchorage but do not remain absolutely stationary throughout orthodontic loading.

Schnelle et al[13] (2004) did a panoramic radiographic evaluation of the availability of bone for mini implant placement in 30 patients. 14 inter-radicular sites were measured using a digital caliper and the existence of 3-4mm interradicular bone was considered as the minimum requirement. They found that this minimum requirement was present only in posterior regions mesial and distal to maxillary and mandibular first molar. They concluded that the clinician needs to be aware that it may not be possible to place mini screws always in attached gingival due to lack of sufficient inter-radicular bone.

Costa et al[14] (2005) evaluated hard and soft tissue thickness for implant placement. The bone depth was quantified by volumetric computed tomography (VCT). The mucosal depth was quantified by a needle with a rubber stop. The results indicate that bone thickness will allow mini screws 10 mm in length only in the symphysis, retromolar, and palatal premaxillary regions. Screws 6 to 8 mm in length can be placed in the incisive fossa, in the upper and lower canine fossae. When placing temporary anchorage devices in mobile alveolar mucosa, the results suggest that a transmucosal attachment may be required to traverse the thickness of the soft tissue.

Melsen[15] (2005) in an overview article on mini implants gave a brief description of the evolution of skeletal anchorage and its indications. She also mentioned a few properties about the most commonly used anchorage systems along with insertion sites and techniques. Problems associated with mini screws such as patient related, operator related and screw related were also mentioned.

Deguchi et al[16] (2006) used computed tomgraphic scans from 10 adults to measure the cortical bone thickness of various potential mini screw placement sites in the maxilla and mandible. They took measurements at at the occlusal level (3-4 mm apical to the gingival margin) and at the apical level (6-7 mm apical to the gingival margin). Significantly less cortical bone was seen in the maxillary buccal region at the occlusal level distal to the second molar when compared with other areas in the maxilla. Additionally, maxillary cortical bone was significantly thicker on the lingual side of the second molar site when compared to the buccal side. In the mandible, there was significantly more cortical bone mesial and distal to the second molar when compared with the maxilla.

Morais et al[17] (2006) analysed immediately loaded commercially pure titanium and titanium alloy implants fixation and gauged the vanadium ion release during the healing process in the tibiae of rabbits. A stress analysis was done to predict the torque at which both types deform plastically and the shear strength at the interface. It was found that removal torque of commercially pure titanium was close to its yield stress and that the concentration of vanadium from the titanium alloy mini implants did not reach toxic levels in the animals.

Park et al[18] (2006) evaluated the factors affecting the clinical success of screw implants used as orthodontic anchorage in eighty seven patients. A total of 227 self tapping mini implants of diameter 1.2mm and lengths 4-8mm and 10mm and diameter 2mm with lengths 10-15mm were used. The overall success rate was 91.6%. They found that screws placed on the right side of the jaw and in the mandible had a higher failure rate.

Poggio et al[3] (2006) did a study to determine the safe zones for mini implant placement using computed tomography. In the maxilla, the greatest amount of bone was seen on the palatal side between second premolar and first molar and least was seen in the tuberosity region. In the mandible, the greatest thickness of bone was noted between first and second premolars and the least bone was observed between first premolar and canine.

Cornelis et al[19] (2007) did a systemic review of the experimental usage of temporary skeletal anchorage devices from electronic databases. Diameters and lengths of screws used varied between 1-2.2mm and 4-10mm respectively. The healing times reported ranged from 0-12weeks, amount of force applied varied from 25-500g. Direct bone-screw contact was reported to be 10-58%. They concluded that long term bone-implant adaptation has not yet been well categorized and future research needs to target specific issues with well controlled experimental models.

Kuroda et al[6] (2007) evaluated root proximity as a risk factor for failure of mini screws in 116 patients with a total of 216 titanium screws. Each screw was classified according to its proximity to the adjacent root into three categories. They found that the average success rate for the screws was 80% and maxillary screws had greater success than mandibular screws. The concluded that proximity of screw to the root was indeed a major risk factor and also that this tendency was more obvious in the mandible.

Motoyoshi et al[20] (2007) determined the effect of cortical bone thickness and implant placement torque on the success of mini implants. After computerized tomography examination, mini implants 1.6x8mm were inserted and orthodontic force was applied for 6 months. They found the success rate higher in implants with insertion torque 8-10Ncm and also when the bone had a minimal cortical bone thickness of 1mm.

Song et al[21] (2007) evaluated the effect of cortical bone thickness on the maximum insertion and removal torque of different types of self-drilling mini-screws and determined if torque depends on the screw design. Titanium alloy cylindrical and tapered screws of dimensions 1.5x6mm, 1.6x6mm and 1.6x7mm were inserted into artificial bone blocks. The mini-screw tip was placed perpendicular to the artificial bone sample, and was inserted to the end of the screw thread by rotating the torque tester and insertion torque was noted. Removal torque was similarly noted by rotating the driver in the opposite direction. They found that a tapered form, with the outer diameter increasing, is the design that increases the torque the most.

Elias et al[22] (2008) enumerated some of the materials used for biomedical applications with emphasis on the importance of titanium and its alloys and their use in the field of implant dentistry. They stated that orthodontic implants are mainly composed of titanium alloy instead of pure titanium due to the former's superior strength. However its corrosion resistance is lower allowing for metal ion release. They proposed the use of ultrafine grain titanium due to its superior biocompatibility and higher mechanical properties than commercially pure titanium.

Gonzalez[23] (2008) did an in vitro study on the cortical bone thickness of maxilla and mandible using computed tomographic scans from seventy eight skulls.

The cemento-enamel junction interproximally of each tooth was determined as the reference point for measurements. In the maxilla the mean cortical bone thickness was below 1 mm at the 6 mm location while at 9 mm and 12 mm locations the mean cortical bone thickness varied from 0.78 mm to 1.31 mm. There was a wide range of measurements in the mandible from 0.62 mm to 3.65 mm with the majority of the means over 1mm. The mandible overall had more thickness of the cortical bone than the maxilla.

Lim et al[24] (2008) determined the variation in the insertion torque of cylindrical and tapered orthodontic mini screws according to the screw length, diameter. Cylindrical and tapered screws of various diameters and length were inserted into artificial bone blocks and the torque was measured. In both types of screws, the maximum insertion torque increased with increasing diameter and length of the orthodontic miniscrews as well as increasing cortical bone thickness. A significant increase in insertion torque was seen mainly in the taper type miniscrew. They concluded that, increase in screw diameter can efficiently reinforce the initial stability of miniscrews, but the proximity of the root at the implant site should be considered.

Park et al[25] (2008) quantitatively evaluated density of the alveolar and basal bones of the maxilla and the mandible using computed tomographic scans of twenty three men and forty women. Cortical density of the maxillary alveolar bone was between 810 and 940 HU, except for the tuberosity, which was approximately 443 HU in the buccal and 615 HU in the palatal alveolar bone. Cortical density of the mandible was between 810 and 1580 HU at the alveolar bone and between 1320 and

1560 HU at the basal bone. Cortical bone of the mandible was denser than that of the maxilla, whereas cancellous bone had similar densities between the mandible and the maxilla. Basal bone generally showed higher density that alveolar bone. They stated that these data could provide valuable information when selecting sites and choosing placement methods for miniscrews.

Prates de Nova et al[26] (2008) evaluated mini-implants of different sizes for insertion, removal, fracture torque, shear tension, and type of fracture. Twenty commercial self-drilling mini-implants of 1.6mm diameter and 7mm and 8mm lengths with and without necks were inserted into bovine tibias. To ensure mini-implant insertion into cortical bone alone, a hole was drilled in the center of the bone specimen. The mini-implant was inserted following perforation with the insertion key attached to the handpiece with a torquimeter. The mini-implants were removed with the same hand piece using the reverse rotation option microscopy. Mini implants with neck showed the greatest insertion torque. All mini implants showed removal torques lesser than insertion torques and experienced ductile fracture.

Pithon et al[27] (2008) designed a study to assess the deformation and fracture of orthodontic mini implants of different commercial brands by submitting them to loads perpendicularly applied along their lengths. Seventy five mini-implants were inserted perpendicularly into swine cortical bones. The different forces required to fracture mini-implants after undergoing 0.5mm, 1mm, 1.5mm and 2mm deformation were assessed. All mini-implants tested in this study proved adequate for use in orthodontic anchorage. Mini-implant shape was directly related to the flexural strength afforded by these devices when perpendicular forces were applied along their axes.

Pithon et al[28] (2008) assessed the maximum torsional strength of orthodontic mini-implants of different diameters. Eighteen titanium alloy mini-implants measuring 10 mm in length and diameters of 1.2mm, 1.4mm, and 1.6 mm were used. Mini-implants with greater diameter had the highest mean torsional values, whereas those with smaller diameter had the lowest ones. The torsional strength analysis for mini-implants has showed that fracture torque is relatively high compared to that used for mini-implants inserted in osseous substrates. Furthermore, the use of greater-diameter mini-implants provides safer conditions regarding fracture.

Salmoria et al[29] (2008) did a study to evaluate insertion torque of mini-implants for orthodontic anchorage, to compare their axial pull-out strength, to determine initial and peri-implant cortical bone thickness, and to analyze the correlations among these variables. Sixty self tapping titanium alloy screws 1.6 mm diameter and 6.0 mm length were placed in the mandibles of 10 dogs. Peak insertion torque values were recorded. Cortical bone thickness was measured after removal of the mini implants.

Authors concluded that pull-out strength is greater immediately after placement of mini screws, cortical bone thickness decreases because of bone resorption, and insertional torque is not an efficient method for predicting the retention of mini implants.

Wilmes et al[30] (2008) analyzed the impact of implant design of six commercially available mini implants on primary stability using porcine rib. They observed that conical implants had higher primary stability than cylindrical ones. They concluded that the diameter and design of mini implant thread have an impact on primary stability. Depending on the region of insertion and local bone quality, the choice of mini implant is crucial.

Baumgaertel et al[31] (2009) investigated the buccal cortical bone thickness of every interdental area as an aid in planning mini-implant placement using cone-beam computed tomography scans of 30 dry skulls. They found that buccal cortical bone thickness was greater in the mandible than in the maxilla. The thickness increased with increasing distance from the alveolar crest in the mandible and in the maxillary anterior sextant, it behaved differently in the maxillary buccal sextants; it was thinnest at the 4-mm level. They proposed that future studies are needed to determine the exact relationship between cortical bone thickness, the method of implant site preparation, and success rates.

Chen et al[32] (2009) published a review article on the factors critical to ensure mini implant success. Most mini implants were found to withstand 100 to 200g of horizontal early or immediate loading successfully and direct orthodontic loading offered shorter treatment time. A volumetric tomographic image analysis for the maxilla and the mandible suggested that safe zones for placement of mini-screws was a maximum diameter of 1.2 to 1.3 mm, and implants with a diameter of 2 mm cannot be considered safe for placement in the posterior interradicular spaces of the maxilla, except between the first molar and the second premolar on the palatal side, and between the canine and the first premolar. Mini implants with a diameter less than 1.5 mm were intended for tooth-bearing areas, particularly in the inter-radicular area.

Hu et al[33] (2009) elucidated the relationship between dental roots and the surrounding tissues to prevent complications after mini screw placement.200 cross sections of human maxillae and mandibles from 20 individual bones were obtained. The inter-root distance, total bucco-lingual bone width, cortical bone thickness and mucosal thickness was measured. It was seen that for all the above mentioned parameters the values increased from anterior to the posterior region and also from the cervical region to the tooth apex.

Luzi et al[34] (2009) provided an overview regarding the guidelines for success in mini implant placement. 137 patients were treated using self-drilling mini implants of lengths either 9.6mm or 11.6mm and diameters either 1.5mm or 2mm. All mini screws were immediately loaded to achieve various dental movements. They recorded a 9% failure rate and divided the possible causes into factors related to clinician, the patient and the screw itself. They concluded that large multicenter studies are needed to gain information on skeletal anchorage and reduce failure rates.

Mayer et al[35] (2009) evaluated the implant angulations and alignments with neighbouring teeth or implants and compared them to angulations and alignments measured with pre-insertion gauges. The mesio-distal angular relationship between gauge and implant, gauge and gauge, gauge and teeth, implant and teeth, and gauge and inferior border of mandible were measured

Results revealed that maxillary implants were more divergent than mandibular implants. Implants placed away from the clinician's side had smaller guage-implant discrepancies than those on the ipsilateral side. Implants placed in the anterior region diverging more than those in the premolar and molar regions and those placed adjacent to teeth had greater divergence than implants in the edentulous region. They proposed that placement of implants without the use of a guide can result in adequate dental implant angular relationships. However, an implant's location and the presence of adjacent teeth can affect the angular relationships.

