Title
Developing a Corticopuncture system to accelerate the rate of tooth movement

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Publication Date
2014

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Developing a Corticopuncture system to accelerate the rate of tooth movement

A thesis submitted in partial satisfaction of the requirements for the degree Master of Science in Oral Biology

By: Mohamed Moharam Mostafa

2014
ABSTRACT OF THE THESIS

Developing a Corticopuncture system to accelerate the rate of tooth movement

by

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Master of Science in Oral Biology
University of California, Los Angeles, 2014

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Introduction:

Tooth movement is caused by inflammatory and cellular reactions within the bone in response to applied orthodontic forces. Several attempts have been made to increase the rate of bone turnover in order to achieve accelerated tooth movement. These attempts can be classified into two categories: physical trauma (such as Alveolar Corticotomy "Wilckodontics"[1], Piezopuncture[2], Laser[3] and Resonance Vibrations[4] and application of drugs (such as the systemic and local application of Vitamin D[5], Prostaglandins[6] and Corticosteroids[7]).

Some of these techniques showed inconsistent results such as laser[3, 8, 9]; while others had undesired side effects (such as osteoporosis induced with the use of corticosteroids[7] or headache associated with the use of a pain mediator as prostaglandin[6]) that pose a challenge to their application in clinical practice. Even though the alveolar corticotomy procedure has had more consistent results, it remains an invasive surgical procedure that causes a tremendous amount of trauma. The unanswered question remains: Is there a mechanical system capable of delivering the lowest threshold of trauma that can cause accelerated tooth movement with minimal discomfort and invasion?.

The aim of this study is to design a system that delivers varying levels of trauma to the bone through the gingiva without raising a flap in order to accelerate the rate of tooth movement. It should be minimally invasive with low level forces. In addition, forces and
frequency should be accurately and easily controlled by the operator, and the system should be autoclavable.

**Material and methods:**

Initial trials were conducted using three needles with varying thickness to qualitatively assess the force level needed to pierce a synthetic bone block of medium hardness. Further quantification of these force levels was done using the Instron machine. The needle length can be determined clinically using a periodontal probe and the needle material will be made of both surgical SS 316 & Titanium. A handpiece was designed containing a pressure gauge, a solenoid valve connected to a timer and a pneumatic piston. FEM analysis was performed on the external and needle design of the handpiece to ensure that a uniform thickness of 1mm and the selected needle thickness were sufficient to withstand the forces generated by this system. A prototype was made and it was tested on a pig jaw for the force calibration. Micro Ct scan was used to quantify bone trauma.

**Results:**

The forces that pierced the synthetic bone block were qualitatively assessed by two different examiners. It was found that the needle of 0.23 mm thickness at the apex and 0.4 mm at the base could penetrate the bone with trauma that could be easily tolerated.
by the patient. The Instron machine quantified these forces to range from 10-17 N. The assembled prototype, using the mechanical design described above, produced desired forces and pierced the synthetic bone block, and the operator was able to control the level of force. Further trials on pig jaws proved that the system was able to pierce the gingival tissue with an adequate force to cause bone trauma. FEM study illustrated that a handpiece with 1 mm thick external casing (surgical SS 316 L), and a needle with thickness of 0.23 mm at the apex and 0.4 mm at the base (Surgical SS 316L and Titanium) could safely withstand the forces generated by this system.

**Conclusion:**

The corticopuncture system designed in this study provides a fully controlled, minimally invasive, effective and efficient technique of causing bone trauma. An external handpiece casing could be safely manufactured using surgical stainless steel 316 L with a uniform thickness of 1mm. A needle with a thickness of 0.23 mm at the apex and 0.4 mm at the base could cause adequate bone trauma at force levels that can be tolerated by patients.
Committee Page

REUBEN KIM
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2014
DEDICATION

I would like to dedicate this dissertation to my family who have supported and encouraged me through my life’s endeavors throughout all these years.
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ACKNOWLEDGEMENTS

I would like to express my deepest appreciation to my mentor, Dr Won Moon, without his guidance and persistent help this dissertation would not have been possible.

I would like to thank my committee members; Professor Sotirios Tetradi, Professor Reuben Kim and Professor Ki-Hyuk Shin for their support with this project.

Special thanks to Professor Ben Wu, Dr Christine Hong, Dr Michael Azer, Dr Ehab Bar, Dr Hameed Eyaz and Abigail Corin for their help and support with this project.
Background and rationale

Wearing braces for a long period of time is neither desired nor hygienic; however the average orthodontic treatment usually lasts between 1.5 to 2 years and more time is required for extraction cases[5].

