COMPARISON OF INSERTION ANGULATIONS MEASURED VIA BONE-TO-IMPLANT CONTACT OF MINI-SCREWS USING MICRO-CT

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ABSTRACT

Bone-to-implant contact of orthodontic mini-screws has been determined via Micro-CT to be a strong predictor of primary stability. Various insertion angulations, including both 90^0 and 50^0, have been reported as ideal for providing optimum primary stability. The aim of this investigation was to determine if a statistically significant difference exists in the bone-to-implant contact of mini-screws placed with an insertion angulation of 90^0 compared to those placed at 50^0 as determined via Micro-CT.

Ten self-drilling, self-tapping orthodontic mini-screws (Aarhus,1.5mmx6mm) (n=5) were inserted into the posterior ramus of an adult pig mandible, an analog to an adult human mandible. A custom stent with ten holes, five at 90^0 and five at 50^0, was fitted to the bone surface to control insertion angulation. The bone was cut to 1.5x1.5x1cm segments and scanned using SkyScan 1127 with ideal specifications (8mm pixel size, medium camera, 80Kv, 100mA, 10W, 180^0 rotation, and 0.5Al+0.25Cu filter). The raw scans were reconstructed using NReconV1.6.10 and these datasets were then reoriented using DataviewerV1.5.2 along the Z-axis to standardize the peri-implant bone for analysis. A custom task-list was used with CT-AnalyzerV1.14.41 to determine the percent of bone-to-implant contact per mini-screw.

A Mann-Whitney U test indicated that the bone-to-implant contact was not statistically significantly different for the mini-screws placed at 90^0 (Mdn= 72.34) compared to the mini-screws placed at 50^0 (Mdn= 53.25), U=5, p=.1443. Therefore, the results do not significantly differ at $p<.05$. There is no statistically significant difference in the bone-to-implant contact between the Aarhus 1.5mmx6mm orthodontic mini-screws placed at 90^0 compared to those placed at 50^0 as measured by Micro-CT. This may lead to

the conclusion that there is no significant difference in the primary stability of Aarhus mini-screws placed at 90^0 and 50^0.

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TABLE OF CONTENTS

Page

ABSTRACT…………………………………………………………………………….I

ACKNOWLEDGEMENTS…………………………………………………………...III

LIST OF FIGURES……………………………………………………………….. VI

LIST OF TABLES……………………………………………………………….. VII

CHAPTERS

1. INTRODUCTION……………………………………………………….........1

2. REVIEW OF THE LITERATURE……………………………………………...3

    2.1. History of Orthodontic Anchorage……………………………………………3

    2.2. History of Dental Implants……………………………………………………5

    2.3. History of Orthodontic Mini-Screws…………………………………………..6

    2.4. Mini-Screw Specifics…………………………………………………………..9

    2.5. Static Factors of Mini-Screw Primary Stability………………………………10

    2.6. Dynamic Factors of Mini-Screw Primary Stability……………………………..11

    2.7. Mini-Screw Insertion Angulation……………………………………………13

    2.8. Bone Physiology……………………………………………………………...14

    2.9. Measurements of Mini-Screw Primary Stability……………………………..16

    2.10. Micro-Computed Tomography (Micro-CT)……………..……………………17

3. AIMS OF THE INVESTIGATION…………………………………………………19

4. MATERIALS AND METHODS……………………………………………………20

4.1. Introduction……………………………………………………………………...20

4.2. Obtaining the Micro-CT scans……………………………………………………..22

4.3. Reconstruction……………………………………………………………………..24

4.4. Reorientation………………………………………………………………………24

4.5. Selecting the Volume of Interest (VOI)……………………………………………26

4.6. Analysis of Bone-to-Implant Contact……………………………………………...28

5. RESULTS…………………………………………………………………………30

5.1. Statistical Analysis…………………………………………………………………30

5.2. Quantitative Analysis………………………………………………………………32

6. DISCUSSION……………………………………………………………………..33

6.1. Results in Comparison to Current Literature………………………………………33

6.2. Specific Findings and Clinical Significance ………………………………………34

6.3. Limitations of the Study and Future Considerations……………………………35

7. CONCLUSIONS…………………………………………………………………37

REFERENCES………………………………………………………………………..38

LIST OF FIGURES

Figure Page

1. Stone model of pig ramus with baseplate wax-up for stent……………..………20
2. Acrylic stent seated on pig ramus with 50^0 and 90^0 insertion holes……………..20
3. Pig mandible anatomy indicating location of mini-screw placement……………21
4. Ten samples of mini-screw and bone separated into cubes……………………...22
5. Scanning specifications for camera size, rotation, frame averaging, and random movement…………………………………………………………………...…23
6. Reconstruction specifications for smoothing, ring artifact reduction, and bean hardening correction……………………………………………………………..24
7. Initial unaligned mini-screw in X,Y,Z planes……………………………………25
8. Reoriented mini-screw aligned orthogonal with the Z axis……………………...26
9. Reoriented data-set loaded into the CT-Analyzer V1.14.4.1 software………..…27
10. Selection of region of interest to include peri-implant bone…………………..….27
11. Samples loaded with custom task list for analysis…………………………….…29
12. Data set visual analysis…………………………………………………….……….30
13. Raw image sample 9…………………………………………………………..…….31
14. Raw image sample 3…………………………………………………………….….31

LIST OF TABLES

Table Page

1. Bone to Implant Contact……………………………………………………..……..32

CHAPTER 1

INTRODUCTION

Anchorage is an integral part of successful orthodontic treatment. Thomas D. Creekmore, DDS, said “tooth borne anchorage is one of the greatest limitations of modern orthodontic treatment” (Creekmore, 1983). Historically, the most commonly used alternative to tooth borne anchorage was extra-oral anchorage, which is dependent on excellent patient cooperation. However, with the increase in adult patients, esthetic concerns, and overall decrease in patient compliance an increased control over unwanted reactive forces has been accomplished with the help of orthodontic mini-screws. Orthodontic mini-screws have a variable success rate ranging from 59.2% to 98% depending on the study. There are numerous factors that can affect this success rate including orthodontist experience, the patient’s oral environment including gingival thickness and bone mineral density, physical characteristics of the mini-screw including length, pitch, depth, and dynamic factors including insertion site and angulation (Decoster, 1990).

The effect of insertion angulation on the success rate, also called the primary stability, of orthodontic mini-screws has been studied by various destructive methods including histomorphometry and pull-out tests, but recently micro-CT has been used (Sakin, 2013). Micro-CT is a non-destructive method to measure mini-screw primary stability via three dimensional radiographs that allows for analysis of the amount of bone contacting the mini-screw surface (Donnelly, 2011). This bone-to-implant contact (BIC) has been directly correlated to primary stability. A recent study completed by Dr. Epshteyn entitled, “Optimal Specifications for Measuring Bone-to-Implant Contact of Mini-Screws

Using Micro-CT" determined the optimal scanning specifications for this data to be retrieved. Thanks to that pilot study, other confounding variables, such as mini-screw insertion angulation can be accurately compared via micro-CT scanning (Epshteyn, 2016).

CHAPTER 2

REVIEW OF THE LITERATURE

2.1 History of Orthodontic Anchorage

Isaac Newton's third law, which states, "when one body exerts a force on a second body, the second body simultaneously exerts a force equal in magnitude and opposite in direction on the first body," is the basis for orthodontic anchorage. Orthodontists are able to expertly move teeth by producing specific forces and moments. As explained by Newton, each orthodontic force has an equal and opposite reaction that must be controlled to either compliment the treatment or be diverted to avoid unwanted tooth movement. Orthodontic anchorage is defined by Daskalogiannakis as "resistance to unwanted tooth movement," and can be accomplished by a number of ways (Daskalogiannakis, 2000). Anchorage can be provided by the teeth, periodontal structures, the head, neck, and more recently, the alveolar bone.