Reyenders et al[36] (2009) reviewed the literature to quantify success and complications with the use of mini-implants for orthodontic anchorage, and to analyze factors associated with success or failure. Adverse effects of miniscrews included biologic damage, inflammation, and pain and discomfort. Only a few articles reported negative outcomes. They concluded that interpretation of findings was conditioned by lack of clarity and poor methodology of most studies.

Cha et al[37] (2010) determined the effect of bone mineral density, cortical bone thickness, screw position, and screw design on the stability of mini screws. Ninety-six miniscrews of both cylindrical and tapered types were placed in 6 beagle dogs manually at 70 to 90° to gingival surface. In all miniscrews, a force of 250 to 300 g was applied.Placement and removal torque values were significantly higher in the mandible compared with the maxilla and was affected by screw position, screw type, and density of cortical bone. Tapered mini screws had higher placement torque than did the cylindrical type. However, the removal torque was similar in both groups.

The authors concluded that bone mineral density of cortical bone, screw type, and screw position significantly influence the primary stability of mini screws.

Crismani et al[38] (2010) did a systematic review of effects related to patient, screw, surgery, and loading on the stability of miniscrews. The mean overall success rate was 83.8%-67.4%. Patient sex showed no significant difference. 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.

Laurito et al[39] (2010) determined the feasibility of temperature recording during implant site preparation using bovine bone using a fluoroptic thermometer. They found the method to be appropriate for real time temperature data recording and concluded that further studies are needed to define standardized procedures.

Lee et al[40] (2010) investigated the effects of the diameter and shape of orthodontic mini-implants on micro damage to the cortical bone during implant placement.Twenty eight self drilling screws of length 6mm and diameters 1.5 and 2mm of cylindrical and tapered shapes were placed in the tibias of seven rabbits. Maximum insertion torque was measured and immediately after placement of the mini implants, the bone with screws was harvested. Cortical bone thickness was measured by using micro computed tomography, and histomorphometric analyses of the cracks were performed.

There was a significant increase in maximum insertion torque correlated to increased diameter and taper. Similarly, there was a significant increase in the number of cracks with increased diameter and tapering. They concluded that further studies about the effect of micro damage on bone remodelling and stability of the mini screws are needed.

Motoyoshi et al[41] (2010) measured the placement and removal torques of mini-implants placed in buccal posterior alveolar bone of fifty seven patients and assessed the relationships among placement and removal torques, placement period, age, sex, and cortical bone thickness. Computerized tomography was used to measure the cortical bone thickness. A torque screwdriver was used to measure the peak torque values. The mean placement and removal torques were 8 and 4 N cm, respectively. A torque of 4 N cm suggested sufficient anchorage capability for mini-implants. No significant correlation between placement and removal torques was found. Placement torque was significantly related to age and cortical bone thickness in the maxilla, whereas removal torque was not significantly related to placement period, age, sex, or cortical bone thickness.

Qamaruddin et al[42] (2010) published a literature review regarding the factors that contribute to the failure of orthodontic mini implants. The various factors proposed by them included improper length and diameter and a weak neck of the mini implant. The maximum load tolerated by the mini implant was 50N-450N. Anatomical constraints for implant placement needed to be kept in mind to avoid inadvertent injury to root, or perforation of sinus. With regards to operator related causes, implants needed to be inserted without wiggling or overheating of the bone and a self drilling mini implant was preferable for a flapless procedure.

Barros et al[43] (2011) evaluated the effect of mini implant diameter on fracture risk and self drilling efficiency. 405 titanium alloy mini-implants with 9 diameters from 1.2 to 2.0 mm of length 8mm were used. Ten mini-implants of each diameter were placed in artificial bone, and twenty five were placed in pig iliac bone to evaluate placement torque and axial placement load. Increases in diameters significantly affected the placement and fracture torque and reduced the fracture risk for each 0.1-mm change in diameter. The diameter had more influence on fracture risk than on drill-free placement efficacy. Placement torque and placement load showed antagonistic behaviour during drill-free placement characterized by progressive torque increases and gradual axial load reductions.

Chatzigianni et al[44] (2011) investigated the influence of implant diameter and length on primary stability by measuring the deflection during high and low force application in vitro. A total of 62 mini implants of length 9mm and diameter 1.5mm and length 7mm and diameter either 1.5mm or 2mm were inserted into bovine rib segments fixed in autopolymerizing resin. The cortical bone thickness was measured to be around 2mm clinically. Bovine ribs have the same architectural pattern as the mandible with clearly defined cortical and cancellous bone. They are the material of choice for studies focussing on maxillofacial implantation.

At low force levels, no significant difference in displacement was noted between the various implants. At higher force levels (2.5N), the 9mm long and 2mm wide mini implants were displaced less than the 7mm long and 1.5mm wide ones. The results showed that implant length and diameter were significant influencing factors on stability when force level exceeded 1N.

Chatzigianni et al[45] (2011) compared numerical simulation data derived from finite element analysis to experimental data on mini implant loading. The purpose was to investigate the effect of implant length, diameter and method of insertion on the primary stability of Aarhus and LOMAS mini implants, each of 1.5mm diameter and lengths 7mm and 9mm. The implants were inserted in bone either perpendicular or at 45° to the surface. A force of 0.5N was applied to the neck of the mini implant, parallel to the bone surface using closed nickel-titanium springs and a similar condition was simulated in the finite element method. Both the results showed that at low force levels, there was no statistical significance in implant displacement according to length, diameter and insertion angle. Rotation of implant was influenced by implant type- LOMAS mini implants rotated more than the Aarhus ones.

Farnsworth et al[46] (2011) did a CBCT evaluation of cortical bone thickness at common implant placement sites in 26 adults and 26 adolescents. Cortical thickness was measured as the shortest distance between the endosteal and periosteal surface at each site. Results showed no difference in thickness based on sex of the patient. Adult cortical bone showed increased thickness compared to adolescent bone in all areas except infrazygomatic crest, mandibular buccal and posterior palate. In both groups, cortical bone appeared thicker in the posterior than anterior region. Anterior paramedian palatal bone was significantly thicker than bone located more posteriorly. They concluded that the mandibular buccal and infrazygomatic crest regions had the thickest cortical bone and the differences between the other areas were small. Also adults had thicker cortical bone when compared to adolescents.

Lemieux et al[47] (2011) used computed tomography imaging to measure placement pattern, bone density, and thickness surrounding sixty mini implants. 1.8mm diameter implants of lengths 6, 8, and 10mm were placed in the maxilla and mandible of 5 human cadavers. Results showed that shorter mini implants tended to have lesser penetration into buccal cortical bone compared to longer implants but they also posed lesser chances of damages to surrounding structures and bicortical perforations. The most important factors in determining maximum mechanical anchorage were found to be bone density placement depth, and mini implant length.

McManus et al[4] (2011) explored the relationship between maximum placement torque during miniscrew placement and miniscrew resistance to movement under load. Ninety-six titanium screws were placed into 24 hemi-maxillae and 24 hemi-mandibles from cadavers between the first and second premolars by using a digital torque screwdriver. All screws were subjected to a force parallel to the occlusal plane, pulling mesially until the miniscrews were displaced by 0.6 mm.

Mean buccal cortical bone thickness and mean maximum screw placement torque were significantly greater in the mandible than in the maxilla. Mean mandibular screw resistance to movement was significantly greater than in the maxilla. The principal finding of this ex-vivo study was that mini screws with higher placement torque values provided greater mean resistance to movement than did screws with placement torque values lesser than 5 Ncm.

Suzuki et al[48] (2011) analyzed the placement and removal torque values of 280 orthodontic minscrew implants in the maxilla and mandible of patients. Both self drilling and pre-drilling screws of 1.5mm in diameter and 6 or 8mm length were used. Maximum insertion torque and maximum removal torque were assessed with a torque wrench. For both the pre-drilling and self-drilling miniscrews, the maximum removal torque was higher than the maximum insertion torque. Though, maximum placement torque values were found to be greater for self- drilling implants, the maximum removal torque values were found to be higher for the pre-drilling screws. They concluded that placement torque was a valid parameter to assess the quality of recipient bone.

Wilmes et al[49] (2011) analysed the threshold torque which resulted in fracture of mini implants of varying types and diameters. Forty one titanium grade 5 mini implants with diameters ranging from 1.3 to 2.0mm were inserted into acrylic glass after pre-drilling. It was found that the fracture torque varied significantly with greater diameter implants having increased fracture torque and almost all mini implants fractured at level of acrylic block at the region of the thread starting in the mini implant. The fracture torque ranged from 108.9Nmm (for 1.3 x 11mm screws) to 640.9Nmm (for 2.0 x 11mm screws). They concluded that the risk of mini implant fracture should be considered if implants of smaller diameters are used. In case of high bone density, pre-drilling should be done to avoid implant breakage.

Whang et al[50] (2011) compared the peak insertion torque values of six commercially available self-drilling mini-implants.. Twenty implants each were drilled into acrylic rods and the insertion torque values were recorded. The mini-implant was lowered until it was in contact with the substrate material and then held to maintain the pressure.Significant differences were found for peak torque values between the different implant manufacturers. This study failed to demonstrate an inverse correlation between the diameter of the mini-implants and their peak torque values. It hence appears that factors such as material composition, production technique, and the ratio between core and thread play an important role in determining the torque resistance of mini-implants.

Woodall et al[51] (2011) did an experimental study on cadavers and a three dimensional finite element anlaysis to investigate if mini screw angulation affected screw anchorage resistance. 3-D finite element models of a cylindrical miniscrew of 1.5mm diameter and bone block 6mm thick with 1.79mm thick cortical layer were made. The screws were then placed at 30°, 60° and 90° to the bone surface. Results showed maximum anchorage resistance forces of 678, 2273, 3700N for screws placed at 30°, 60°, and 90° respectively. Cortical bone stress was greatest for screws placed at 30° and least for screws placed at 90° to bone surface.

Chang et al[52] (2012) evaluated the effect of thread depth, taper shape and taper length on the mechanical properties of mini implants using both finite element method and mechanical testing using artificial bone blocks. Titanium alloy mini implants of size 2mm x 9.82mm and pitch 0.75mm were used. The thread depths were varied as 0.16, 0.24, 0.32, 0.40 and 0.48mm and the taper was varied as 0°, 3°, 5°, 7°, 11°. Mini implants with greater thread depths, smaller tapers and shorter taper length generated higher maximum stresses on the bone and threads elements and also had larger relative displacements.

Jasmine et al[53] (2012) generated finite element models of maxilla, mandible and mini implant to simulate orthodontic loading for en-masse retraction and to simulate the stress patterns in the bone and microimplant immediately after loading with different insertion angulations. AbsoAnchor mini implants of diameter 1.5mm and lengths 7mm and 8mm were considered. The authors found that the stress levels both in microimplant and cortical bone decreased as insertion angle increased from 30° to 90°. Also as the insertion angle increased, little stress was transmitted to the cancellous bone. Hence they concluded that mini implants should be placed as perpendicular to the bone as possible for better stability.

Kim et al[54] (2012) measured the cortical bone thickness in 15 men and women in the mandibular buccal and lingual areas using computed tomography. The cortical bone in the mandibular buccal and lingual areas was thicker in men than in women. In men, the mandibular lingual cortical bone was thicker than buccal region except between 1st and 2nd molars on both sides. In women, the lingual cortical bone was thicker in all areas compared to buccal cortical bone. In general, buccal cortical bone thickness increased from canines to molars. The lingual cortical bone thickness was greatest between 1st and 2nd premolars.

Liu et al[55] (2012) investigated the role of bone quality, loading conditions, screw effects using finite element analysis. A three dimensional bone block with cortical and cancellous bone in varying degrees was modelled. Pure titanium screws of diameters 1.2, 1.5 and 2mm and lengths 7, 9, 11, 13 and 15mm were modelled. Force magnitudes of 2, 4 and 6N and directions of 60°, 90° and 120° were loaded. Maximum stress around screw and bone occurred under 2N and 90° force near the entrance of screw to cortical bone. Stress in cancellous bone was much lesser than cortical bone and both stress and displacement increased with increasing cortical bone thickness.

Massey et al[56] (2012) evaluated the effects force on bony adaptations around mini screw implants and also whether bone around mini implants subjected to compressive loads adapts differently than bone around unloaded implants using micro computed tomography in six foxhounds. Results showed that loaded mini screws displayed less bone than unloaded in the cortical region but more bone than unloaded screws in the non cortical region. Larger loads produced less bone than smaller loads in the non cortical region. The layer of bone closest to the mini implants showed less bone than layers farther away. Cortical and non cortical zones under compression exhibited greater amount of bone than zones not under compression.