Disadvantages of having braces for a long period of time

There are biological and psychological disadvantages of prolonged orthodontic treatment. Biological disadvantages include food and plaque accumulation around the brackets as the area around the bracket wings acts as a food trap making it difficult to clean. This can lead to a decrease in the pH level, causing decalcification and mineral loss [10]. Wearing braces may also lead to gingival inflammation. This may progress to periodontal disease, which is more common in adult patients [11], or gingival hyperplasia which is more common in adolescent patients due to hormonal levels [12]. Prolonged orthodontic treatment can also increase the risk of further root shortening and resorption [13]. Psychological disadvantages of prolonged orthodontic treatment include esthetics, which is a concern especially for adult patients. As a result, adults usually refrain from orthodontic treatment. In addition, prolonged treatment time may
lead to in-compliance among adolescent patients, making it more difficult to finish cases[14].

For a long time, the biological concepts behind tooth movement were ignored and consequently, research efforts were directed more towards inventing various appliances and bracket systems such as bidimensional and self ligating brackets. These systems had marginal improvements over the rate of orthodontic tooth movement, focusing on reducing friction between the arch wire and the brackets, while the biological responses remained the same. Before discussing previous attempts to accelerate the rate of tooth movement, we need to briefly discuss the theories of orthodontic tooth movement

**Biological concepts of tooth movement**

There are several different theories about the biological response leading to tooth movement [15] .

The bioelectric potential theory [16] suggests that metabolically active bone or connective tissue produce electronegative charges that are generally proportional to their activity. Inactive cells and areas are nearly electrically neutral. Although the purpose of this bioelectric potential is unknown, cellular activity can be modified by adding exogenous electric signals.

The pressure tension theory [17], which is the most widely accepted theory in orthodontics, suggests that chemical rather than electrical signals trigger tooth movement. External orthodontic pressure shifts the tooth position, causing stretching of
the PDL space in one area, and compression on the opposing area. Alterations in blood flow quickly create changes in the chemical environment. The chemical changes stimulate the migration of immune cells, which along with native cells such as fibroblasts and osteoblasts, produce inflammatory cytokines that include: lymphocyte and monocyte derived factors, growth factors, high concentrations of inflammatory cytokines such as interleukin-1 (IL-1), IL-2, IL-3, IL-6, IL-8, tumor necrosis factor-α (TNFα), interferon-γ (IFNγ), and osteoclast differentiation factor. Cellular element such as macrophages or osteoclasts from adjacent tissues invade the inflamed area. Bone deposition occurs along the tension side and resorption occurs along the compression side.

The bone bending theory [18] suggests that when an orthodontic appliance is activated, forces delivered to the tooth are transmitted to all tissues near the area of applied force. These forces bend bone, tooth, and PDL. Bone is more elastic than teeth and bends more readily in response to force application. The active biologic processes that follow bone bending involve accelerated bone turnover and renewal of cellular and inorganic fractions.

The multiple cell signaling pathways theory [19] suggests that cytokines such as IL-1 produced locally by mechanically activated cells are responsible for the resorptive and formative phase of connective tissue remodeling.
Previous attempts to accelerate tooth movement

Several attempts have been made to accelerate the biological response to orthodontic forces based on the aforementioned theories. These attempts can be divided into 2 main categories:

- **Direct physical Trauma as:**
  - Alveolar decortiotion (Wilckodontics)
  - Corticision
  - Application of laser
  - Pulsed electromagnetic field
  - Direct electric current
  - Piezopuncture

- **Local or systemic application of drugs:**
  - Application of vitamin D
  - Injection of prostaglandin
  - Use of corticosteroids

Some of these procedures proved effective, however, their undesired side effects make their direct application in daily clinical practice challenging while others showed inconsistent results. For example, alveolar corticotomy [1] technique is promising but it is an invasive surgical technique. The common risks are morbidity, inflammation, and swelling. In addition, alveolar corticotomy must be performed by a periodontist or an oral surgeon. Thus, it is not feasible to perform alveolar corticotomy with every patient or in an orthodontics office.
Laser application is controversial. Some studies found it to be effective [8], while others proved otherwise [3]. This is because the effect of the laser is governed by several variables, such as the bone density, wave length of the laser beam, and thickness of bone, which can vary even between the mandible and the maxilla of the same patient. The pulsed electromagnetic field [20] technique showed an increase in cellular activity which was reflected by the presence of significantly greater numbers of osteoclasts in the alveolar bone. However, changes in the systemic blood chemistry of the experimental animals related to an increase in protein metabolism and muscle activity or degradation which makes it impossible to be applied in clinical practice. The use of medications also has some systemic effects. For example, prostaglandins are mediators of pain receptors, so its use induced pain and headache [6]. Animal studies have proven that corticosteroids causes osteoporosis [7], so it is not a viable option for human use. Vitamin D [1] has also proven to be highly effective. However, it has several limitations: the optimum dosage, frequency of injection, and effects over a longer period of time have not been determined, and may be variable depending on age, sex, weight and the general health condition of the patient.