The magnitude and therefore type of anchorage needed in a particular treatment is influenced by multiple factors including "size of root surface, attachment level, density and structure of alveolar bone, periodontal reactivity, muscular activity, occlusal forces, craniofacial morphology and the nature of the tooth movement resulting from the planned correction" (Diedrich, 1993).

Gianelly and Goldman suggested the terms maximum, moderate, and minimum to rank types of anchorage based off of how much the active and reactive units of teeth should move when a force is applied. According to his original terminology, in a maximum anchorage system, the anchor should remain in place. For example, extra-oral headgear or a Hayes Nance should maintain the anchorage unit in its original position. Minimum anchorage, also commonly referred to as reciprocal, is when the reaction force should move the teeth at least to the same extent as the action force. For example, closing a maxillary diastema and moving each central incisors equidistant amounts. Moderate anchorage applies to any situation between these two. Examples of moderate anchorage use of a trans-palatal arch while retracting maxillary anteriors, using a lower lingual holding arch to retract lower canines, or use of differential tooth mass to properly manage space closure (Gianelly, 1971).

Orthodontic treatment without sufficient anchorage can result in compromised results (Melsen, 1997). With the increase in social media and decrease of paternalism in the medical field compliance with orthodontic appliances has decreased. A study completed in 2006 in which a timing device was placed in the headgear appliance, showed that although patients reported wearing the headgear a mean of 13.6hrs/day they were actually only wearing it a mean of 5.6hr/day (Brandao, 2006).

Orthodontic mini-screws have been defined as “a device temporarily fixed to bone for the purpose of enhancing orthodontic anchorage by either supporting the teeth of the reactive unit or by obviating the need for the reactive unit all together, and is subsequently removed after use” (Singh, 2010). Orthodontic mini-screws provide the advantage of anchorage control without depending on patient compliance. Mini-screws allow for a new magnitude of anchorage called skeletal or “absolute.” Mini-screws will not have “absolute” anchorage if the maximum force is above their maximum failure force, which, according to the literature, ranges from 50g (0.5N) to 450g (4.5N) (Cuijpers, 2014). Mini-screws can be used as direct anchorage where the reactive force directly effects the mini-screw or as indirect anchorage where the mini-screws are used as reinforcement to the reactive unit of teeth to reduce the negative effect of the reactive force. As long as the mini-screw is successful, both instances allow for more controlled and therefore predictable orthodontic treatment.

2.2 History of Dental Implants

The first evidence of dental implants was attributed to the Mayans who, around 600 AD, placed shells into the alveolar bone to replace teeth (Weinberger, 1926). In the 1930's, orthopedic surgeons were using chromium-cobalt alloy screws in hip surgeries. After observing this, the Strock brothers experimented by placing these screws into the mandibles of dogs and humans to replace individual teeth. Huge advances occurred in the 1950's when Dr. Branemark accidently discovered the concept of osseointegration. He placed titanium implants into rabbit femurs to observe the blood flow and healing and discovered that he was unable to remove the implants because "the metal had fused to the bone." He called this process osseointegration and concluded that titanium could be used to create an anchor for artificial teeth (Abraham, 2014). Presently, mandibular dental implants have reached a success rate of up to 95% but they depend on a three month healing period in order for osseointegration to complete.

Osseointegration has been defined as "a direct structural and functional connection between ordered, living bone and the surface of a load-carrying implant" (Parithimarkalaignan, 2013). Osseointegration must be achieved prior to implant loading and is a requirement for long term implant success. There are mechanical properties involving the alveolar bone locking around the implant, followed by biological fixation where the bone remodels around the implant. Osseointegration is a time dependent process and the histological results resemble a “functional ankyloses” (Zarb, 1985). Parithimarkalaignan et al describes six components involved in achieving dental implant osseointegration as follows: 1) biocompatible material 2) macroscopic and microscopic implant surface design 3) bone quality 4) surgical technique 5) undisturbed healing phase 6) loading conditions.

Dental implants are made from Titanium because it is highly biocompatible: it is nontoxic to macrophages and is not rejected by the human body. Titanium forms an oxide layer which has the ability to absorb calcium and phosphate ions leading to its ability to osseointegrate. Surface roughening and acid treatment increases the thickness of the fragile oxide layer, increasing the success of osseointegration (Liu, 2017).

Most orthodontic mini-screws are also made from Titanium, however a major difference is that mini-screws have a smooth surface because they are not intended to osseointegrate. Even with this smooth surface, one study, which histologically examined immediately loaded orthodontic mini-screws maintained in beagle dogs for at least 6 months, found that there can be small islets of osseointegration on orthodontic mini-screws (Vannet, 2007).

2.3 History of Orthodontic Mini-Screws

The requirement for skeletal anchorage grew from the increase in adult patients seeking orthodontic care. Many of these patients presented with periodontal disease, missing teeth, as well as esthetic concerns. The standard methods of anchorage, such as extra-oral appliances or differential tooth mass were not providing sufficient anchorage for complicated tooth movements. In addition to complicated tooth movements, orthodontic mini-screws are indicated to be used as anchorage in patients in preparation for prosthetic treatment who have "absence of multiple teeth, reduction of periodontal support, and inadequate occlusion [which] prevents the use of conventional anchorage" (Melson 2004).

The evolution of orthodontic mini-screws is related to the advances in dental and surgical implants. As the use of dental implants increased for prosthetic reconstruction, these smaller implants, or mini-screws, were being developed for use in retro-molar areas providing indirect anchorage in orthodontic patients with missing teeth (Melson, 2005). On the other side were the advancing surgical implants which were developed with the idea of direct anchorage by reducing patient compliance on headgear and elastics. Oral surgeons had difficulties with post-surgical retention using only tooth borne anchorage, and the advent of surgical screws and pins as skeletal anchorage as an adjunct to post-surgical fixation increased retention. With these ideas in mind, Creekmore and Eklund inserted a surgical vitallium bone screw just occlusal to the anterior nasal spine and continuously tied it with elastic thread to an orthodontic arch wire in a deep bite patient in order to intrude the maxillary anterior teeth 6mm within one year. Their innovative case study sparked other research on the possible uses of skeletal anchorage and orthodontic tooth movement (Creekmore, 1983).

The first documented mini-screws specifically designed for orthodontic anchorage was described by Kanomi in 1997. He described characteristics that differentiate orthodontic mini-screws from dental implants, such as 1) small enough to be inserted in a variety of locations within the alveolus, 2) can be placed with a simple surgical procedure by an orthodontist, and 3) easy removal (Kanomi, 1997). Since Kanomi, other variables that differentiate orthodontic mini-screws from prosthetic implants have been determined and include that orthodontic mini-screws can have immediate force application, that they rely on primary stability rather than osseointegration, and that their maximum failure force is dependent on primary stability rather than osseointegration.

One major difference between orthodontic mini-screws and dental implants is that orthodontic mini-screws are meant to be immediately loaded. This idea was tested by Dr. Birte Melsen who placed 8mm orthodontic mini-screws in adult male macaca fascicularis monkeys and immediately applied either a 25g or 50g continuous spring force. The mini-screws were removed after 1, 2, 4, and 6 months, and after histological examination, it was determined that the type of bone the mini-screws were inserted into and their location affected their stability, but the force level applied did not affect stability. This study supports the idea that as long as the force does not exceed the maximum failure force of the primary stability, immediate loading of orthodontic mini-screws is successful (Melsen 2000).