Migliorati et al[57] (2012) evaluated the correlations between bone characteristics, orthodontic mini screw designs, and primary stability. Four different miniscrews of sizes 1.7x10mm, 1.65 x9mm, 1.6x10mm and 1.8x10mm were placed in pig ribs. The miniscrews were first scanned with a scanning electron microscope to obtain measurable images of their threads and then inserted to a depth of 7mm. Maximum insertion torque and pull out force was measured for each screw. A positive correlation between the pitch of the mini screw and maximum insertion torque values was found. A strong correlation between maximum insertion torque and pullout force was noted. A direct correlation among cortical thickness, marrow bone density, and pullout force was observed. Differences in cortical bone thickness were more relevant for initial stability of the miniscrews than cortical bone quality.

Papageorgiou et al[58] (2012) summarized the knowledge from published clinical trials regarding the failure rates of miniscrew implants used for orthodontic anchorage purposes and attempted to the factors that possibly affect them. An overall failure rate of 13.5% was seen. Higher overall failure rates were observed when the miniscrews were inserted in the mandible than in the maxilla. No significant difference was found between the failure rates of self-drilling and not self-drilling miniscrews. Lastly, no significant differences of the failure rates of implants were observed concerning the time of orthodontic force application: ie, immediate and delayed loading.

Rao et al[59] (2012) published a review article regarding the importance of primary stability and the factors that influence it. They proposed that initial implant stability was mainly determined by bone quality and quantity and a positive correlation was found between primary stability and cortical bone thickness. The literature regarding primary stability and implant design was filled with contradictory conclusions. Cylindrical and surface roughened implants were seen to have lower failure rates. Also it was confirmed by many studies that implant diameter mainly and not implant length influences primary stability. However other studies showed that shorter implants fail more often than longer ones.

Reyenders et al[60] (2012) published a systematic review article about the insertion torque and success of mini implants and analysed whether the recommended maximum insertion torque values of 5 to 10Ncm were associated with higher success rates compared to mini implants inserted at torque values beyond this range. They concluded that an association between maximum insertion torque values and success was analysed only in nonrandomized studies of low quality and that success is a subjective recording and should not be considered as a reliable factor for testing associations with maximum insertion torques. Subsequent studies should be done with a digital torque sensor and the review should be considered as a negative study as no evidence based conclusions could be drawn.

Shah et al[61] (2012) did a study to determine the effect of altering implant length, diameter, cortical bone thickness and density on the primary stability of mini implants. Results showed that the shorter mini implants had insertion torque and pull out strength lower than the longer implants. Increasing the outer diameter by 0.25mm significantly increased the primary stability. Decreasing the cortical thickness and density also resulted in lower insertion torque and pull out strength.

Singh et al[62] (2012) analyzed the stress distribution and displacement patterns that develop in miniscrew implant made of stainless steel and titanium alloy and its surrounding bone. It was seen that stress distribution was not significantly different between the 2 types of implant material. Increased stress values were located at the necks of the implants and the surrounding cortical bone. Stainless steel screws had greater stress compared to titanium alloy screws. Bending of the titanium miniscrew was observed in the neck region under horizontal traction. Amount of stress transferred to cancellous bone was minimal.

Tachibana et al[63] (2012) measured the placement torque value of self-drilling mini implants 1.6mm in diameter and 8mm long in pig ribs. The peak mini-implant placement torque was measured using a digital torque tester. In the maxillary bone model, the torque in the self-drilling group was 8.2 N cm; in the pre-drilling group with a 1.0 mm diameter pilot hole, the torque was 7.1 N cm. These values were in the range 5– 10 N cm. Therefore, it is preferable to use the self-drilling method or the pre-drilling method with a 1.0 mm diameter pilot hole for the maxillary bone. In the mandibular alveolar bone model, the torque in all the self-drilling groups and the pre-drilling group, except for the case with a 1.3 mm diameter pilot hole, exceeded 10 N cm. Therefore, in the mandible, the authors suggest that a 1.3 mm diameter pilot hole be used to place the mini-implant to ensure an acceptable torque range. The histological results concurred with the placement torque results.

Lin et al[64] (2013) did a finite element study and factorial analysis to determine the effect of exposure length of mini implant, insertion angle and the direction of orthodontic force. Stainless steel implants of diameter 2mm and lengths 8, 10, 12mm were modelled. Computed tomographic images were obtained for the mandible. Insertion of mini implant was simulated between premolar and molar at 60°, 90° and 120° with thread depth of insertion at 3mm, 5mm and 7mm and orthodontic force of 2N was applied to the top surface of the mini implant and inclined in the proximal direction to imitate en masse retraction.

They found that maximum stress occurred when the exposure length of the mini implant was 7mm and as the length decreased, the stress also decreased. Stress in cancellous bone was greatest with insertion angle of 60°. Most stresses were concentrated around the region of insertion of mini implant.

Pithon et al[65] (2013) evaluated the influence of length of mini implant and cortical bone thickness on insertion torque. Mini implants of 1.5mm diameter and lengths 6mm, 8mm and 10mm were tested in pig ribs of varying cortical bone thickness. They found that the insertion torque increased with increasing screw length and increasing cortical bone thickness and concluded that though increasing the length of the screw doesn't increase its mechanical strength, it can efficiently reinforce the initial stability of mini implants.

Sana et al[66] (2013) provided an overview on the current literature available on mini implants with regards to their material properties. They stated the ideal qualities required of a mini implant material and gave the mechanical properties of stainless steel, cobalt-chromium alloys, titanium and its alloys which are currently in use.

Serra et al[67] **(2013)** 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.

Alrbata et al[68] **(2014)** 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.

Kalra et al[69] (2014) compared the accuracy of two dimensional radiographs to CBCT for mini implant placement and found a significant difference between both groups for deviation from ideal height of placement. They concluded that although CBCT provides an accurate three dimensional visualization of the inter-radicular space, the two dimensional intra oral radiographs provide sufficient information for implant placement.

MATERIALS&METHODS

MATERIALS USED IN THE STUDY:

1. Eighty Absoanchor Ti-6Al-4V alloy mini implants by Dentos®, Korea
2. Long handle implant driver with torque gauge, Dentos®, Korea
3. Bovine rib bone
4. Normal saline
5. Osteotome (Orthomax)
6. Auto-polymerizing resin (DPI)
7. Spirit level
8. Customized stand for implant placement
9. Discovery XR656 digital radiographic machine by G.E.®
10. G.E. Media Viewer software for image analysis
11. 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 and was approved by the Ethical Committee of the institution.

Eighty Absoanchor self-drilling, mini implants made of Titanium-6Aluminium-4Vanadium [Ti-6Al-4V] alloy implants from Dentos® Korea, of the following dimensions were used for the experiment [fig 1-4]:

Length 6mm with diameters: 1.3mm, 1.4mm, 1.5mm and 1.6mm	5 screws of each dimension Total 20 mini implants
Length 7mm with diameters: 1.3mm, 1.4mm, 1.5mm and 1.6mm	5 screws of each dimension Total 20 mini implants
Length 8mm with diameters: 1.3mm, 1.4mm, 1.5mm and 1.6mm	5 screws of each dimension Total 20 mini implants
Length 10mm with diameters: 1.3mm, 1.4mm, 1.5mm and 1.6mm	5 screws of each dimension Total 20 mini implants

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.

Preparation of bone segments:

An osteotome [fig 5] was used to segment fresh bovine rib into pieces 1.5cm wide. The segments were embedded in autopolymerising resin blocks of 15cm x 5cm x 2cm. Four rib segments were embedded in each block.

To ensure that the point of insertion of the implant was truly horizontal, a spirit level was placed on the surface of each of the rib segments during embedding [fig 6, 7]. Twenty bone segment blocks were thus prepared and were segregated for implant insertion such that one block had four mini screws of similar length and varying diameter. A pictorial representation of the resin block with rib bone segments is shown in Figure 8.

Bovine rib was used in this study as previous studies by **Laurito et al**[39] have shown that bovine rib architecture is similar to the human mandibular architecture. Bovine rib is one of the preferred human bone substitutes in ex-vivo implantology studies. The bovine rib was stored in normal saline and kept moist till the time of insertion as done by **Chatzigianni et al**[44] [fig 9]

Insertion of mini implants:

A long handle implant driver from Dentos®, Korea with torque gauge fixed at 1kg/cm [i.e.9.8N] was used for the study [fig 10, 11]. The torque force can be adjusted from 0.5Kg.cm to 2Kg.cm. The driver emitted a clicking noise when the torque level exceeded the set value.

A stand was custom fabricated for the study using polymerized nylon and chrome plated steel [fig 12, 13]. The implant, implant driver and the resin block were held perpendicular to each other in the custom made stand [fig 14, 15] The stand was made with telescopic axes to enable adjustment of the bone block and driver interface in all three planes of space. The mini implant was inserted into the bone segment by slow continuous manual insertion. Likewise, all the remaining implants were also inserted one mini implant per bone segment.

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 with the X-ray source 100cm from the object set at 80kV and 292mAs was used with radiographic exposure time of 1milli second [fig 16, 17]. 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 [fig 18, 19].

Image analysis for deflection measurement:

The radiographic image obtained was adjusted for optimum contrast and magnification prior to obtaining the mini implant deflection values [fig 20].A pictorial representation of the image analysis is shown in Figure 21.In the image, the black line AB represents the true horizontal line passing through the centre of point of insertion of the implant. The red line XY represents the long axis of the mini implant passing through its apex and tip. Ø is the angle between the two lines AB and XY and represents the degree of deflection of the mini implant.

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 [Fig 22]. 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 23]. The procedure was thus repeated for all the 80 mini implants.

Figure 1- Length 6mm- diameters from left to right-1.3mm, 1.4mm, 1.5mm, 1.6mm

Figure 2-Length 7mm- diameters from left to right-1.3mm, 1.4mm, 1.5mm, 1.6mm

Figure 3-Length 8mm- diameters from left to right-1.3mm, 1.4mm, 1.5mm, 1.6mm

Figure 4-Length 10mm- diameters from left to right-1.3mm, 1.4mm, 1.5mm, 1.6mm

Figure 5- Osteotome for cutting bovine rib bone bovine rib embedded in resin

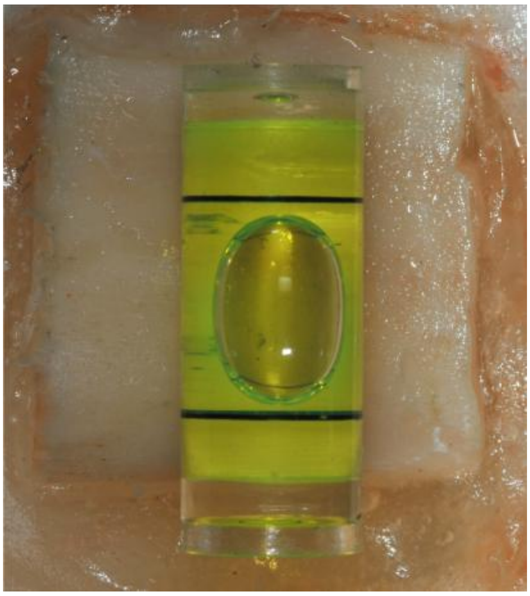


Figure 6-Spirit level used to ensure bone segment is horizontal

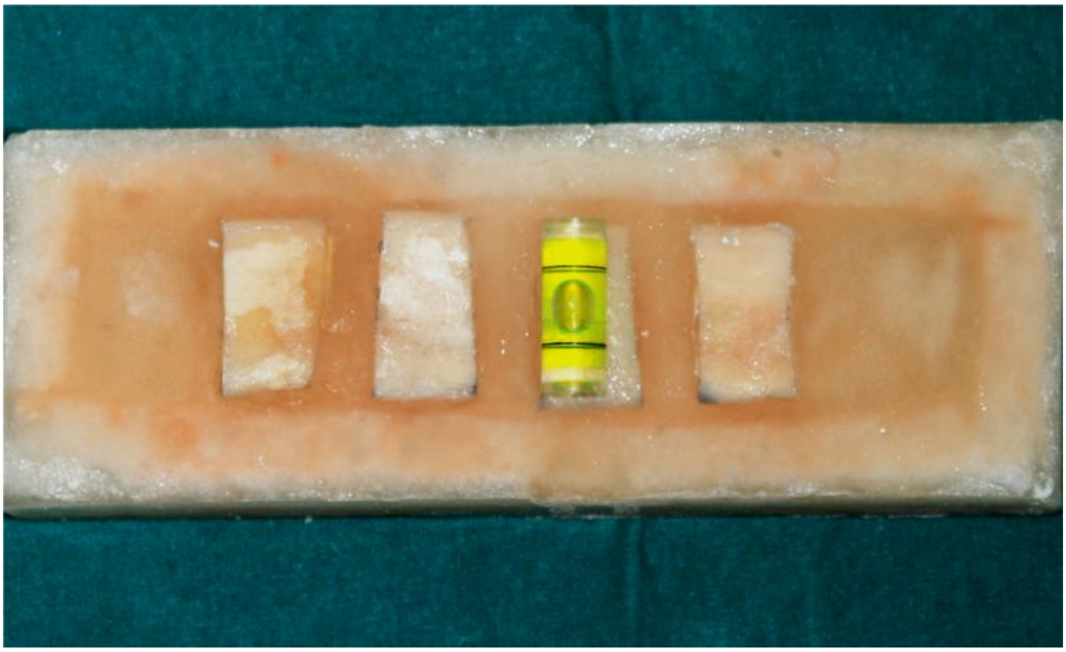


Figure 7- Bone block prepared with four segments in each block

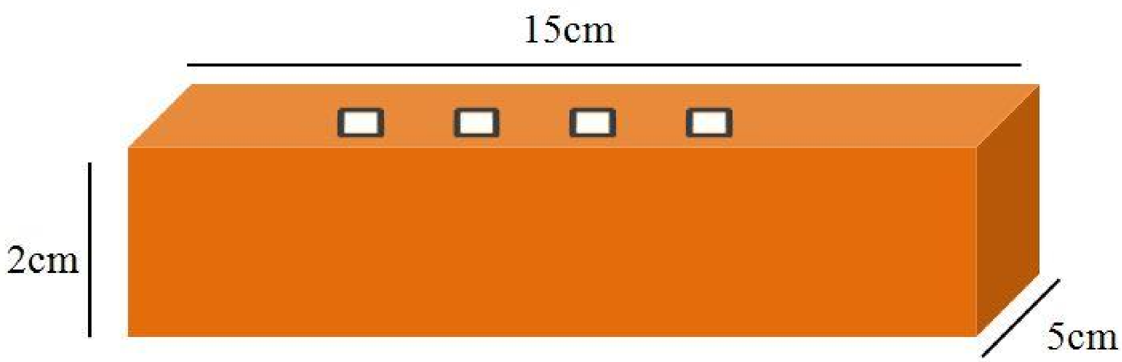


Figure 8- Pictorial representation of acrylic block with bovine rib bone segments