Alveolar decortication attempts to accelerate tooth movement

In 1959, Köle attempted to accelerate the rate of tooth movement via reducing the resistance exerted by the surrounding cortical bone. He introduced a surgical technique that involved both osteotomy and corticotomy which proved to be efficient[21].
In 1981, it was found that the acceleration in orthodontic tooth movement was due to a temporary stage of localized hard-tissue remodeling that resulted in rebuilding of the injured sites to a normal state through recruitment of osteoclasts and osteoblasts via local intercellular mediator mechanisms involving precursor supporting cells, blood capillaries and lymph. This was phenomenon was named the “Regional Acceleratory Phenomenon” (RAP)[22].

“Wilckodontics,” which refers to surgical injury to alveolar bone, has been proven to temporarily accelerate tooth movement [21]. This occurs because surgical trauma triggers the release of inflammatory mediators [23], causing vasodilation of the blood vessels and increasing the recruitment of osteoclasts. This accelerates bone turnover and thus increases the rate of tooth movement. In addition to increasing the remodeling rate of alveolar bone, corticotomy decreases bone mineral density [24] thus decreasing the mechanical resistance of dentoalveolar tissues to orthodontic forces and decreasing the risk of orthodontic related external root resorption. Corticotomy has proven to be a highly effective procedure, and it is the most commonly used technique. However, it is an invasive surgical procedure (Figure 1) that must be performed by a periodontist or an oral surgeon.

To overcome this problem, attempts have been made to make the technique less invasive. For example, in corticision, a surgical mallet taps upon a reinforced scalpel that is used as a thin chisel [25]. This was shown to be an effective technique to induce
cortical bone trauma without reflecting a flap. However, it remains an invasive procedure (Figure 2) that cannot be performed by an orthodontists.

In another attempt, a bur and a high speed handpiece were used to cause trauma through the gingiva [26]. This attempt was a much less complicated clinical procedure than the corticotomy discussed above. However, it is still an invasive procedure (Figure 3) in which the length of drilling was not controlled and excessive gingival trauma was induced due to the rotary nature of the drilling machine.

A new technique called piezopuncture is currently being tested [2], in which an endodontic piezotome is used to apply trauma to the bone through the gingiva. Animal study proved that piezopuncture is an effective technique that can accelerate the rate of tooth movement 2-3 fold, However, this remains an invasive procedure that may lead to bone necrosis due to the generated heat.

The lowest threshold of bone trauma required to accelerate the tooth movement is unknown at this time, and development of a novel instrument that can deliver varying levels of force with minimal invasiveness is necessary in order to establish this threshold.
Objectives and specific aims

The objective of this study is to design a force delivery system that is able to causes minor trauma to the cortical bone penetrating through the gingiva with minimal invasiveness.

Specific aims:

1- Assess qualitatively and quantitatively the forces needed to apply trauma to cortical bone using various needle sizes and shapes.
2- Design an internal mechanical system that will control the needle movements and deliver the required force.
3- To explore the clinical applicability of the novel handpiece design.
4- Design an ergonomic external casing that can withstand the applied forces.

Hypothesis:

The novel corticopuncture instrument can produce tolerable oscillating forces that puncture the cortical bone through the gingiva.
Materials and methods

Materials:

Initial testing was performed on biosynthetic bone blocks (Figure 4)(Sawbones®, Pacific Research Laboratories Inc, Vashon, WA). The cortical bone simulation is 2mm thick [27, 28] and is made of solid rigid polyurethane foam (SRPF). The remaining and middle material of the sample was used to simulate cancellous bone and is made of cellular rigid polyurethane foam (CRPF). The density of SRPF was 30 pounds per cubic foot (pcf). The glue that was used to laminate the two materials together is of negligible thickness and made of 40 pcf solid rigid polyurethane foam[29].

Trauma was conducted using stainless steel nails on the synthetic bone block (Figure 5). Three different nail sizes were used and the thickness at the nail apex and at 3 mm of length was measured using a digital caliber (Table 1). Needles were covered with acrylic to control the penetration depth.

Solenoid valve “744100115B” manufactured by Parker company (Figures 6,7), pneumatic division was included in the prototype. This solenoid valve is designed to control the flow of inert liquids and gases. It is a 3-way, normally closed valve with a pipe exhaust. The port size is 1/4” SS. The electric current supply was controlled by the solenoid timer.
The solenoid timer “CLRB-115A-2-30S-10S” manufactured by R.K. Electronics (Figure 8) was connected to the solenoid valve. This timer works under an input voltage of 115A and has 2 knobs on top to adjust the on / off time. The off time ranges from 0.3 - 30 sec and the on time ranges from 0.1 to 10 sec.