In contrast to dental implants, orthodontic mini-screws are meant to be used temporarily as anchorage. In order to be temporary, they should not osseointegrate and must be able to be removed with appropriate ease. Due to these reasons, the stability of orthodontic mini-screws depends not on osseointegration but on primary stability.

Primary stability has been defined as “the mechanical interlock between bone and implant.” It is also defined as the “implant stability immediately after insertion” (Araghbidikashani, 2016). This primary stability is related to the stiffness of the bone, the implant, and the amount of intersection of the implant and bone. This intersection is called the bone-to-implant contact (BIC) (Gedrange, 2005). Due to their dependence on primary stability, mini-screws are able to be loaded immediately, compared to prosthodontic implants that require a "healing time of three to six months before functional loading because function allows micro-motion, which permits fibrous tissue growth and subsequent [implant] failure" (Cope, 2004).

Another difference between orthodontic mini-screws and prosthodontic implants is that the maximum failure force for osseointegrated implants is proportional to the quantity of osseointegration whereas for orthodontic mini-screws the maximum failure force is proportional to the bone-to-implant contact. There are many factors affecting the amount of bone-to-implant contact, which ultimately effects the mini-screw's primary stability. These include static factors, such as the quality and quantity of bone at the insertion site, as well as the physical and dynamic characteristics of mini-screw placement (Migliorati, 2012).

2.4 Mini-Screw Specifics

The mini-screws used in this study have four distinct anatomic zones, including a head, neck, collar and body. The head is the working part visible to the orthodontist once inserted, and can have different configurations. The neck is a smooth transitional area that joins the head and the collar. It can also serve functionally to allow for ligation. The soft tissue collar can be different lengths, to allow for different thicknesses of gingiva based on patient populations. It is wider than the body, allowing it to serve as a stop during insertion. The body is the portion of the screw inserted into the bone, and can vary in length, diameter, pitch, thread shape factor, and shape. Mini-screws are usually made from a titanium alloy, as discussed previously.

The mini-screws used in this study are self-drilling; the actual tip of the mini-screw is what cuts through the bone. Other types of mini-screws are self-tapping, meaning that they require a pre-drilling their full length at a slightly small diameter. The disadvantages with a self-tapping system include the loss of tactile sensation during insertion and a decrease in primary stability due to bone loss during the pre-drilling (Melson, 2013).

2.5 Static Factors of Mini-Screw Primary Stability

Some of the physical characteristics of the mini-screw include pitch, length, diameter, and thread shape factor. Petrey et al. found that mini-screws with increased length (comparing 6mm, 8mm, and 10mm) had greater primary stability (Petrey, 2010). The pitch, defined as the vertical distance between the threads, has been shown to be a key factor in primary stability. DeCoster et al studied implant design via pullout tests in synthetic bone. He found that the pull out force increased as the major diameter increased and as the pitch decreased (Decoster, 1990).

Another characteristic that varies between mini-screws is the thread shape factor (TSF). TSF is the ratio of mean thread depth to pitch (D/P) expressed as a percentage. Migliorati et al determined the importance of thread shape factor by first measuring the thread depth and pitch of different mini-screws via scanning electron microscope and then, after controlling for other variables, completed pullout tests to compare the maximum failure force. He concluded that mini-screws with higher TSF values provided greater primary stability (Migliorati, 2011, 2012).

Another static factor affecting the primary stability of the mini-screw is its shape. Pithon et al compared the pull out strength of conical and cylindrical-shaped mini-screws placed in pig mandibles and found that the conical-shaped mini-screws had lower maximum failure force due to displacement of the bone to a broader region. The cylindrical-shaped mini-screws had increased osseous contact (bone-to-implant contact) therefore showing increased pull out strength indicating greater primary stability (Pithon, 2012). In contrast, Yano et al found that when histologically analyzed via scanning electron microscope, conical mini-screws had increased bone-to-implant contact compared to cylindrical mini-screws (Yano, 2006). In agreement with Yano's findings is a study completed by Florvaag et al. which compared cylindrical and conical implants via pull out tests placed in pig femoral heads with the same bone mineral density. They found the conical mini-screws were able to withstand statistically significantly greater pull out forces (Florvaag, 2010).

2.6 Dynamic Factors of Mini-Screw Primary Stability

In addition to all the static factor affecting primary stability of mini-screws, there are also dynamic factors such as insertion technique and clinician experience. One of the main dynamic factors is placing a self-tapping mini-screw with a pilot hole or the use of a self-drilling mini-screw. The pilot hole associated with self-tapping mini-screws has been correlated with increased clinical problems, including increased mini-screw failure defined by loosening or complete loss of the mini-screw. Chapman et al. found a statistically significant reduction in pullout force for screws placed with a pilot hole compared to those that were self-drilling. He found that bone was lost from the pilot hole drilling process so the resulting bone-to-implant contact was decreased (Chapman, 1996). Pithon et al. found that the self-tapping mini-screws had decreased primary stability because they tend to have a decreased thread shape factor (Pithon, 2012). Yano et al. assessed the bone-to-implant contact histologically using scanning electron microscope with both self-drilling and self-tapping mini-implants. He discovered that for self-tapping mini-screws the BIC increased when an eight week healing period was allowed between the pilot hole and the mini-screw placement (Yano, 2006). Given the orthodontic requirement of immediate loading of mini-screws, self-drilling is better due to the increased primary stability, decreased chance of heat induced bone necrosis, and decreased healing time to ensure adequate primary stability (Pithon, 2012).

Lim et al reported that the insertion site and the clinician's experience are the most influential variables in the successful use of mini-screws. Lim looked at 407 mini-screws placed by 17 orthodontic residents and looked at the success at one-week post-insertion. This study determined that operator induced loosening of the mini-screw from "wobbling" during insertion decreases significantly once the clinician has placed 20 mini-screws (Lim 2011).

2.7 Mini-Screw Insertion Angulation

Insertion angulation of the mini-screw is a relevant clinical factor that remains unclear. Petrey et al studied the failure force of mini-screws, defined as the magnitude of force that can be placed on a mini-screw before it fails. He found that mini-screws inserted at 90^0 had a greater failure force compared to those inserted at 45^0 and 135^0. They found no significant difference in the failure force between the 45^0 and 135^0 insertion angles. However, this pull out test was completed tangent to the mini-screw, regardless of its insertion angulation. Often times when using mini-screws as anchorage, the force is not tangent to the insertion angulation (Petrey, 2010).

Alternatively, Park et al studied failure rates based on mobility during active treatment. He found no significant differences between mini-screws placed anywhere from 10^0 to 90^0, but did note that mini-screws placed at an angle (i.e. not at 90^0) could be longer in length due to the Pythagorean Theorem, which states that the sum of the squares of the lengths of each of the triangle's legs is the same as the square of the length of the triangle's hypotenuse. A longer mini-screw can fit within the alveolar bone when placed at an angle and Park, along with many other studies, have agreed that increased length is directly correlated with increased primary stability (Park, 2006).

From an oral surgery perspective, Uckan et al studied the insertion angulation of screws for mandibular fixation after a 5mm mandibular advancement in sheep via sagittal split ramus osteotomy. Previously, surgeons could place three screws either at 90^0 extraorally or at 60^0 intraorally. Uckan et al showed there is no difference in the rigidity of the fixation with the insertion angulation because screws inserted perpendicular to the bone

(90^0) provide approximately 20% more resistance to displacement of the segments and screws inserted at 60^0 have a greater contact area with the bone (Uckan, 2001).