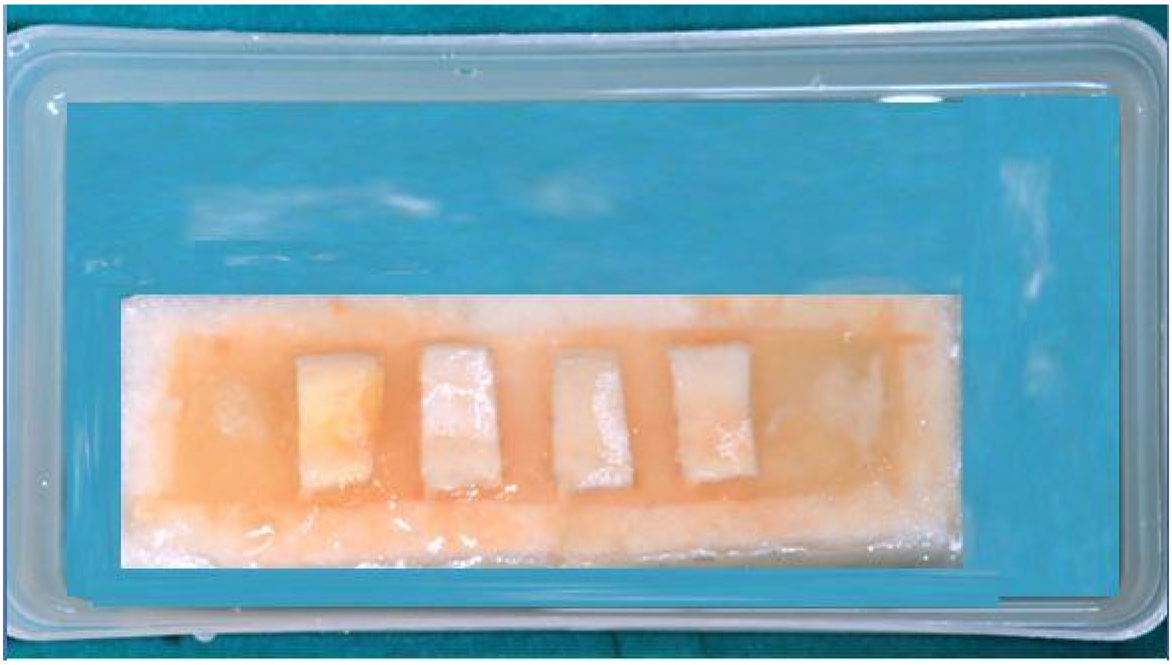


Figure 9- Embedded bone is stored in normal saline

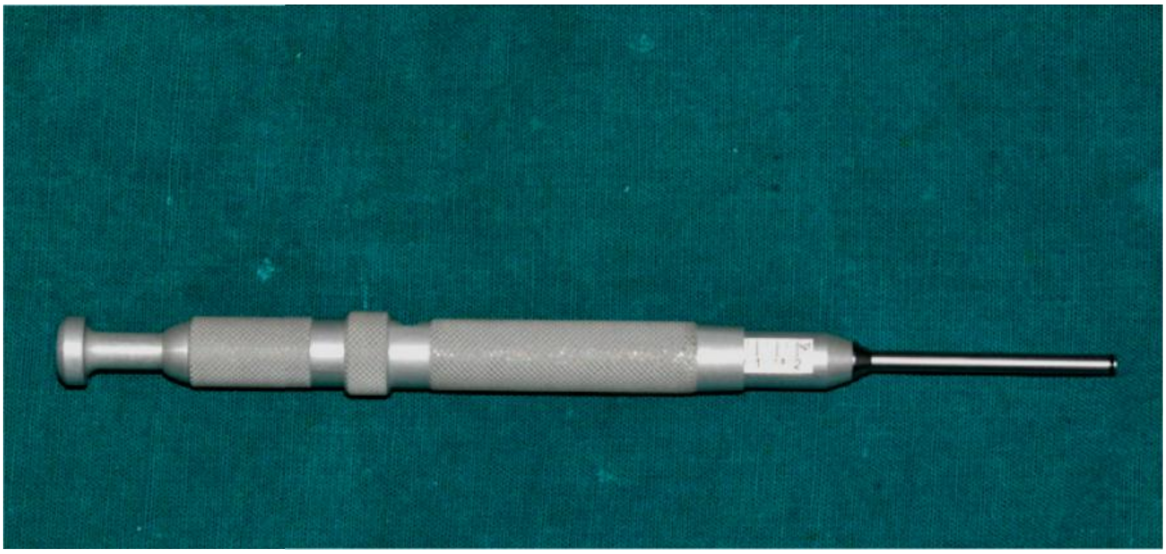


Figure10- Long handle implant driver with adjustable torque gauge

Figure 11- Torque set at 1kg.cm

Figure 12- Customized stand lateral view

Figure 13- Customized stand frontal view

Figure 14- Stand with bone, mini implant and driver

Figure 15- Torque level kept at 1kg.cm during insertion

Figure 16- G.E. Discovery XR656 radiographic machine

Figure 17- Settings used for the radiographic imaging of the block

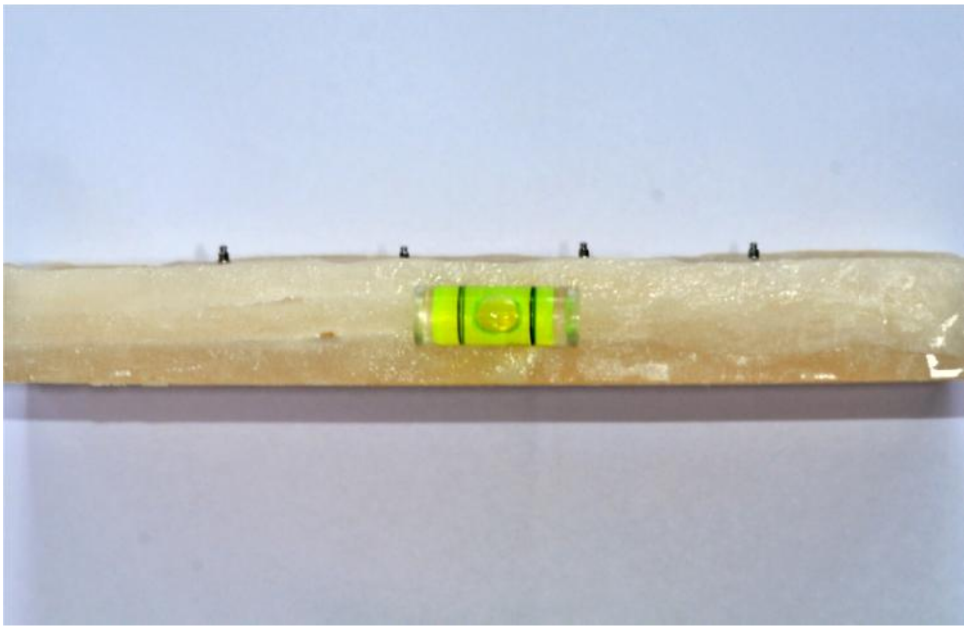


Figure 18- Block surface checked with spirit level

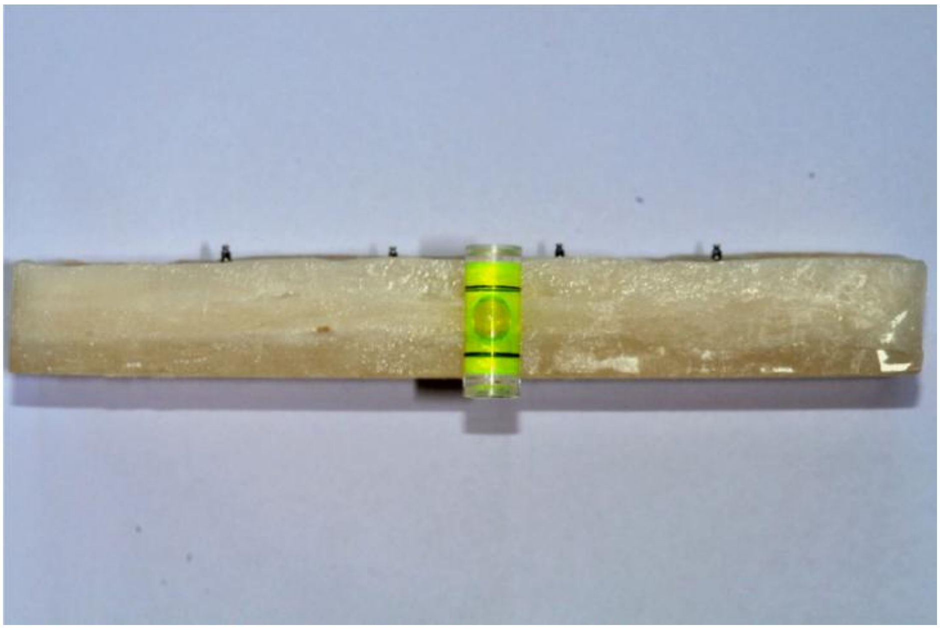


Figure 19- Block surface checked with spirit level

Figure 20- Radiographic image obtained of a bone block

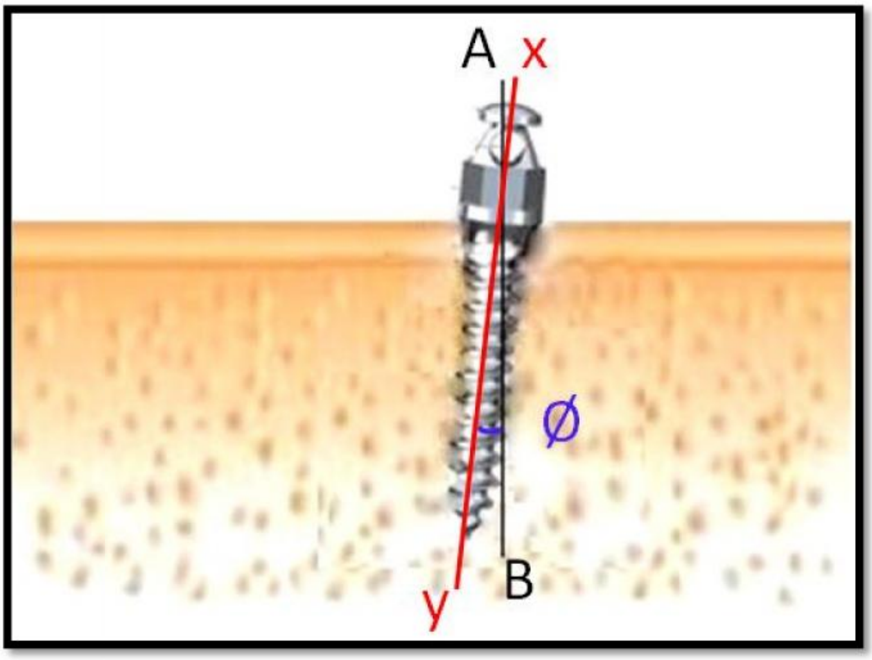


Figure 21- Pictorial representation of the image analysis

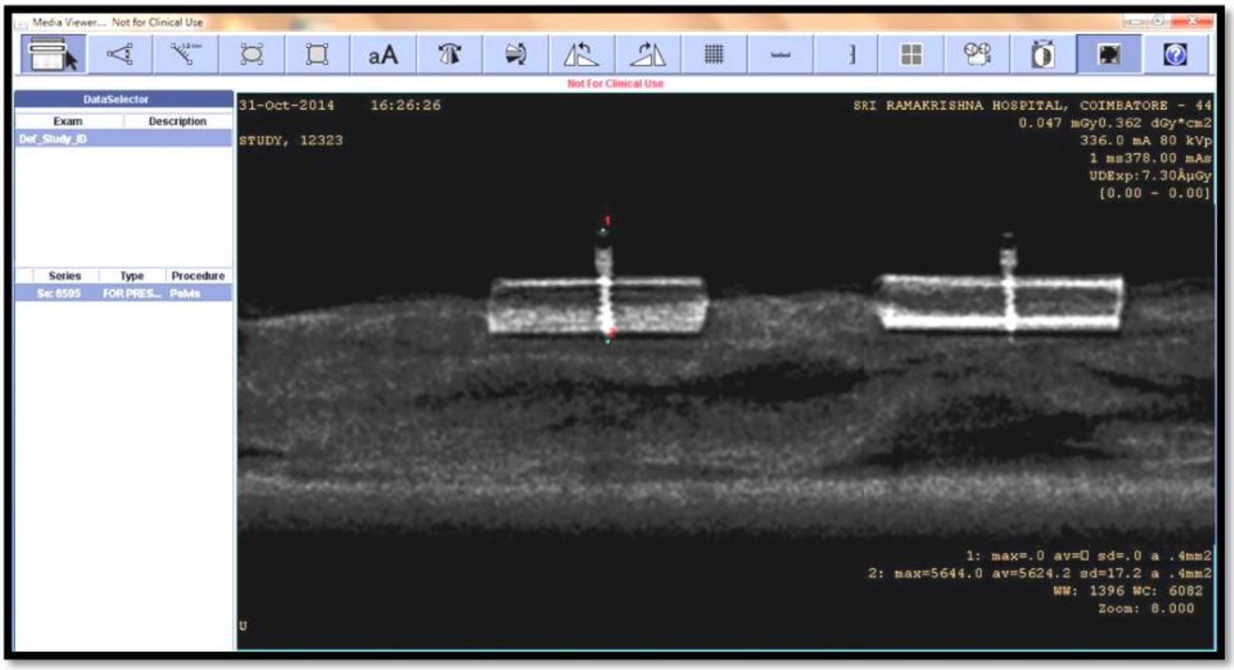


Figure 22- Points marked to draw line through long axis of mini implant

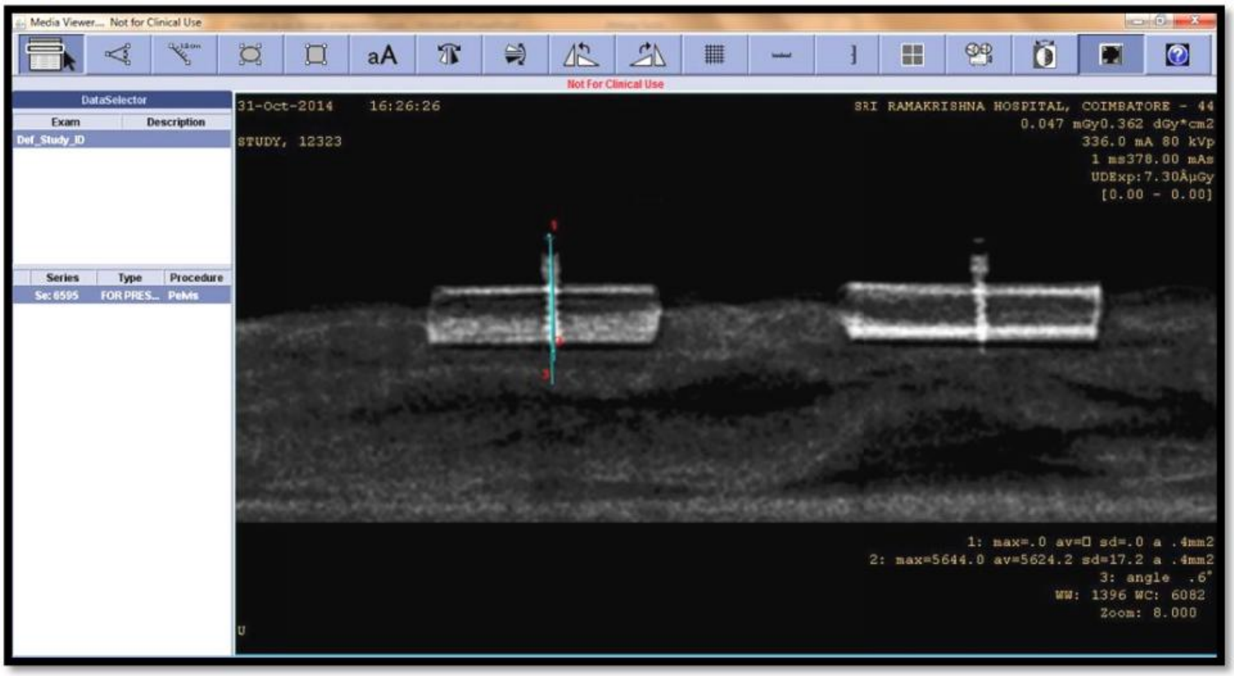


Figure 23- Analysis of image using G.E. Media Viewer software

RESULTS

A total of 80 mini implants were tested of which 2 mini implants of size 1.6mm x 8mm fractured and hence were not included in the study. The results of this in-vitro study using titanium alloy mini implants in comparing the deflection produced by implants of various diameters and length is presented as follows:

Descriptive statistics for the measurements were computed with SPSS statistical software package and the assumption of normality of the variables was investigated by the Kolmogorov-Smirnov test. The mean deflections of the various dimensions of implants used in the study are shown in table 1. All mini implants underwent deflection upon insertion with a maximum mean deflection 2.9 degrees and a minimum of 0.6 degrees. A test of between subjects' effects was done to assess the influence of length and diameter and also the combined effects of length and diameter on deflection. The influence of all three parameters was found to be statistically significant (Table 2).

The individual effect of constant diameter with varying length and also constant length and varying diameter was assessed using one way Analysis of Variance and Post Hoc comparisons at 95% confidence interval.

Parameters assessed:

I. Comparison of the deflection in various diameters of varying length

II. Comparison of the deflection in various lengths of varying diameters

I. **Comparison of the deflection in various diameters of varying length:**

There was decrease in the amount of deflection observed with the corresponding decrease in length.

(a) Comparison of 1.3mm diameter mini implants of lengths 10mm, 8mm, 7mm and 6mm:

The 10mm mini screws showed maximum deflection followed by 8mm then 7mm and the least deflection was seen in the 6mm long screws. This difference was seen to be statistically significant ($p < 0.05$).The maximum difference of 1 degree was seen between the 10mm and 6mm screws and a minimum difference of 0.28 degrees was seen between the 8mm and 7mm long screws. The above result is depicted in table 3 and graph 1.

(b) Comparison of 1.4mm diameter mini implants of lengths 10mm, 8mm, 7mm and 6mm:

The 10mm mini screws showed maximum deflection followed by 8mm then 7mm and the least deflection was seen in the 6mm long screws. The maximum difference of 1.08 degrees was seen between the 10mm and 6mm screws and a minimum difference of 0.30 degrees was seen between the 10mm and 8mm screws and 8mm and 7mm long screws. The difference in the mean deflection observed was statistically significant ($p < 0.05$). The above result is depicted in table 4 and graph 2.