The pneumatic cylinder is manufactured by the Parker Hannifin Ltd company (Figure 9,10). These cylinders are threaded on the outside, allowing new needles to be inserted for each patient. This allows for a compact and efficient design as there is no additional space required for the attachment mechanism between the needle and the handpiece. Different pneumatic pistons are supplied by the manufacturer, however piston “P1G-S016SS-0005” was used for this experiment. This specific piston is a single acting piston with an internal spring included which is needed for the retracting stroke. According to the manufacturer, this piston works under maximum air pressure of 7 bar (101 PSI) and minimal air pressure of 2 bar (29 PSI). Air pressure will be controlled at the level of the compressed air using the air pressure valve. The stroke length is 5mm.

The husky mini pressure regulator was used (Figure 11,12). It is pressure regulator that can be used between the air supply and an air tool to control the air pressure on lines carrying up to a maximum of 300 PSI. The regulator output pressure range is 0-125 PSI.
Methods

Initial experimentation:

Needles, covered with acrylic, were attached to a hammer and trauma was applied to the synthetic bone block (Figure 4). Force levels were qualitatively assessed to establish the range of tolerable forces that can penetrate cortical bone with a level of trauma that can be easily tolerated by the patient. This was done by placing the synthetic bone block on the operator’s cheek and applying trauma to the synthetic bone block using different needle sizes. To insure inter-rater and intra-rater reliability, forces were assessed by two different examiners at two separate time points.

Further quantification was done using the Instron machine. The Instron machine is designed to undergo a variety of tests such as displacement, strain and stress (Figure 13). For the sake of this experiment, the loading cell was inverted to allow for sample placement and manual application of forces on top of it (Figure 14). Forces previously assessed were applied manually by the operator and measurements were obtained.

Mechanical Design:

A system designed with a pressure regulator, solenoid valve attached to a timer and a pneumatic piston was suggested to generate adequate forces that can duplicate
those measured manually while allowing full control by the operator. 1/4” nylon tubing was used to assemble the prototype parts together allowing for continuous compressed air flow through the mechanical system. Tubing was inserted into 1/4” straight male connector attached to both ends of the solenoid valve, pressure regulator and the inlet end of the pneumatic piston (Figure 15). The system was connected to the dental unit using a chair side adaptor (Figure 16). The small needle was attached to the pneumatic piston threads using composite resin. The air pressure regulator was set to allow compressed air of 75 PSI into the system and initial testing was done using synthetic bone block. The bone block was visually inspected for the amount of trauma induced; and qualitative assessment of the forces used was done by placing the bone block on the operator’s cheek. This was done to ensure that the mechanical system can duplicate the tolerable forces produced by the manual trauma.

**Micro CT Scan**

Further Prototype testing was done using procaine bone. Pig jaw was obtained and trauma was delivered at 4 different air pressures: 35 PSI, 55 PSI, 75 PSI, and 90 PSI through the gingival tissue into the bone. A flap was raised and the bone trauma was visually inspected. After blinding the operator to remove any bias, the trauma area of the pig jaw was dissected into two smaller samples and measurements were taken. Sky scan 1172 high resolution Micro CT scan manufactured by Bruker was used to quantify the pitting of the bone; via measuring the depth and volume. Analysis of variance (ANOVA) was performed to analyze the difference between groups.
**Finite element analysis:**

The external design of the handpiece was created using 3D CAD modeling software (MAYA). Solenoid valve, timer and pressure regulator were placed in a small chair side box with buttons to control the frequency of trauma and air pressure allowed into the system (Figure 17). The pneumatic piston was placed in an ergonomically designed handpiece to deliver trauma intraorally. STL file of the handpiece external design was obtained and imported into Ansys Vr. 12 software to form the mesh. The hand-piece was simulated to have a uniform thickness of 1 mm and physical properties of surgical stainless steel 316L “Young’s Modulus (195 MPa) and Poisson’s Ratio (0.3)” were used; while the needle with an apex of 0.23 mm and a base of 0.4 mm was simulated using both surgical stainless steel 316L and titanium “Young’s Modulus (485 MPa) and Poisson’s Ratio (0.35)”.

Linear static Analysis was performed

Total Number of nodes: 20248

Total Number of Elements: 19633

Used Element: Shell 63

Boundary condition: Fixed top and side areas (areas supported by the dentist, (Figures 18,19))

Load: A load of 35 N was used to simulate the forces applied to the handpiece.
Results:

**Initial experimentation**

Forces exerted by small and medium sized needles were assessed by the examiners to penetrate the synthetic bone block with an impact that can be tolerated by the patient; however, forces produced by the smaller needle (0.23mm at the apex and 0.4 mm at the base) were found to be easier to handle than those of the medium size needle. Forces produced by the large needle were found to be excessive and would be highly uncomfortable to the patients.