Pickard et al was the first to study the insertion angulation of mini-screws in relation to shear force, which is more representative of clinical orthodontic forces. They placed mini-screws at 45^0 in the direction of the shear force and 45^0 opposing the shear force (similar to a tent peg) in the buccal surface of cadaver human mandibles. The mini-screws angled in the same direction as the shear force had a significantly greater stability and resistance to failure compared to the mini-screws angled away from the shear force. This is clinically significant because orthodontists may want to consider the direction of force applied to their mini-screw prior to placement. Pickard et al also studied the maximum pull out force along the long axis for mini-screws placed at 90^0 and 45^0 and found the 90^0 group to have statistically significant higher failure force (341N vs. 141N). Both of these studies indicate that mini-screws can withstand greater force when the force is applied along the long axis, however most orthodontic forces on mini-screws are shear forces (Pickard, 2010).

2.8 Bone Physiology

Piau Pigs (Sus scropha) was the experimental animal model used in this study because of the features similar to humans. As outlined by Glowacki et al. the minipig and human share similarities including mandibular anatomy, temporomandibular function, chewing patterns, and bone turnover rates (Glowacki, 2004). Nkenke et al compared the bone-to-implant contact and peri-implant bone density in immediately loaded dental implants with traditional dental implants in pig mandibles (Nkenke, 2003).

Pithon et al placed mini-screws in various locations within the pig maxilla and mandible and tested primary stability via pull out tests. He found the highest pull out values in the upper and lower molar areas and the lowest value in the mid-palatal suture. He measured the mean cortical thickness of the sites and found the palatal suture to have a cortical thickness of 0.92+/-0.13mm while cortical thickness of the lower molar area is 2.24+/-0.08mm. It was concluded that mean cortical thickness is directly correlated to mini-screw primary stability. Studies by Dalstra el al agreed with Pithon that the thickness of the cortical bone is the most important factor for mini-screw primary stability. In contrast, studies by Lim et al found the insertion site with the greatest success was the palatal suture. Presently there is not a general consensus for the insertion location with the greatest primary stability. In humans there are multiple other factors affecting success including hygiene, periodontal thickness, and all of the static and dynamic factors previously discussed (Dalstra, 2004).

Lekholm et al have categorized human bone into four groups: Type 1 is almost entirely composed of homogenous cortical bone, Type 2 is mostly cortical bone with an inner section of trabecular bone, Type 3 is a thin layer of cortical bone surrounding a large core of dense trabecular bone, and Type 4 is a thin layer of cortical bone surrounding a core of weak, low density trabecular bone. It has been agreed upon that in both the mandible and maxilla the cortical thickness and trabecular density tends to decrease as you move posteriorly. It has been correlated that the success rate of implants decreases as the Type of bone increases (Lekholm, 1985).

2.9 Measurements of Mini-Screw Primary Stability

Mini-screw primary stability can be measured in multiple ways. One method is cutting torque resistance analysis, which is using the action of drilling a pilot hole for a self-tapping mini-screw to determine the hardness, and therefore bone mineral density of the insertion site. Friberg et al confirmed the correlation between the energy (J/mm^3) to cut through bone with the bone mineral density measured through microradiographs. This is not a very applicable method because it requires the use of self-tapping implants, and the result is not discovered until after the pilot hole is drilled (Friberg, 1995).

Insertion torque, the force required to insert the mini-screw, has been shown to be directly related to bone mineral density and positively correlated with primary stability. Removal torque value, which is measuring the force required to remove the mini-screw, has been correlated to the bone-to-implant contact but this method is less accurate due to the confounding variable of osseointegration, which begins immediately and is a continuous process. The initiation of osseointegration occurs with the transformation of the initial blood clot by phagocytes which peaks between day 1-3 post-surgery. Depending on how long the mini-screw is present in the bone prior to the removal torque test, various increases in torque value can be contributed to osseointegration (Johansson, 1991).

Another invasive method to measure primary stability is through the pull out test, which measures the force required to pull the mini-screw out of the bone along its long axis. Increased pull out force has been correlated with increased primary stability, increased bone mineral density, increase cortical bone thickness, and increased thread shape factor. The downside to pullout tests is that orthodontic forces on mini-screws are usually oblique, not typically along the long axis of the mini-screw (Migliorati, 2012).

Histomorphometry, which is a quantitative study of the form or shape of tissue, can also be used to measure bone to implant contact. With this method, samples of the implant and bone are ground down and analyzed under a microscope where the bone-to-implant contact can be quantified. Often a scanning electron microscope (SEM) is used, which is ideal for analyzing the surface topography. In this method, the number of osteocytes within a sample can be counted and then extrapolated to the rest of the sample. One limitation to the SEM is that it only produces 2D images and does not provide information on bone trabeculation or density (Cuijpers, 2014).

The micro computed tomography scanner (micro-CT) is a 3D x-ray on a small scale with increased resolution that can also be used to measure BIC. It allows for viewing the 3D internal structure of specimen without destruction. In the micro-CT scanner, the specimen rotates on a stage between the xray source and detector in order to replicate the microarchitecture with isotropic resolutions as small as 1 to 6μm (Donnelly, 2011).

2.10 Micro-Computed Tomography (Micro-CT)

Compared to other methods, the micro-CT is superior because it is a non-destructive technique that provides a three-dimensional method to study structures. Applied to mini-screws in bone, the micro-CT is useful because it allows for the study of the bone trabeculation and structure in addition to bone mineral density, which is often a variable factor playing an important role in the success of mini-screws. Compared to histomorphometry, micro-CT has increased accuracy in determining the thickness of the slices examined. Cuijpers et al compared the efficacy of measuring bone-to-implant contact, among other variables, using micro-CT, nano-CT, and histomorphometry. He concluded that both the micro-CT and histomophometry are efficacious. However, the micro-CT has the disadvantage of beam hardening and scatter which, in Cuijpers study, increased due to the high attenuation of the bone and the metal mini-screw. Fortunately, in the micro-CT reconstruction "Beam hardening correction" can be used which helps eliminate the dark shadows, streaks, and other artifacts that are obtained during the scanning. Professor Dominik Fleischmann from Stanford recommends scanning with increased kV in order to have a "harder X-ray beam, and thus less beam hardening artifacts" (Boas, 2012).

CHAPTER 3

AIMS OF THE INVESTIGATION

The aim of this study is to determine which angulation of mini-screw insertion will provide more primary stability using micro-CT technology. Specific aims include:

- To fabricate a stent to standardize the insertion angulations of the mini-screws
- To determine if the bone-to-implant contact of the mini-screws will be significantly different for those placed at 90^0 compared to those placed at 50^0
- To acquire images of titanium mini-screws inserted into bone using three-dimensional micro-computed tomography using ideal specifications determined from Epshteyn et al., "Optimal Specifications for Measuring Bone-to-Implant Contact of Mini-Screws Using Micro-CT.2016. These images will be reconstructed with NRecon and used to measure the bone-to-implant contact with CTAn software.
- To use the bone-to-implant contact measurements and compare them for statistical significance

Significance

The information gathered in this study may provide orthodontists with a clinically significant insertion angulation in which mini-screw primary stability is increased as measured by bone-to-implant contact.

CHAPTER 4

MATERIALS AND METHODS

4.1 Introduction

For this study, an adult pig mandible from Clemens Food Group in Hatfield PA served as an analog to the human mandible. An alginate impression was taken of the portion of the body of the mandible and poured in yellow lab stone. To fabricate the stent, the stone model was lubricated with Vaseline and three layers of hard baseplate wax were molded around the model. Ten holes were measured out 2cm apart in all directions and indented into the baseplate wax (Figure 1). With the help of Prosthodontist David Donatelli, DDS, a dental surveyor was used to accurately obtain the 90^0 and 50^0 insertion paths to the intaglio surface of the wax up. The wax up was sent to NewTech Dental Lab and invested into clear acrylic resin. Once returned, the stent was seated on the bone to confirm fit (Figure 2).