(c) Comparison of 1.5mm diameter mini implants of lengths 10mm, 8mm, 7mm and 6mm:

The 10mm mini screws showed maximum deflection followed by 8mm then 7mm and the least deflection was seen in the 6mm long screws. The maximum difference of 1.30 degrees was seen between the 10mm and 6mm screws and a minimum difference of 0.28 degrees was seen between the 10mm and 8mm screws.

This difference was seen to be statistically significant ($p < 0.05$). The above result is depicted in table 5 and graph 3.

(d) Comparison of 1.6mm diameter mini implants of lengths 10mm, 8mm, 7mm and 6mm:

The 10mm mini screws showed maximum deflection followed by 8mm then 7mm and the least deflection was seen in the 6mm long screws. On comparison of mean difference of deflection, it was seen to be statistically significant ($p < 0.05$). The maximum difference of 1.30 degrees was seen between the 10mm and 6mm screws and a minimum difference of 0.20 degrees was seen between the 10mm and 8mm screws. The above result is depicted in table 6 and graph 4.

II. **Comparison of various lengths of varying diameter:**

There was an inverse relation seen with respect to the effect of varying the diameter of the mini implant.

(a) Comparison of 10mm long implants of diameters 1.3mm, 1.4mm, 1.5mm and 1.6mm

The smaller diameter i.e. 1.3mm implants showed the greatest deflection followed by 1.4mm, 1.5mm and the least deflection was observed for the 1.6mm wide mini implants. The maximum difference of 1 degree was seen between the 1.3mm and 1.6mm diameter screws and a minimum difference of 0.28 degrees was seen between the 1.5mm and 1.6mm screws. These discrete values were found to be statistically significant ($p < 0.05$). This is represented in table 7 and graph 5.

(b) Comparison of 8mm long implants of diameters 1.3mm, 1.4mm, 1.5mm and 1.6mm

The smallest diameter i.e. 1.3mm implants showed the greatest deflection followed by 1.4mm, 1.5mm and the least deflection was observed for the 1.6mm wide mini implants.

The maximum difference of 0.9 degrees was seen between the 1.3mm and 1.6mm diameter screws and a minimum difference of 0.20 degrees was seen between the 1.5mm and 1.6mm screws. The difference in values was found to be statistically significant ($p < 0.05$). This is represented in table 8 and graph 6.

(c) Comparison of 7mm long implants of diameters 1.3mm, 1.4mm, 1.5mm and 1.6mm:

The smallest diameter i.e. 1.3mm implants showed the greatest deflection followed by 1.4mm, 1.5mm and the least deflection was observed for the 1.6mm wide mini implants. The maximum difference of 1.24 degrees was seen between the 1.3mm and 1.6mm diameter screws and a minimum difference of 0.40 degrees was seen between the 1.4mm and 1.5mm screws. These discrete values were found to be statistically significant ($p < 0.05$). This is represented in table 9 and graph 7.

(d) Comparison of 6mm long implants of diameters 1.3mm, 1.4mm, 1.5mm and 1.6mm:

The 1.3mm implants showed the greatest deflection followed by 1.4mm, 1.5mm and the least deflection was observed for the 1.6mm wide mini implants. A maximum difference of 1.30 degrees was seen between the 1.3mm and 1.6mm diameter screws and a minimum difference of 0.28 degrees was seen between the

1.4mm and 1.5mm screws. These discrete values were found to be statistically significant ($p < 0.05$). This is represented in table 10 and graph 8.

The overall comparison of the deflection values of mean of diameters for various lengths is represented in graph 9. It shows a progressive decrease in deflection with both increase in diameter and also decrease in length.