Further quantification of the applied forces was done using the instron machine (Graphs 1,2,3,4). It was found that the medium sized needle needs forces of a minimum value of 25N and a max value of 42 N with an average of 34 N; while the small needle can penetrate the synthetic bone with forces ranging from 10-17 N with an average of 15 N.

**Mechanical Design:**

The handpiece prototype was assembled and initial experimentation proved that the forces produced manually can be duplicated using the suggested mechanical design. Forces as well as the frequency of trauma application can be accurately controlled by the operator. This can be done by controlling the amount of compressed
air pressure allowed into the mechanical system using the pressure regulator, and the
frequency of the applied trauma by controlling the flow of electric current into the
solenoid valve using the solenoid timer. Further experimentation using the pig jaw
proved that the needle can penetrate through the gingival tissue with these forces. A
flap was raised and bone trauma appeared evident as pitting in the buccal bone that
could be visually verified by the operator (Figures 20, 21, 22 and 23). Quantification of
the trauma was done using the Micro CT Scan.

**Micro CT Scan**

Thickness of cortical bone, depth and volume of bone trauma were calculated
(Table 2, 3 and 4). It was found that the depth of bone deformation was highest in the 90
PSI (group 4) with a mean of 1.17 +/- 0.09 mm, followed by 75 PSI (group 1) with a
mean of 1.140 +/- 0.06 mm then 55 PSI (group 3) with a mean of 0.787 +/- 0.25 mm
and finally 35 PSI (group 2) which had a mean of 0.32 +/- 0.717 mm. The four groups
showed significantly different mean depths, with the highest difference noticed among
group 1 and 4 (1.1400 and 1.1771 respectively) followed by the third group while the
lowest was among the second group.

On the other hand, It was found that trauma volume in the 90 PSI (group 4) had a
mean of 0.328 +/- 0.34; 75 PSI (group 1) had a mean of 0.090 +/- 0.0055 mm3; 55 PSI
(group 3) had a mean of 0.139 +/- 0.205 mm3 and 35 PSI (group 2) 0.104 +/- 0.198
mm3. The four groups did not show statistically significant differences.
(Figure 24) Trauma was applied to the buccal bone in 3 different areas (apical, middle and coronal) using the same compressed air pressure (90 PSI). It was noted that in the coronal portion, buccal bone thickness was significantly greater than that in the middle and apical parts. Depth of trauma measured 0.844 mm for the coronal portion while in the middle and apical portions, trauma went beyond the thickness of buccal bone to measure 1.6 and 1.37 mm respectively. Trauma to the tooth follicle was obvious in both the middle and apical portions. This demonstrates the effect of the buccal bone thickness on the depth of trauma, indicating that less air pressure should be used in areas of thin bone thickness.

**FEM**

Analysis of stress and deformations occurring under forces of 35 N applied to the handpiece’s outer casing made of surgical stainless steel 316 L and the needle made of surgical stainless steel 316 L (Figures 31-36) and titanium (Figures 37-42) were:

S1: Maximum tensile stress

S3: Maximum Compressive stress

Usum: Total deformation.

Maximum stresses appeared at the junction between the needle and handpiece while maximum deformations was found to be at the needle tip. All the deformations are measured in Mpa and were shown to be lower than the endurance limit of both the surgical stainless steel and titanium.
Needle Design Experimentation:

Needle design was divided into 3 main parts: length, width and material. Needle width was determined based on the initial clinical experimentation; which proved that a force of 15 N can penetrate simulated cortical bone block with forces easily handled by the patient.

To determine the ideal needle material; a comparison between surgical stainless steel (316L) and titanium (Ti) alloys was made, as those are the two most commonly used metals in orthopedic appliances.

Physical Properties of Surgical Stainless Steel vs. Titanium

Corrosion Resistance

While stainless steel used for implants has about 18wt% Cr and 8wt% Ni, type 316 stainless steel contains molybendum (Mo) which increases its corrosion resistance. In addition, type 316L has greater corrosion resistance as its carbon (C) content is lowered from 0.08 to 0.3wt%. High corrosion resistance can be attributed to the formation of a protective film of chromium oxide passive layer formed upon oxidation.
On the other hand, it is well documented that titanium has a high corrosion resistance. This is due to the formation of an inert oxide film that helps the metals resist corrosion. Even in the case of trauma and scratching of the outer layer, the film is self-healing and re-forms almost immediately when the needle is exposed to oxygen. Both titanium and stainless steel have excellent corrosion resistance.