Figure 1: stone model

Figure 2: custom stent

Ten self-drilling orthodontic mini-screws from Aarhus (American Orthodontics, Sheboygan, Wisc, 1.5mm diameter and 6mm long) were inserted in the mandibular retromolar area from the lingual cortex through the stent until all threads were completely buried by bone (Figure 3). There were ten mini-screws, five inserted at each angulation: 90^0 and 50^0. Each mini-screw was cut from the mandible with a handsaw so the total sample 1cm x 1cm x 1.5cm to facilitate micro-CT scanning. These cube samples were placed in a sample holder with phosphate-buffered saline and kept refrigerated at 4^0 Celsius (Figure 4). Each sample was positioned in a holder with parafilm and scanned using micro-CT (SkyScan 1172; Skyscan, Aartselaar, Belgium) in the Skyskan Micro-CT Laboratory run by Dr. Mary Barbe at the Temple University School of Medicine. The scan specifications, 8mm pixel size, medium camera size, 80 Kv, 100mA, 10W, 180^0 rotation, and 0.5Al + 0.25Cu filter, were adopted from a previous Master's Thesis.

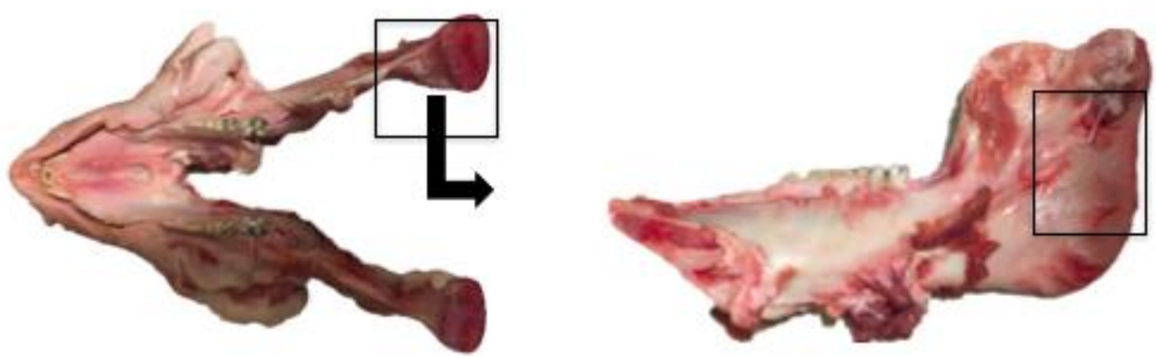

Figure 3: sample pig mandible

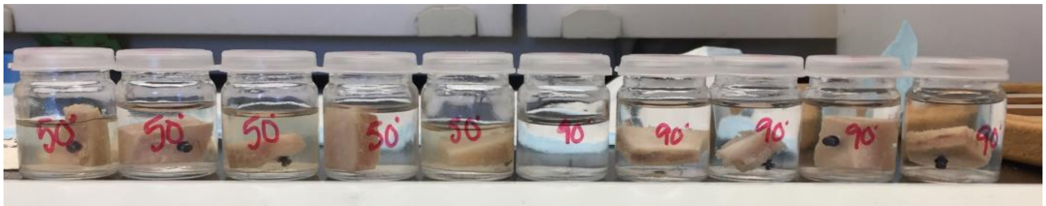

Figure 4: samples segmented in 1x1.5x1.5mm sections

4.2 Obtaining the Micro-CT Scans

Once the sample was ready for the micro-CT scanning, the SkyScan 1172 was turned on and the following specifications were set in the main pop-up window: voltage at 80kV, current at 100μA, power at 10W. At the bottom of the screen the remaining parameters are set including the 0.5Al+0.25Cu filter, 8μm pixel size, and medium camera. This 0.5Al+0.25Cu filter is chosen to reduce beam hardening by reducing the amount of low energy x-rays in the beam. The medium camera, also called the resolution level, was chosen as a balance between the spatial resolution and duration of the scan. The scanning mode was set up by ensuring the mean intensity of the empty field without flat field was ~60% and then the flat-field was corrected prior to placing the sample in the scanner. The scan was performed with 0.3 degrees of rotation, frame averaging of 6, and random movement of 10 (Figure 5). The small rotation step and frame averaging are to allow for a large number of projections which reduce noise.

Figure 5: 0.3 degrees of rotation, frame averaging of 6, random movement 10

4.3 Reconstruction

Each scan was reconstructed using NRecon V1.6.10. The specifications are seen under Figure 6. Smoothing was set to 2 and ring artifact was set to 6 which helped limit the amount of secondary ring artifact from the metal mini-screw. Lastly, beam-hardening correction was moved to 40%, which can change depending on the bone and the amount of fluid absorbed in the bone.

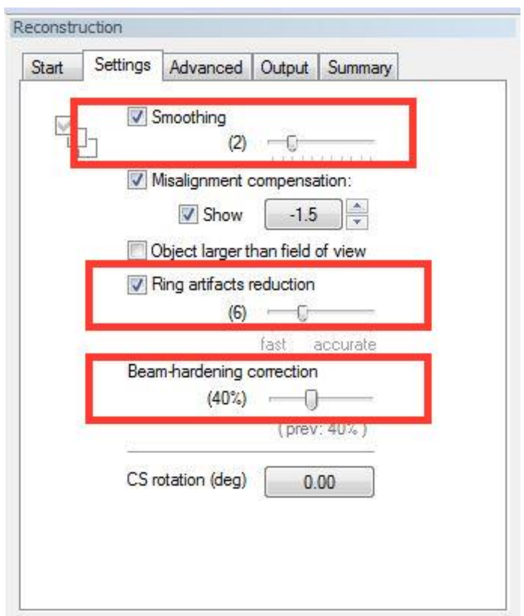

Figure 6: smoothing of 2, ring artifact at 6, and beam hardening at 40%

4.4 Reorientation

The reconstructed dataset was then reoriented using Dataviewer V1.5.2. The mini-screw needs to be reoriented so that the long axis of the mini-screw is along the Z-axis so that the analysis of the peri-implant bone can be standardized between samples. The reconstructed file was opened by loading the entire dataset. The initial unaligned dataset is seen in Figure 7. To align, the "ctl" button was held down and the mouse was dragged in each X,Y, and Z plane. The mini-screw was correctly aligned when it appeared as a solid circle that remained in place when one scrolled up and down in the Z-axis. The X-Y dataset was saved so the mini-screw was orthogonal with the Z axis (Figure 8).