Table 1- descriptive statistics of data

Dependent Variable: Data

		length of mini implant	diameter of mini implant			
			1.3mm	1.4mm	1.5mm	1.6mm
Data	N	10mm	5	5	5	5
		8mm	5	5	5	3
		7mm	5	5	5	5
		6mm	5	5	5	5
	Mean	10mm	2.9	2.5	2.18	1.9
		8mm	2.6	2.2	1.9	1.7
		7mm	2.32	1.9	1.5	1.08
		6mm	1.9	1.42	0.88	0.6

Table 2- test of between subjects effect

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Corrected Model	29.834(a)	15	1.989	3082.831	.000
Intercept	260.721	1	260.721	404117.736	.000
LENGTH	15.357	3	5.119	7934.469	.000
DIAMETER	13.118	3	4.373	6777.423	.000
LENGTH * DIAMETER	.359	9	.040	61.849	.000
Error	.040	62	.001		
Total	295.720	78			
Corrected Total	29.874	77			

Table 3- ANOVA test 1.3mm diameter implants

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	2.714	3	.905	1809.333	.000
Within Groups	.008	16	.000		
Total	2.722	19			

Post Hoc comparison

(I) length of mini implant	(J) length of mini implant	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
10mm	8mm	.3000(*)	.01414	.000	.2595	.3405
	7mm	.5800(*)	.01414	.000	.5395	.6205
	6mm	1.0000(*)	.01414	.000	.9595	1.0405
8mm	10mm	-.3000(*)	.01414	.000	-.3405	-.2595
	7mm	.2800(*)	.01414	.000	.2395	.3205
	6mm	.7000(*)	.01414	.000	.6595	.7405
7mm	10mm	-.5800(*)	.01414	.000	-.6205	-.5395
	8mm	-.2800(*)	.01414	.000	-.3205	-.2395
	6mm	.4200(*)	.01414	.000	.3795	.4605
6mm	10mm	-1.0000(*)	.01414	.000	-1.0405	-.9595
	8mm	-.7000(*)	.01414	.000	-.7405	-.6595
	7mm	-.4200(*)	.01414	.000	-.4605	-.3795

The mean difference is significant at the .05 level

Table 4- ANOVA test for 1.4mm diameter mini implants

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	3.182	3	1.061	2121.000	.000
Within Groups	.008	16	.001		
Total	3.190	19			

Post Hoc Comparisons

(I) length of mini implant	(J) length of mini implant	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
10mm	8mm	.3000(*)	.01414	.000	.2595	.3405
	7mm	.6000(*)	.01414	.000	.5595	.6405
	6mm	1.0800(*)	.01414	.000	1.0395	1.1205
8mm	10mm	-.3000(*)	.01414	.000	-.3405	-.2595
	7mm	.3000(*)	.01414	.000	.2595	.3405
	6mm	.7800(*)	.01414	.000	.7395	.8205
7mm	10mm	-.6000(*)	.01414	.000	-.6405	-.5595
	8mm	-.3000(*)	.01414	.000	-.3405	-.2595
	6mm	.4800(*)	.01414	.000	.4395	.5205
6mm	10mm	-1.0800(*)	.01414	.000	-1.1205	-1.0395
	8mm	-.7800(*)	.01414	.000	-.8205	-.7395
	7mm	-.4800(*)	.01414	.000	-.5205	-.4395

The mean difference is significant at the .05 level.

Table 5- ANOVA for 1.5mm mini implants

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	4.770	3	1.590	1589.833	.000
Within Groups	.016	16	.001		
Total	4.786	19			

Post Hoc Comparisons

(I) length of mini implant	(J) length of mini implant	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
10mm	8mm	.2800(*)	.02000	.000	.2228	.3372
	7mm	.6800(*)	.02000	.000	.6228	.7372
	6mm	1.3000(*)	.02000	.000	1.2428	1.3572
8mm	10mm	-.2800(*)	.02000	.000	-.3372	-.2228
	7mm	.4000(*)	.02000	.000	.3428	.4572
	6mm	1.0200(*)	.02000	.000	.9628	1.0772
7mm	10mm	-.6800(*)	.02000	.000	-.7372	-.6228
	8mm	-.4000(*)	.02000	.000	-.4572	-.3428
	6mm	.6200(*)	.02000	.000	.5628	.6772
6mm	10mm	-1.3000(*)	.02000	.000	-1.3572	-1.2428
	8mm	-1.0200(*)	.02000	.000	-1.0772	-.9628
	7mm	-.6200(*)	.02000	.000	-.6772	-.5628

The mean difference is significant at the .05 level.

Table 6- ANOVA test for 1.6mm diameter mini implants

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	4.963	3	1.654	2895.148	.000
Within Groups	.008	14	.001		
Total	4.971	17			

Post Hoc comparisons

(I) length of mini implant	(J) length of mini implant	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
10mm	8mm	.2000(*)	.01746	.000	.1493	.2507
	7mm	.8200(*)	.01512	.000	.7761	.8639
	6mm	1.3000(*)	.01512	.000	1.2561	1.3439
8mm	10mm	-.2000(*)	.01746	.000	-.2507	-.1493
	7mm	.6200(*)	.01746	.000	.5693	.6707
	6mm	1.1000(*)	.01746	.000	1.0493	1.1507
7mm	10mm	-.8200(*)	.01512	.000	-.8639	-.7761
	8mm	-.6200(*)	.01746	.000	-.6707	-.5693
	6mm	.4800(*)	.01512	.000	.4361	.5239
6mm	10mm	-1.3000(*)	.01512	.000	-1.3439	-1.2561
	8mm	-1.1000(*)	.01746	.000	-1.1507	-1.0493
	7mm	-.4800(*)	.01512	.000	-.5239	-.4361

The mean difference is significant at the .05 level.

Table 7- ANOVA test for length 10mm mini implants

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	2.774	3	.925	1849.333	.000
Within Groups	.008	16	.000		
Total	2.782	19			

Post Hoc Comparisons

(I) diameter of mini implant	(J) diameter of mini implant	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
1.3mm	1.4mm	.4000(*)	.01414	.000	.3595	.4405
	1.5mm	.7200(*)	.01414	.000	.6795	.7605
	1.6mm	1.0000(*)	.01414	.000	.9595	1.0405
1.4mm	1.3mm	-.4000(*)	.01414	.000	-.4405	-.3595
	1.5mm	.3200(*)	.01414	.000	.2795	.3605
	1.6mm	.6000(*)	.01414	.000	.5595	.6405
1.5mm	1.3mm	-.7200(*)	.01414	.000	-.7605	-.6795
	1.4mm	-.3200(*)	.01414	.000	-.3605	-.2795
	1.6mm	.2800(*)	.01414	.000	.2395	.3205
1.6mm	1.3mm	-1.0000(*)	.01414	.000	-1.0405	-.9595
	1.4mm	-.6000(*)	.01414	.000	-.6405	-.5595
	1.5mm	-.2800(*)	.01414	.000	-.3205	-.2395

The mean difference is significant at the .05 level.

Table 8- ANOVA test for length 8mm mini implants

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	1.944	3	.648	27599870709250900000000000000000.000	.000
Within Groups	.000	14	.000		
Total	1.944	17			

Post Hoc Comparisons

(I) diameter of mini implant	(J) diameter of mini implant	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
1.3mm	1.4mm	.4000(*)	.00000	.000	.4000	.4000
	1.5mm	.7000(*)	.00000	.000	.7000	.7000
	1.6mm	.9000(*)	.00000	.000	.9000	.9000
1.4mm	1.3mm	-.4000(*)	.00000	.000	-.4000	-.4000
	1.5mm	.3000(*)	.00000	.000	.3000	.3000
	1.6mm	.5000(*)	.00000	.000	.5000	.5000
1.5mm	1.3mm	-.7000(*)	.00000	.000	-.7000	-.7000
	1.4mm	-.3000(*)	.00000	.000	-.3000	-.3000
	1.6mm	.2000(*)	.00000	.000	.2000	.2000
1.6mm	1.3mm	-.9000(*)	.00000	.000	-.9000	-.9000
	1.4mm	-.5000(*)	.00000	.000	-.5000	-.5000
	1.5mm	-.2000(*)	.00000	.000	-.2000	-.2000

The mean difference is significant at the .05 level.

Table 9- ANOVA test for length 7mm mini implants

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	4.244	3	1.415	1414.667	.000
Within Groups	.016	16	.001		
Total	4.260	19			

Post Hoc Comparisons

(I) diameter of mini implant	(J) diameter of mini implant	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
1.3mm	1.4mm	.4200(*)	.02000	.000	.3628	.4772
	1.5mm	.8200(*)	.02000	.000	.7628	.8772
	1.6mm	1.2400(*)	.02000	.000	1.1828	1.2972
1.4mm	1.3mm	-.4200(*)	.02000	.000	-.4772	-.3628
	1.5mm	.4000(*)	.02000	.000	.3428	.4572
	1.6mm	.8200(*)	.02000	.000	.7628	.8772
1.5mm	1.3mm	-.8200(*)	.02000	.000	-.8772	-.7628
	1.4mm	-.4000(*)	.02000	.000	-.4572	-.3428
	1.6mm	.4200(*)	.02000	.000	.3628	.4772
1.6mm	1.3mm	-1.2400(*)	.02000	.000	-1.2972	-1.1828
	1.4mm	-.8200(*)	.02000	.000	-.8772	-.7628
	1.5mm	-.4200(*)	.02000	.000	-.4772	-.3628

The mean difference is significant at the .05 level.

Table 10- ANOVA test for length 6mm mini implants

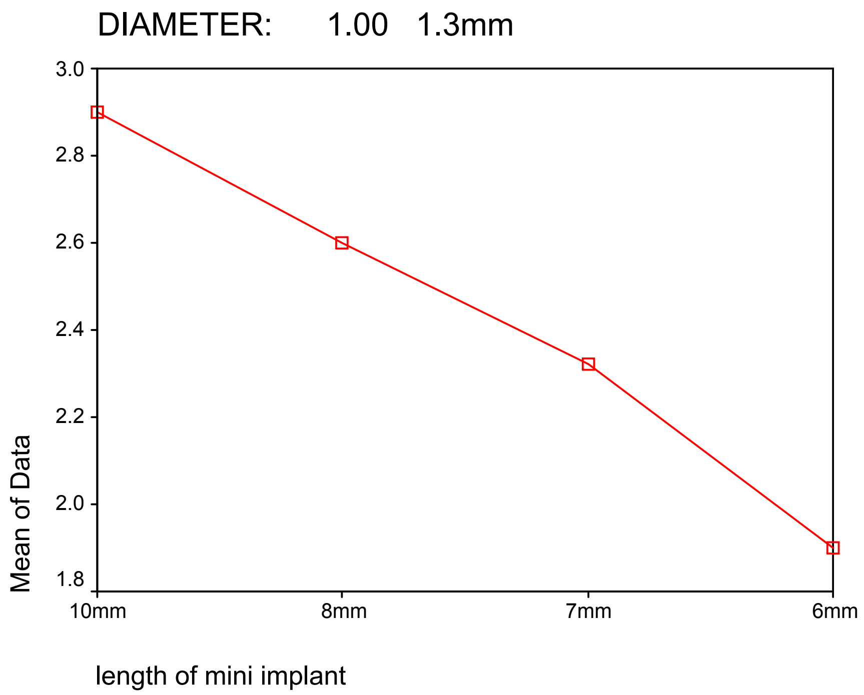
	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	5.004	3	1.668	1668.000	.000
Within Groups	.016	16	.001		
Total	5.020	19			

Post Hoc Comparison

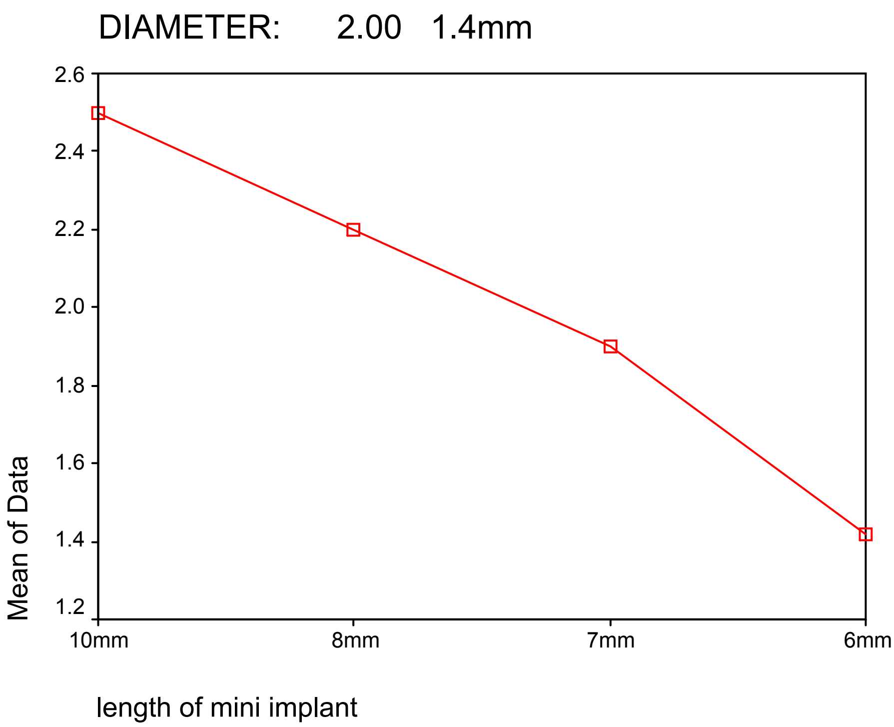
(I) diameter of mini implant	(J) diameter of mini implant	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
1.3mm	1.4mm	.4800(*)	.02000	.000	.4228	.5372
	1.5mm	1.0200(*)	.02000	.000	.9628	1.0772
	1.6mm	1.3000(*)	.02000	.000	1.2428	1.3572
1.4mm	1.3mm	-.4800(*)	.02000	.000	-.5372	-.4228
	1.5mm	.5400(*)	.02000	.000	.4828	.5972
	1.6mm	.8200(*)	.02000	.000	.7628	.8772
1.5mm	1.3mm	-1.0200(*)	.02000	.000	-1.0772	-.9628
	1.4mm	-.5400(*)	.02000	.000	-.5972	-.4828
	1.6mm	.2800(*)	.02000	.000	.2228	.3372
1.6mm	1.3mm	-1.3000(*)	.02000	.000	-1.3572	-1.2428
	1.4mm	-.8200(*)	.02000	.000	-.8772	-.7628
	1.5mm	-.2800(*)	.02000	.000	-.3372	-.2228

The mean difference is significant at the .05 level.