**Mechanical Properties**

Analysis of mechanical properties includes assessing the strength, toughness, elastic modulus, and ductility of the metals [30]. The differences between the mechanical properties of type 316L stainless steel and Ti and its alloys are summarized in (Table 5).

Yield strength (Y.S.) is defined as the stress at which a material begins to deform plastically. Surgical Stainless Steel “316L” has a yield strength of 190 MPa while that of titanium and its alloys is 485-795 MPa. This means that titanium has less tendency to deform under stresses than surgical stainless steel.

Ultimate tensile strength “U.T.S.” is the maximum strength that the metal can withstand while being stretched before failure. Surgical Stainless Steel “316L” has an U.T.S. of 490 MPa while that of Titanium and its alloys is 550-860 MPa. This means that titanium is more brittle than surgical stainless steel.
Young's modulus: is used to measure the strain of an object to an applied stress as long as the stress is less than the yield strength of the material. Surgical Stainless Steel 316L has a Y.M. of 190 MPa; while that of Titanium and its alloys is 485-795 MPa. Meaning that a needle made of titanium will be stiffer than one made of stainless steel.

Maximum elongation is the measure of the ductility of a material as determined by a tension test. Surgical stainless steel “316L” has a maximum elongation of 40% while that of titanium and its alloys is 10-15%. This means that a needle made of titanium will be more brittle compared to one made out of surgical stainless steel 316 L.

To determine the indicated needle length, a literature review of the average soft tissue and cortical bone thickens in different quadrants was conducted; however, a lot of variation was found in the cortical bone thickness and not many articles about the thickness of soft tissue could be found (Table 6)[30, 31]. It was also documented that cortical bone thickness can be easily affected by age, race and type of malocclusion (Tables 7-10)[27, 31-33]. One possible way to overcome this is to use a “direct measurement device” such as an ultrasonic gingival-thickness meter. Yet, this will make the whole procedure more complicated and requires more equipment . For the sake of this procedure; a periodontal probe will be used to assess the thickness of gingival tissue. This will make the whole process less complicated, as gingival tissue will be anesthetized before trauma, so measurements will be made at no discomfort to the patient. The length of the needle will vary, ranging from 2-6mm to accommodate for different gingival and bone thicknesses.
**Mechanical Design**

The mechanical design consists of a pressure regulator valve placed at the source of compressed air, a solenoid valve attached to a timer and a pneumatic piston. The pressure regulator valve will control the air pressure in the system and thus allow the operator to control the forces produced by the pneumatic piston depending on the age, cortical bone thickness, location of the applied trauma and the patient’s comfort level. The solenoid timer will control the relay of electricity into the solenoid valve. When an electric current is applied, there is a delay period (off time) where the timer will not allow electric current to pass through the solenoid valve, thus blocking the flow of compressed air from port A to P (Figure 6). At the end of this off timer period, the electric current is allowed to pass and the second delay period (on time) begins where the air passes from port A to P and the stroke is produced. The duration of the off time and the on time will be controlled by separate controllers on solenoid timer and be fully adjustable by the operator. Pneumatic cylinders are mechanical devices that use compressed gas or liquids to produce a force in a reciprocating linear motion. Force is exerted when compressed air or liquids are allowed to enter into one end of the pneumatic piston. This exerts pressure, displacing the piston in a forward direction. Once pressure exerted by the compressed air is stopped, the compressed spring pushes the piston in a backwards direction allowing the system to return to its original state.
Forces exerted by the pneumatic pistons are governed by this equation:

\[ \text{Force} = \text{Pressure} \times \text{Area} \]

Forces exerted depend mainly on the pressure applied by the compressed air. By adjusting the applied pressure using a pressure regulating valve, it is possible to increase or decrease the final forces exerted by the pneumatic cylinder.

Another method of force regulation is to choose a cylinder with either a larger or smaller diameter, i.e. by manipulating the surface area on which the pressure is exerted. The larger the diameter, the greater the surface area of the piston and thus the greater the force exerted at the cylinder rod and piston. Based on the equation explained above; if the compressed air the dental unit exerts has a maximum value of 116 PSI, then a pressure of 43.5 PSI can be used as the average. The “excess air pressure will be controlled by the pressure regulator valve” and the required forces will range from 10-17 N which, is equivalent to 2.25 / 3.37 lbs (1N = 0.224 lbs) the area of pneumatic piston = \( \frac{3.37}{43.5} = 0.077 \text{ in}^2 \), 49.677\( \text{mm}^2 \).