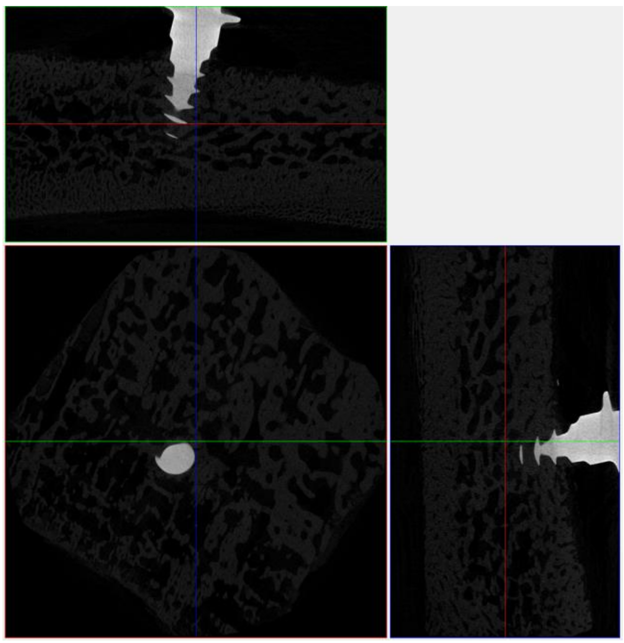

Figure 7: unaligned data set

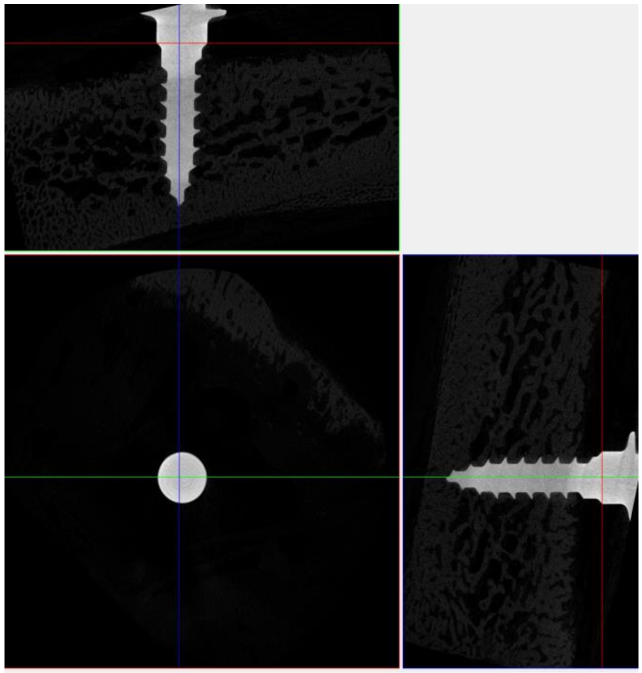

Figure 8: data set aligned orthogonally with the Z axis

4.5 Selecting Volume of Interest (VOI)

The next step was to load the proper volume of interest into the CT-Analyzer V1.14.4.1 software. The dataset was loaded and the apex and base of the mini-screw, just apical to the tissue collar, were hand selected to ensure equivalent areas between all the samples (Figure 9). Next a circular region of interest (ROI) of the same size per sample was selected including the peri-implant bone (Figure 10). This new dataset was then double checked to ensure the mini-screw was orthogonal and entirely encompassed in the ROI.

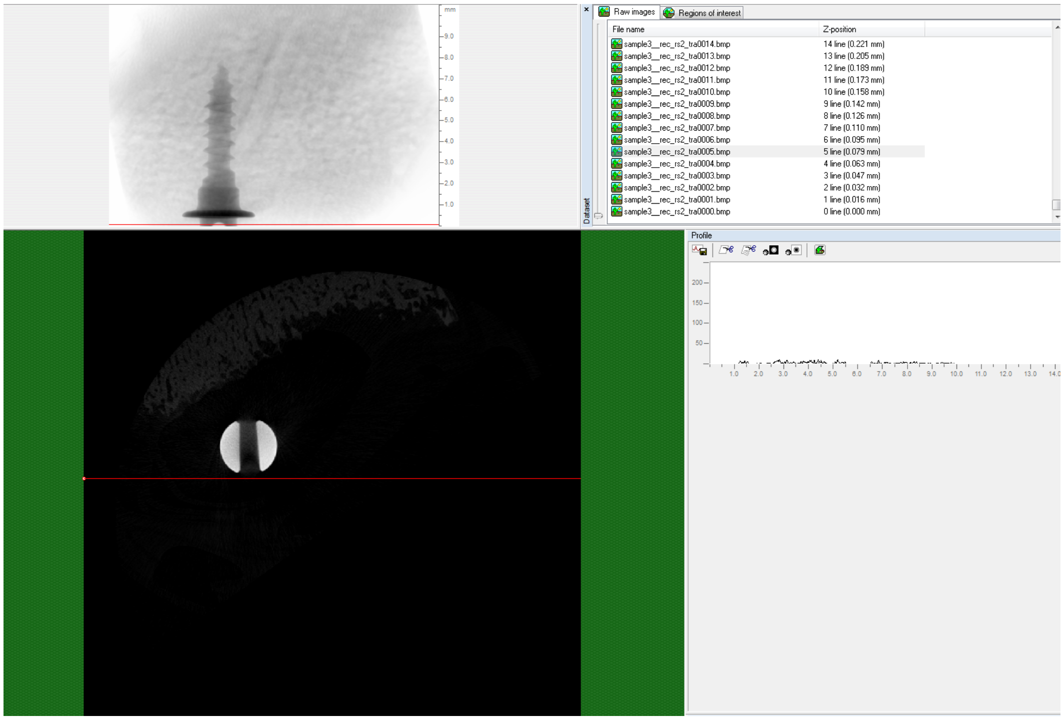

Figure 9: reoriented dataset loaded in CT-Analyzer

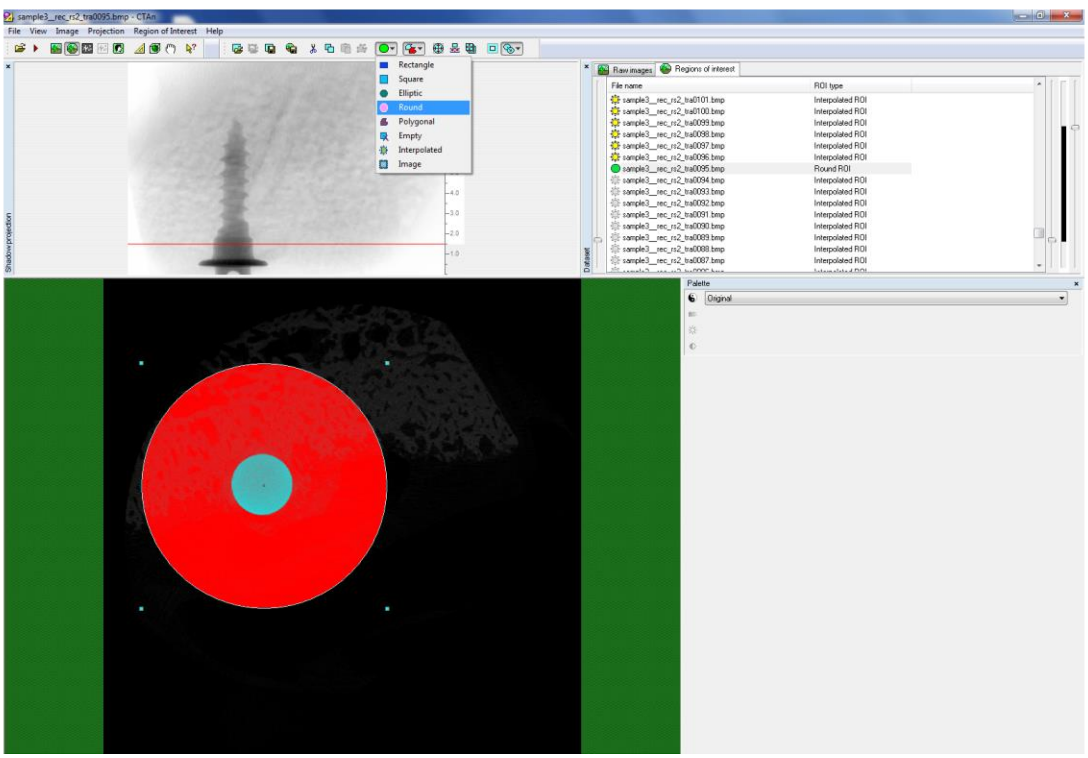

Figure 10: Selecting the ROI

4.6 Analysis of Bone-to-Implant Contact

A custom task list (Figure 11), created by Dr. Lenoid Ephsteyn, was used to analyze the dataset. The standardized ROI files previously created were uploaded to this custom task list. Standard density thresholds were used to distinguish between the bone-metal interphase. The appropriate density thresholds were set as follows based off of accepted norms: 0-28 for empty space/soft tissue, 28-90 for bone, and 90-255 for metal. After running the data through the task-list, the CT-Analyzer software then produced an Excel file with the chosen data including "Percent intersection surface" which represents the bone-to-implant contact. Within this region of interest (ROI), the following data points are of interest:

1) Intersecting Surface (IS)- area of bone within the ROI measured in mm^2
2) Total Surface (TS)- area of the mini-screw surface measured in mm^2
3) Percent Intersection Surface= IS/TS * 100= the percentage of the mini-screw surface that is in contact with bone

This bone-to-implant contact will be compared via statistical analysis to determine if there is a difference in the amount of bone-to-implant contact between the two insertion angulations.