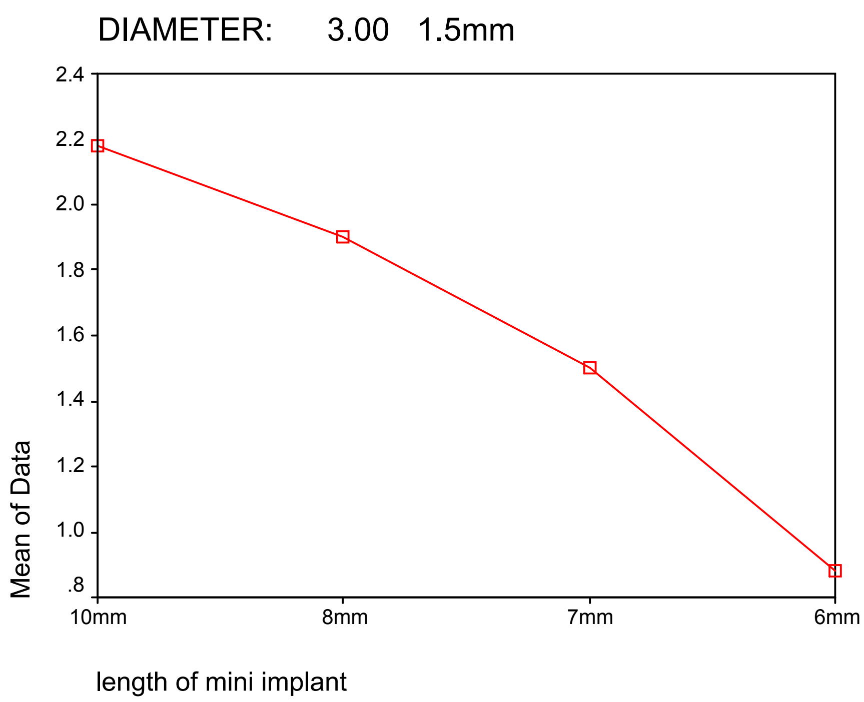
Graph 1- deflection for 1.3mm diameter mini implants



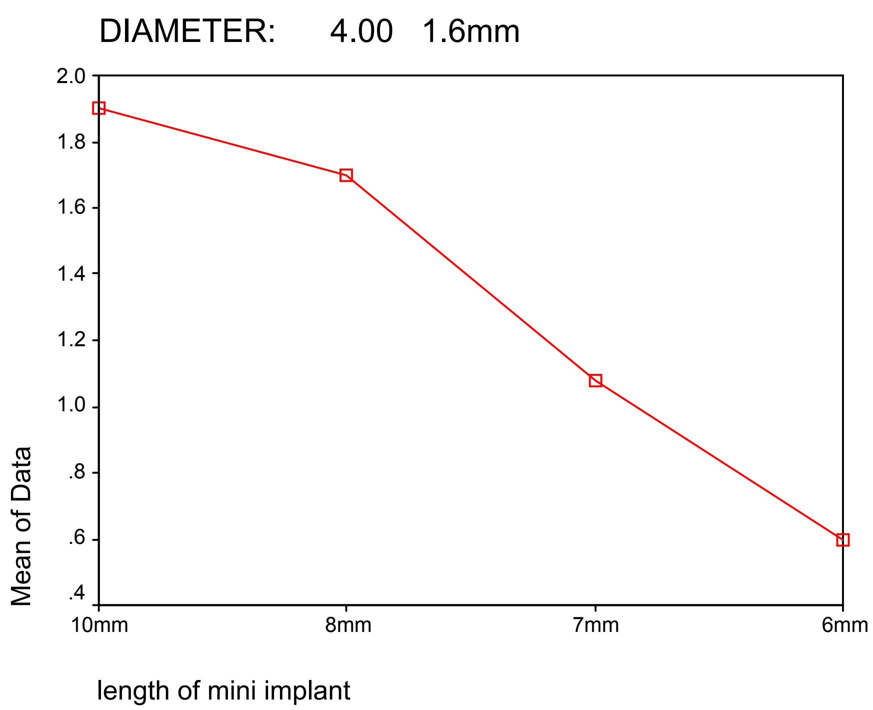
Graph 2- deflection for 1.4mm diameter mini implants



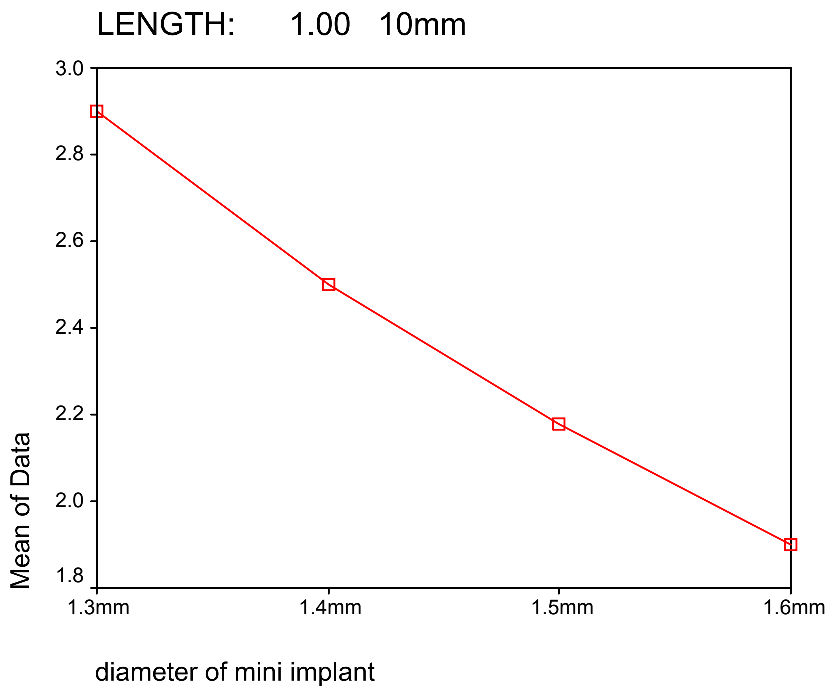
Graph 3- Deflection for 1.5mm diameter mini implants



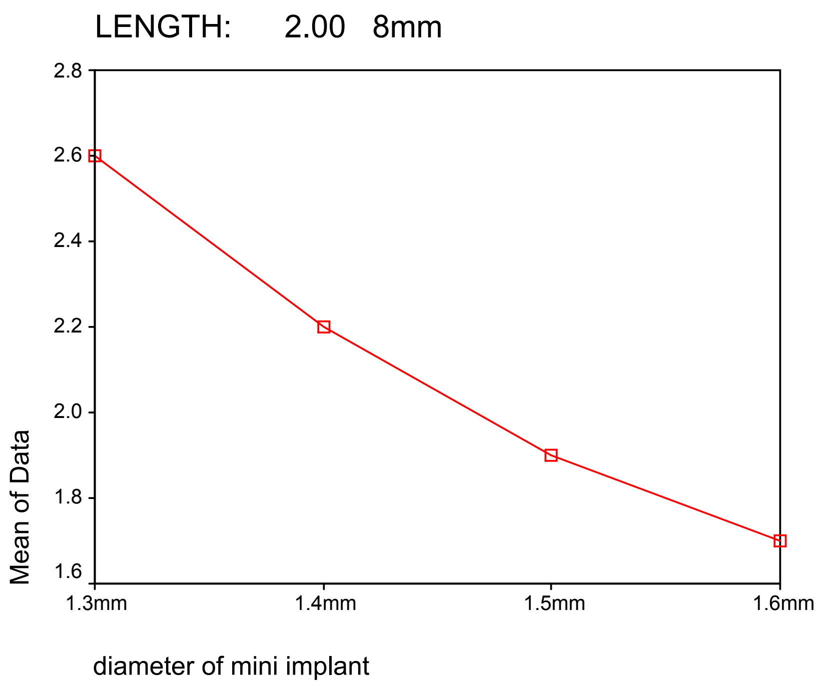
Graph 4- Deflection for 1.6mm diameter mini implants



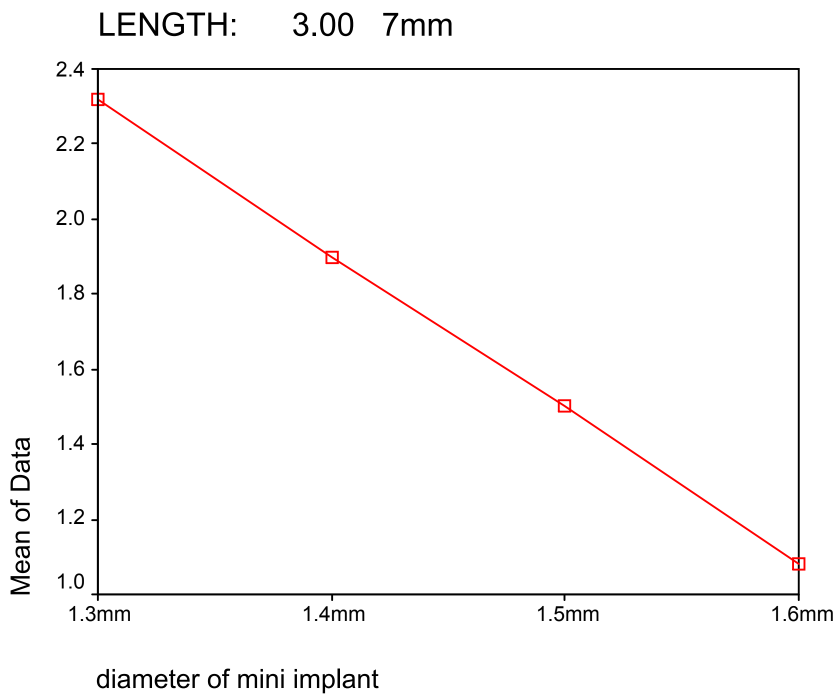
Graph 5- Deflection for length 10mm mini implants



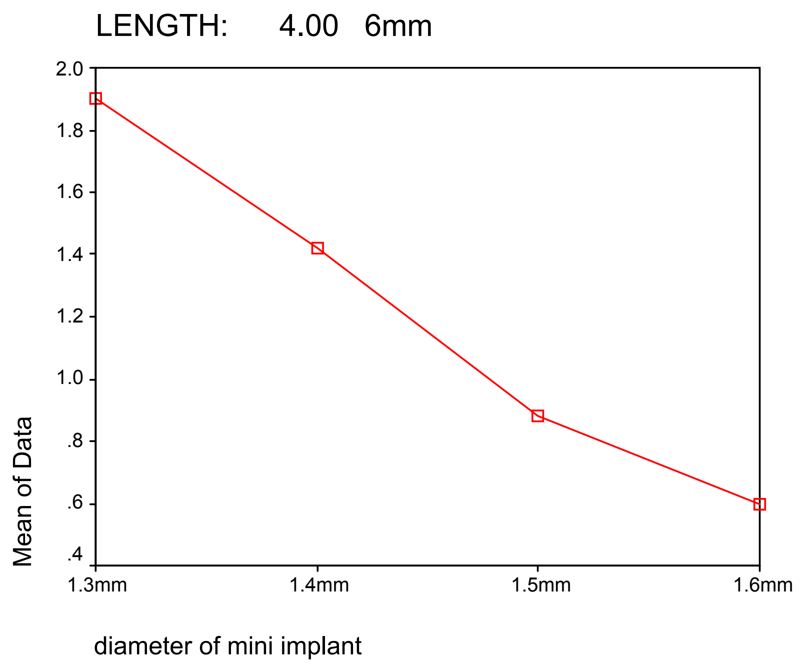
Graph 6- Deflection for length 8mm mini implants