Therefore, the area needed to determine the diameter can be found using the equation

\[ \text{Area} = \pi \times R^2 \]

\[ R = \sqrt{\frac{\text{Area}}{\pi}} = \sqrt{0.077 / 3.142} = 0.157" \]

\[ \text{Diameter} = 2 \times \text{radius} = 2 \times 0.157 = 0.313" \text{, } 7.95 \text{ mm} \]
One essential part of the piston that should not be left out of the equation is the spring. Forces exerted by the spring can be calculated using Hooke’s law. Hooke’s law states that the force needed to extend or compress a spring by some distance is proportional to that distance multiplied by a constant. This constant factor is characteristic of the spring depending on its stiffness.

\[ F = Kx \]

where \( K \) is the constant and \( x \) is the distance

\( x \) is the distance travelled by the pneumatic piston which is 5 mm and \( K \) is a constant that can be calculated using the equation:

\[ K = \frac{Gd^4}{8nD^3} \]

Where \( G \) is young’s modulus of elasticity, \( d \) is the wire diameter, \( n \) is the number of active coils and \( D \) is the mean coil diameter.

The spring will be made of surgical stainless steel “316L” having a young’s modulus of 200 GPa [2], 1mm wire diameter, 3 active coils and a mean diameter of 7mm based on the equation above.

\[ K = \frac{200 \times (1)^4}{8 \times (3)^3} = 2.42954 \text{ newton/meter, 1.387 pound/inch} \]

So the forces exerted by the spring= \( 2.42954 \times 5 = 12.1477 \text{ N} \)

So forces needed = \( 12.14 + 15 = 27.14 \text{ N} \); however, a safety margin should be included to overcome the anticipated friction and to allow force manipulation using the pressure regulator gauge, so a 25% increase in the calculated forces will be included which is equal to 6.7 N making the final amount of forces 33.85 N
So if the compressed gas of the dental unit exerts, on average, a pressure of 43.5 PSI and the required force is approximately 33.85 N which is equivalent to 7.6 lbs the Area of pneumatic piston = \( \frac{7.6}{43.5} = 0.174 \) in\(^2\)

So the area needed to determine the diameter can be found using the equation 
\[
\text{Area} = \pi \times R^2 \\
R = \sqrt{\frac{\text{Area}}{\pi}} = \sqrt{\frac{0.174}{3.142}} = 0.2353"
\]

Diameter = \( 2 \times \text{radius} = 2 \times 0.2256 = 0.4706'' \), 11.953 mm

The needed Pneumatic piston diameter should be approximately 12 mm

**Micro CT Scan**

After assembling the prototype; initial experiments were conducted using a pig’s jaw. Porcine bone was selected because of its close resemblance to human bone in mineral concentration and density[34]. As for thickness, the average buccal bone thicknesses in different areas of the oral cavity were compared to measurements obtained from the microCT scan of the pig jaw. Thickness of cortical bone in the mandibular posterior area ranged from a minimum mean of 3.3 +/- 0.84 in the canine area to a maximum of 6.58 +/- 1.27 mm in the first molar area[35] while that of the anterior maxillary facial bone had a mean of 0.5mm 4mm below the CEJ and 0.6mm at the middle of the root[36].

Our sample had a maximum bone thickness of 2 mm and a minimum bone thickness of 0.8 mm with a mean of 1.178 mm and standard deviation +/- 0.35. The extent of alveolar bone decortication varies as there is no superior technique dictating a specific
pattern, depth or extent of the selective decortication[37], however, it was reported to be around 0.5 mm in depth [38] depending on the thickness of cortical bone in the selected area which varies greatly within the same jaw [39] and between hypodivergent and hyperdivergent patients [31]. Therefore, either a CBCT is recommended before alveolar decortication to assess the cortical bone thickness in the area as it was reported to be a reliable method to assess the cortical bone thickness[40] or trauma should be limited to the inter-radicular areas only to avoid injuring the root.

Trauma produced at 90 PSI and 75 PSI had a mean depth of 1.17 mm and 1.14 mm respectively indicating that forces were too strong and trauma applied was too deep. On the other hand; trauma at 55 PSI and 35 PSI had a mean of 0.78 mm and 0.32 mm respectively indicating that adequate, minimally invasive trauma was produced. Volume differences were not significant between the 90 PSI and other 3 groups despite the significant difference in depth, which means that the level of trauma may be equal to the lighter forces that can be achieved with minimal discomfort to the patient.

**Finite Element Analysis:**

Finite element analysis (FEM) is a computer-aided mathematical technique that predicts the response of physical systems subjected to external influences by approximating numerical solutions for complex problems in engineering, science, and applied
mathematics. FEM uses a complex system of nodes to form a mesh. These nodes are assigned material properties and forces are applied to simulate clinical loading. The expected response of the material to these stresses can be calculated. Ultimately, FEM helps to visualize where structures bend and twist, and reveals the distribution of displacements and stresses[41].

Handpiece outer casing manufactured of surgical stainless steel 316 L of 1mm thickness or more proved to be reliable as stresses produced are much lower than the endurance limit. Needle design from both surgical stainless steel 316 L and titanium are safe enough to be used as the needle material.