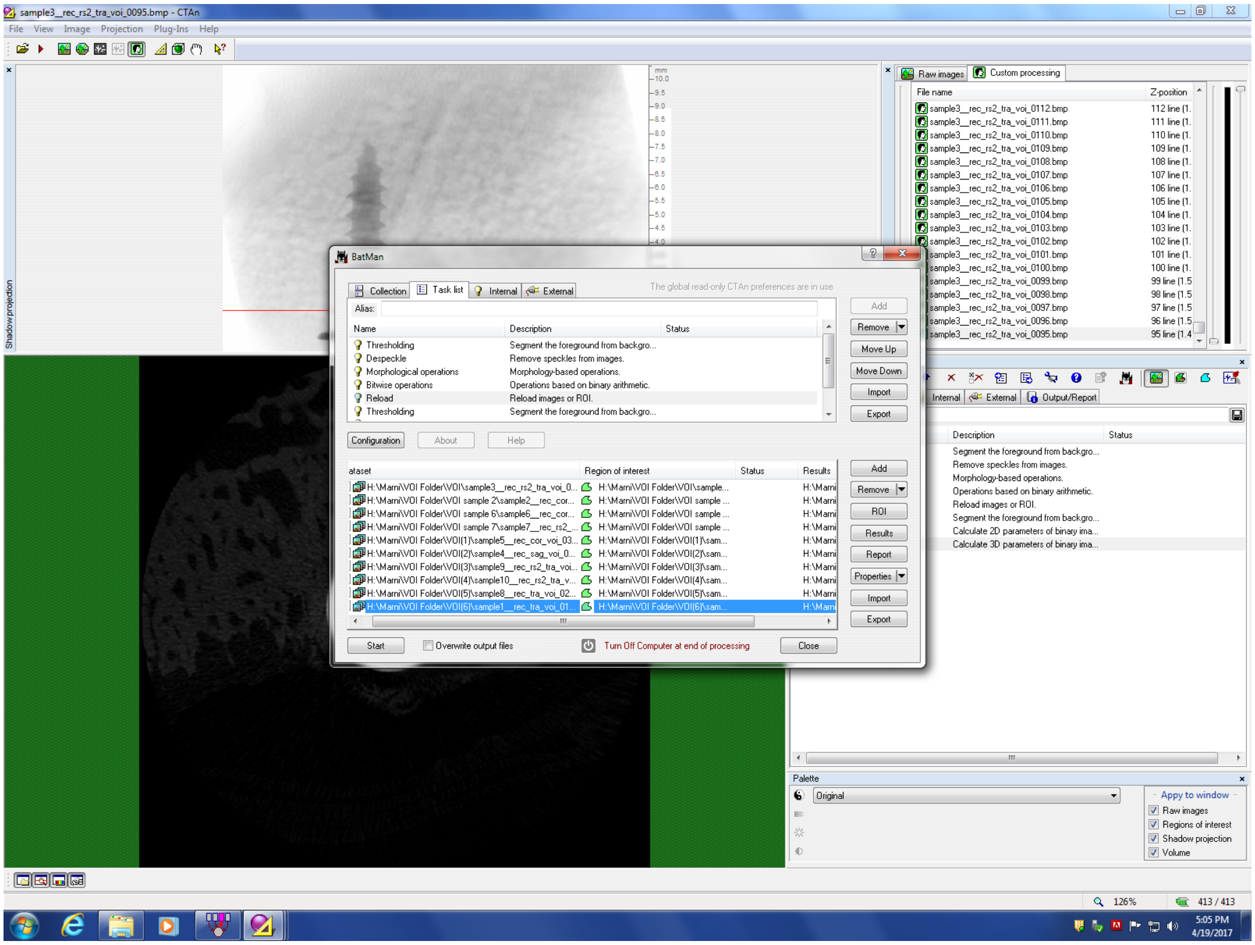

Figure 11: Samples loaded with custom task list for analysis

CHAPTER 5

RESULTS

5.1 Statistical Analysis

All data was entered into an Excel spreadsheet. A curve fit analysis was performed to visualize the data set (Figure 12). A slice from the raw images of sample 9, which was placed at 90^0, and sample 3, which was placed at 50^0 is seen below in Figures 13 and 14.