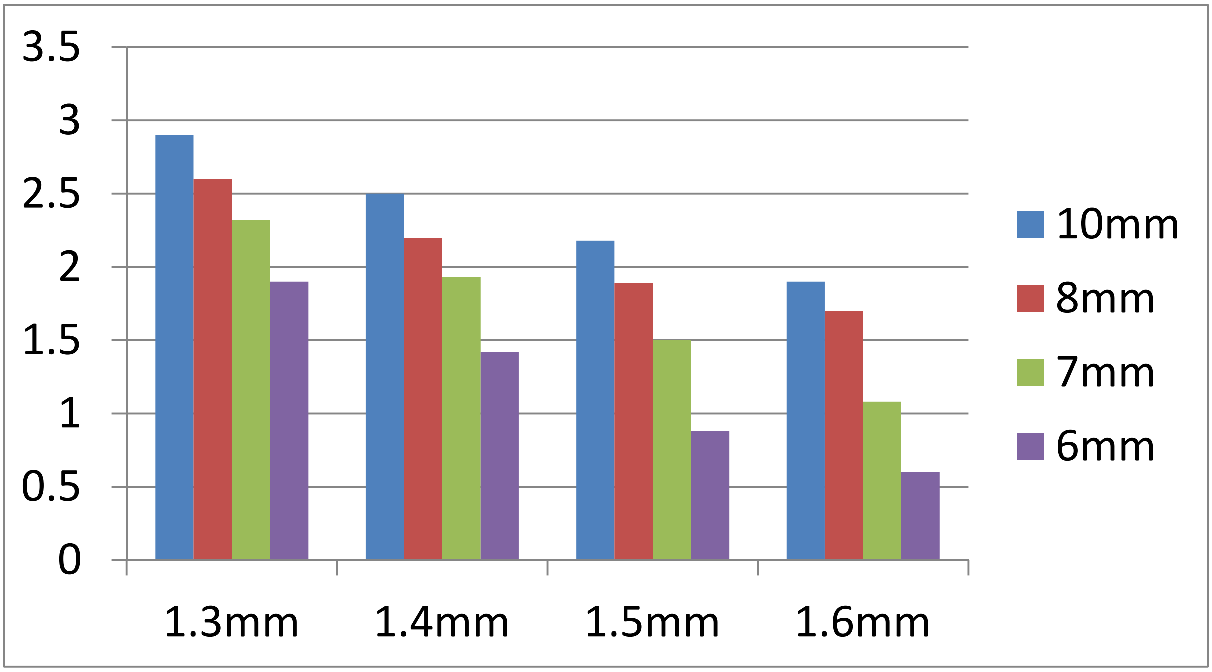
Graph 7- Deflection for length 7mm mini implants



Graph 8- Deflection for length 6mm mini implants



Graph 9- Overall comparison of deflection of mini implants



DISCUSSION

Over the last decade, the use of mini implants for the purpose orthodontic anchorage has increased considerably. For an implant to achieve its goal, the selection of a mini implant of adequate length and diameter best suited to the required area is of prime importance. Various authors like **Kyung et al**[70] and **Park et al**[71] have proposed dimensions of implants to be used in different areas of the jaws. Hence in this study the commonly used dimensions of implants have been used for evaluation and comparison of deflection.

Initially implants were manufactured using cobalt-chromium alloys, but the use of this material was soon discarded due to adverse bone reactions noted. Stainless steel implants are biocompatible with a high Young's modulus of 185GPa[72] and are thus less prone to bending but contact area with bone is reduced and it also interferes with magnetic resonance imaging and computed tomographic investigations.

The biocompatibility and direct bone contact with pure titanium implants has been clearly demonstrated previously but it has a lower yield strength of 180MPa, tensile strength of 290MPa and hardness compared to titanium alloys which have a yield strength of 830MPa and tensile strength of 900MPa[72]. This also permits filigree structures like the turn of the threads to be worked out solidly. Consequently, most orthodontic mini implants in use currently are made of Grade 5 titanium (Ti-6Al-4V) and thus this was the implant material chosen for the present study.

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**[51] 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**[53] and **Lin et al**[64] also 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. Hence the insertion angle was chosen as 90° for the present study.

In the evaluation of the biomechanical performance of screws, methods such as insertional torque and axial pull out tests are the most often used in orthopedics and oral and maxillofacial surgery. **Motoyoshi et al**[20] found the average torque measured at placement to be between 8.3 Ncm in the maxilla and 10 Ncm in the mandible and said that screws placed with maximum torque in the range of 5-10 Ncm had the highest rate of success. Higher torque levels are associated with ischemia and necrosis of surrounding bone and low insertion torques are associated with inadequate primary stability of implants. Thus the optimal insertion torque was set at 1kgf (i.e 9.8Ncm) in this study.

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 loosen and fracture. When mini implants of different diameters produced by the same manufacturer were compared by **Pithon et al**[28], 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[73].

Numerous authors like **Park et al**[25], **Motoyoshi et al**[20] and **Farnsworth et al**[46] have investigated the cortical bone thickness in various areas of the jaws. **Schnelle et al**[13] and **Hu et al**[33] have determined the availability of inter-radicular bone for mini implant placement. Bovine rib bones were chosen for the study as other authors like **Chatzigianni et al**[44] and **Laurito et al**[39] have demonstrated the similarity of architecture of bovine rib bone to human mandible. Hounsfield units of cortical bone in an average human mandible have been observed to be 1400-1600 with a medullary reading of 400-600 Hounsfield units. The cortical bone in bovine ribs has demonstrated to be 1400 Hounsfield units and medullary bone to be 470 Hounsfield units.

In our study, the results of the 80 samples were divided into 2 groups:

(a) Effect of length on deflection with constant diameter

(b) Effect of diameter on deflection with constant length

Irrespective of the size, all the mini implants showed deflection in varying degrees upon insertion into the bone. The test of between subjects effect showed that individual effect of varying length and diameter and also the combined effect of varying both diameter and length on the degree of deflection was statistically significant (refer table 2)

The overall comparison of the first group showed that when the diameter was kept constant, there was a statistically significant progressive decrease in mean deflection with a decrease in length from 10mm, 8mm, 7mm and 6mm implants. This phenomenon was observed for all the implants of diameters 1.3mm to 1.6mm (refer tables 1, 3, 4, 5 and graphs 1-4).

In the second group, keeping the length constant it can be seen that there is a decrease in mean deflection with increase in diameter from 1.3mm to 1.6mm. This progressive decrease in deflection was observed to be statistically significant for all the various lengths of implants used in the study i.e. 10mm, 8mm, 7mm and 6mm (Refer table 1,6,7,8 and graphs 5-8).

It has been known that a change in length or diameter can alter the strength of a material. The strength of a material is directly proportional to the fourth power of its diameter and inversely proportional to the cube of its length[74]. Thus the stronger the implant, the greater is its ability to resist deflection.

The overall comparison of results of this in vitro study are in agreement with the above principle as the result demonstrates that there is a direct relation of the deflection of the implant on its length . In addition, the deflection of the implant is seen to be inversely proportional to its diameter. Hence the greatest deflection was observed for mini implants with the least diameter and longest length i.e. 1.3 x 10mm and the least deflection was experienced by the widest and shortest mini implants i.e. 1.6 x 6mm.

As shown in graph 9, similarities in deflection values can be observed for the various sizes of implants: 1.3 x 6mm, 1.4 x 7mm, 1.5 x 8mm, 1.6 x 10mm. Also 1.3 x 7mm, 1.4 x 8mm and 1.5 x 10mm mini implants are seen to exhibit similar deflections. Since all the implants were inserted into identical bone with a constant insertion torque and are all of the same material, the only factor responsible for the similarities between groups is the interplay of length and diameter. A change in the diameter of the implant is compensated by the change in its length to produce similar deflections for the various sizes of implants used in this study.

In a study done by **Miyajima et al**[75] the following elasticity coefficients were observed for cortical bone, spongy bone and titanium alloy implants: 1.4 x 10^4MPa, 7.9 x 10^3MPa and 1.1 x 10^4MPa 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 in responsible for deflection of the mini implant which is exhibited in our study.

In our study also the deflection was observed at the point of entry of the mini implant into bone. **Singh et al**[62] 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**[55] also who stated that the point of entry of the implant into the cortical bone acts as a pivot for its bending.

Similar to our study, **Kalra et al**[69] also found angular deviation of mini implant from ideal path in their in-vivo study. In addition they also found deviation from the point of entry of mini implants into bone. Contrary to our study, **Meyer et al**[35] found angular deviation between stent placement position and implant after insertion but said that this difference was not significant. However, they used prosthetic implants which were placed in edentulous areas and hence there were minimal chances of contact with adjacent teeth.

Having evaluated the deflection characteristics of various implants used in the study, the clinical implications of the same can be considered. Prior to implant placement, numerous factors liken the amount of available bone in the particular area, the presence of sinus, nerve canal and proximity to roots of adjacent teeth is examined. This is done using investigative tools like radiographs or computed tomographic techniques.

Although, it is seen that choosing a wider diameter implant would be beneficial in terms of ensuring a higher success rate [44] and a lesser degree of deflection as seen by our study, selecting an implant for a particular area is largely dependent on the amount of available bone in that region. **Poggio et al**[3] and **Alrbata et al**[68] have proposed that a minimum of 1mm bone thickness surrounding the mini implant is necessary to ensure its stability. Hence in areas where inter radicular bone availability is less mini implants of smaller diameter can be chosen.

This will also decrease the failure rates of mini implants as **Kuroda et al**[6] have proven that root proximity is one of the major risk factors. However it must be borne in mind that a decrease in diameter will lead to an increase in deflection as shown by our study and also weaken the implant. **Wilmes et al**[49] in their study concluded that the risk of mini implant fracture should be borne in mind at the time of insertion especially if mini implants of small diameters are employed.

Lee et al[76] have proposed that the torsional strength of a screw is directly proportional to the cube of its diameter. This is similar to the results of the study done by **Barros et al**[43] who found that increase in mini implant diameters significantly influences the placement torque and fracture torque on quantities that progressively reduced the fracture risk. This in turn can be co-related to the ability of an implant to withstand loading without fear of failure. **Park et al**[18] recommended loads between 150 and 350 cN and **Chatzigianni et al**[44] found greater screw displacement with increase in force levels. **Kanie et al**[11] have shown that when an implant is malleable, deformation and fracture could occur easily. This study concurs with our study which shows that the thinnest and longest mini implants show the greatest deflection.

In areas where cortical bone is thick, reducing the length of the mini implant will help reduce chances of failure by decreasing the amount of deflection as exhibited by this study. Longer and thinner implants are also seen to be more prone to bending and breakage. Thus reducing the length will also ensure that the insertion torque stays within the optimal range of 5 to 10 Ncm as studies by **Pithon et al**[28, 65] have shown that increasing the length of the screw causes an increase in the torque required for its insertion.

This is a factor in preferring shorter length implants predominantly in the mandible where the cortical bone thickness is inherently thicker than the maxilla. In our study also, longer mini implants showed greater deflection when compared to the shorter mini implants. Another alternative as proposed by **Melsen**[15] is that even with the use of self-drilling screws, pilot drilling may be required if cortical bone thickness is greater than 2mm as the dense bone can bend the fine tip of the screw.

Two mini implants fractured during the study during insertion. This was due to over tightening of the screw during placement. The insertion torque exceeded the set value of 1Kgf. Hence caution and care is advised during implant insertion to decrease fracture rates of mini implants.

Before placement of mini implant, the behaviour of the implant due to its interaction with bone even prior to loading needs to be considered. The importance of the biomechanical behaviour of various lengths and diameters of mini implants used in the study has been evaluated. The above study highlights the fact that when an implant is being placed into bone with increased cortical thickness such as the mandible, a shorter and wider implant needs to be used. Both the mechanical properties of the material in use as well as anatomical constraints in choice of implant to withstand the load applied are of prime importance. Hence, this study will enable the practitioner to select a proper mini implant from his/her available armamentarium for the right anatomical location.

SUMMARY & CONCLUSION

An in vitro study was conducted using self-drilling titanium alloy mini implants. Eighty implants of various diameters and lengths were used. The mini implants were inserted into bovine rib segments perpendicularly, and were then radiographically evaluated to determine the difference in deflection due to the bias in length and diameter. Previous studies on mechanical behaviour of mini implants have evaluated factors like torsional strength, displacement on loading, effect of diameter on fracture risk and stress concentration in different parts of the implant. The present study considered the phenomenon of deviation of the mini implant due to its interaction with cortical bone.

On the basis of the results, when an implant is inserted into a bone of increased cortical thickness, the following inferences can be obtained:

- Deflection of the mini implant does occur upon insertion.
- Increasing the diameter of the implant decreases the amount of deflection.
- Decreasing the length of the mini implant causes a decrease in deflection.
- Similarities in deflections of mini implants are caused due to the interplay between length and diameter of the mini implant.

In an era where the usage of skeletal anchorage for effecting tooth movement is exponentially increasing, the clinical significance of choosing the right armamentarium needs to be considered. This study demonstrates the behaviour of various implants upon insertion prior to loading.

Selecting a proper implant depends on anatomical limitations like cortical bone thickness, proximity to adjacent roots, or any other vital structures. It also depends on the mechanical property of the material of the implants, the minimum length and diameter required to withstand the forces applied on it. Hence an effective balance has to be maintained between the two to ensure high success of treatment.

This study will enable the clinician to make a judicious choice after weighing the pros and cons of selecting a particular dimension of implant. When an implant is planned for insertion into thicker bone such as the mandible, it is preferable to use a thicker and shorter mini implants as they exhibit lesser deflection. In areas of lesser cortical bone, a thinner and longer mini implant can be considered as the resistance offered by the bone will be lesser.

However the above study is an in vitro study hence the exact clinical scenario cannot be simulated. Further studies on a larger sample scale may be needed to validate the results obtained.

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