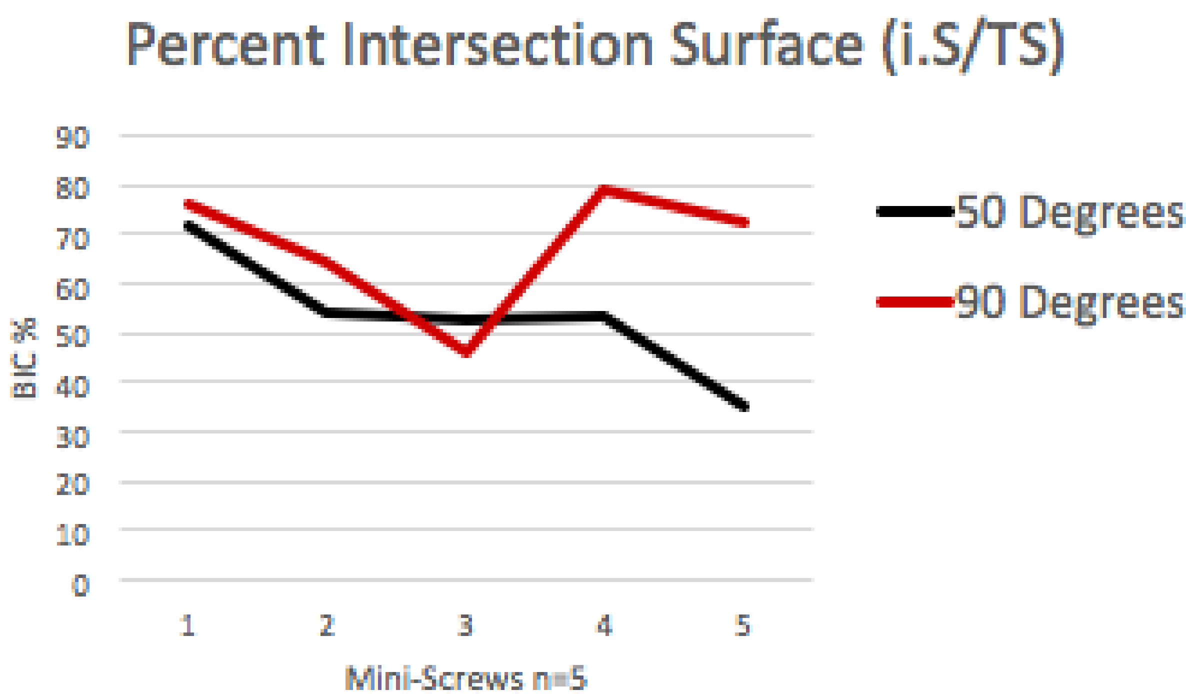

Figure 12: Data set visual analysis

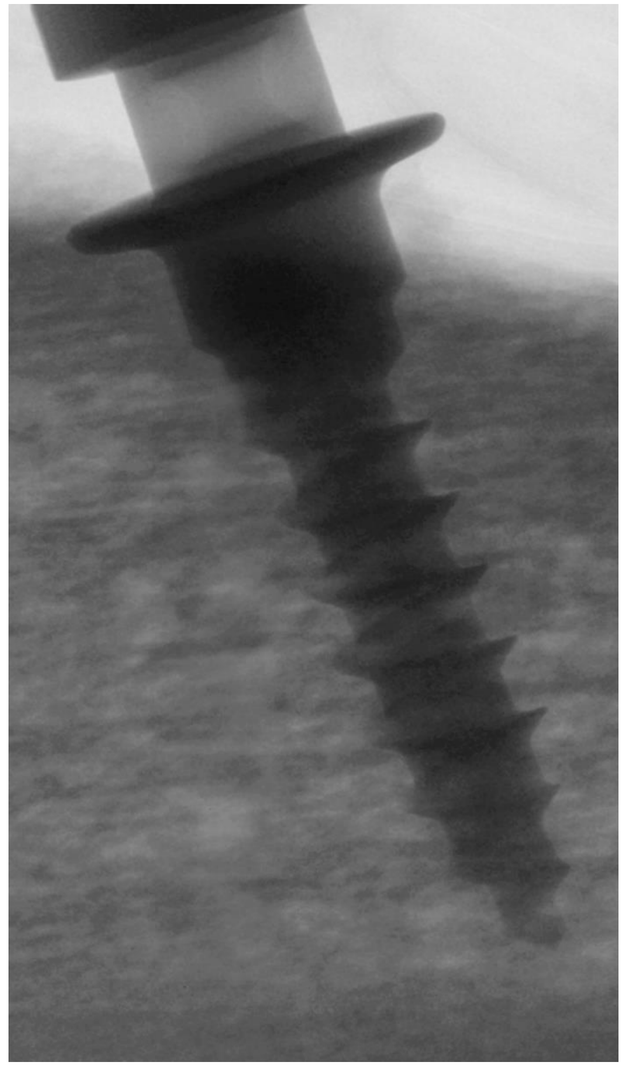

Figure 13:
Sample 9 placed at 90^0

Figure 14:
Sample 3 placed at 50^0

5.2 Quantitative Analysis

The BIC ranged from 35.11% to 78.92%. As seen in Table 1, the mean for the 50^0 insertion group was 53.40% while the mean BIC for the 90^0 insertion group was 67.63%. The ranges between the two groups were similar, with the 50^0 group range at 36.65 and the 90^0 group range at 33.09. The standard deviations were nearly identical, with the 50^0 SD at 11.60 and the 90^0 SD at 11.91.

Insertion Angle	1	2	3	4	5	Mean	SD
50 Degrees	71.76	53.97	52.90	53.25	35.11	53.40	11.60
90 Degrees	76.34	64.72	45.83	78.92	72.34	67.63	11.91

Table 1: Bone to Implant Contact

Statistical significance between the groups was tested via a MannWhitney U test. The test indicated that the BIC was not statistically significantly different for the mini-screws placed at 90^0 (Mdn= 72.34) compared to the mini-screws placed at 50^0 (Mdn= 53.25), $U=5$, $p=.1443$. Therefore, the results do not significantly differ at $p<.05$.

CHAPTER 6

DISCUSSION

6.1 Results in Comparison to Current Literature

With the increasing popularity of mini-screws, understanding which variables increase success is critical. Insertion angulation is one variable that effects the primary stability of mini-screws. Various insertion angulations have been reported as superior in the literature. Petrey et al concluded that 90^0 insertion angulation had higher failure force compared to 45^0 and 135^0. These results differ from our study, given that failure force is highly correlated with primary stability. (Petrey, 2010).

Deguchi et al assessed cortical bone contact with mini-screws placed at 30^0, 45^0, and 90^0 and determined that mini-screws placed at 30^0 have 1.5 times the cortical bone contact as those placed at 90^0. Therefore, they concluded that 30^0 is more ideal, contrary to our results (Deguchi, 2006).

Araghbidkrashani et al took these ideas one step further and included force vectors in the study. They compared the primary stability of mini-screws at insertion angulations 30^0, 45^0, 60^0, and 90^0 along with different vectors of applied force. They determined that the direction of the applied force has a significant impact on the primary stability of mini-screws at various insertion angulations. For example, they found the highest primary stability with a combination of shear force and mini-screws placed at 45^0 along with pullout force with mini-screws placed at 90^0 (Araghibidkashani, 2016). This indicates that the direction of force applied to the mini-screw should be considered when determining if a particular insertion angulation will be beneficial. This study utilized the mini-screws as direct anchorage, not as indirect anchorage.

Xu et al placed mini-screws in living beagle dogs at 30^0, 50^0, 70^0, and 90^0. The bone segments were scanned using Micro-CT and they found statistically significantly higher BIC in the mini-screws placed at 50^0 and 70^0. They concluded that while the depth of the slightly oblique mini-screws was decreased compared with the perpendicular ones, they engaged with a greater percentage of cortical bone, a critical determinant of mini-screw primary stability (Xu, 2013). Our results differ from Xu's in that we did not find any significant difference in the BIC between the two angulations.

6.2 Specific Findings and Clinical Significance

Our research found there to be no significant difference in the BIC between orthodontic mini-screws inserted at 90^0 compared to those inserted at 50^0. This new information allows the orthodontist to place mini-screws with either insertion angulation without fear of increased failure. The orthodontist has the freedom to consider multiple variables, including insertion angulation, to increase the success of their mini-screw. For example, if placing the mini-screw in the palate and using a trans-palatal arch for molar intrusion, according to Araghibidkashani's research on force vectors, it may be advantageous to insert the mini-screw at 90^0 because the force from the intrusion arch is a direct pull out force.

Alternatively, there are multiple scenarios in which an angulated mini-screw can be advantageous, such as 1) to avoid root contact in the buccal shelf, 2) when using shear forces, such as when utilizing mini-screws as skeletal anchorage in maximum retraction cases, 3) to allow for a longer length mini-screw, and 4) to increase contact with cortical bone, and 5) to increase placement ease, for example not being required to hold a straight nose driver perpendicular to the palate. In these situations, it is beneficial for the orthodontist to find a stable insertion position without wobble, even if it is angulated.

6.3 Limitations of the Study and Future Considerations

This study used a total of ten mini-screws, five placed at each insertion angulation. While all the mini-screws were placed in the same pig ramus, the exact bone characteristics in the samples were not taken into consideration. If a ghost sample of pig mandible could have been scanned as a reference, the differences in bone density per sample could have been taken into consideration and could have strengthened the data. However, even though we did not have a ghost sample to scan as a standard for the bone mineral density, we were able to use standardized bone densities from previous animal microCT studies, which removed much of the metal artifact and eliminated a considerable amount of error. A larger sample size and repeat scans would have also strengthened the data.

The stent was created using alginate impression, stone model, a wax set up, and a dental surveyor with a protractor to measure the insertion angulations. It was then cast into acrylic and seated on the bone specimen. While accurate, the accuracy could have been improved by scanning the bone sample and digitally determining the insertion angulations and 3-D printing the stent.

There are many components to improving mini-screw success. Results may have varied if any of the mini-screw characteristics, such as length, diameter, pitch, material, etc were altered.

CHAPTER 7

CONCLUSIONS

1. There is no statistically significant difference in the bone-to-implant contact of mini-screws placed with an insertion angulation of 90^0 compared to those placed with an insertion angulation of 50^0
2. More studies with larger samples sizes and controls for bone mineral density would strengthen this data

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