**Future Directions:**

Tool will be utilized to apply trauma using an animal model, replicating our previous study[2] and thus being able to compare the results. If the rate of tooth movement accelerates in a comparable rate, then further studies can be done using a human model. On the other hand; if minimal or no acceleration of the rate of tooth movement is achieved, modifications of the Corticopuncture system, as by using a smaller needle to apply deeper bone trauma utilizing the same amount of force or increasing the amount of applied force should be done.
Conclusion:

1-Needle of 0.23 mm thickness at the apex and 0.4 mm at the base can penetrate biosynthetic bone blocks with forces that can be tolerated by the patient.

2-The novel instrument developed in this study can deliver the above force range effectively.

3-Experimentation using pig jaw proved that prototype produces enough trauma to pierce the gingival tissue and cause trauma to the cortical bone. This will aid in establishing the “corticopuncture” technique in orthodontics.

4-Micro-CT scan data provided valuable quantitative analysis indicating that the volume of bone deformation were similar in all force levels, suggesting that the lighter force may be able to deliver adequate bone trauma.

5- Finite element analysis proved that outer casing manufactured of surgical stainless steel 316 L of 1mm thickness or more and needle design from both surgical stainless steel 316 L and titanium proved to be reliable, and the stresses produced are much lower than the endurance limit.
Limitations and Potential Pitfalls of the study

- Malocclusion may affect the cortical bone thickness as in class II division II cases where the root is torqued labially causing narrowing of the labial cortical bone. A CBCT is recommended before trauma application for the accurate assessment of the buccal cortical bone thickness.

- Further investigations should be done to assess the effect of gingival trauma on patients with a thin gingival biotype.

- Forces used in the FEM study were static however the real-life process is dynamic.

- Control of the handpiece recoil will differ with different operators.

- Further animal and human studies are needed to confirm the accurate calibration of the handpiece.
Figures 1: Demonstrating Wilckodontics surgery where a full thickness flap is reflected and selective alveolar decortication is performed “A”. Graft material is placed to increase buccal alveolar bone thickness “B” and flap is repositioned and sutured “C” [21]
Figures 2: Demonstrating corticision surgery where a surgical scalpel is used as a chisel and a mallet is used to tap on it. Trauma is introduced to the cortical bone through the gingiva without raising a flap [25]
Figures 3: Trauma caused by flapless decortication of the alveolar bone using round bur. Excessive gingival tissue trauma due to the rotary nature of the handpiece can be noted[26]
Figure 4: Needle, covered with acrylic, attached to a hammer and synthetic bone block
<table>
<thead>
<tr>
<th>Size</th>
<th>Tip</th>
<th>Base</th>
</tr>
</thead>
<tbody>
<tr>
<td>Small</td>
<td>0.23</td>
<td>0.4</td>
</tr>
<tr>
<td>Medium</td>
<td>0.36</td>
<td>0.75</td>
</tr>
<tr>
<td>Large</td>
<td>0.76</td>
<td>1.2</td>
</tr>
</tbody>
</table>

Figure 5: Diameter of the needles used to apply trauma to the synthetic bone block at the tip and the base “3mm from the tip”

Figure 6: Diagram showing 3 way, normally closed solenoid valve with a pipe exhaust.
Figure 7: Solenoid valve “744100115B” manufactured by Parker Company

Figure 8: Solenoid timer by R.K. Electronics to control the electric relay to the solenoid valve
Figure 9: Diagram showing pneumatic cylinder

Figure 10: Diagram showing different components of Pneumatic cylinder manufactured by “Parker Hannifin Ltd company”
Figure 11: Diagram showing different components of pressure regulator

Figure 12: Husky mini pressure regulator
Figures 13: Instron machines used to quantitatively assess manually applied forces

Figures 14: Instron machines with inverted loading cell to apply manual trauma
Figures 15: Assembled prototype

Figures 16: Connector adapter to connect the prototype to the dental unit
Figures 17: External handpiece design where, solenoid valve, timer and pressure regulator will be placed in a small chair side box with buttons to control the frequency of trauma and air pressure allowed into the system. Pneumatic piston will be placed in the handpiece to deliver trauma intraorally.

Figures 18,19: Boundary conditions used to duplicate the clinical situations in the FEM analysis.
Figures 20, 21, 22, and 23: Pig jaw was obtained and trauma was applied using the prototype. Trauma was visually inspected to assess the effect on the gingival tissue and a flap was raised to inspect the extent of bone trauma.
Figure 24: Demonstrates the effect of buccal bone thickness on the applied trauma. Trauma was applied with 90PSI pressure.

(Figure 25), trauma was conducted at a force level of 75 PSI, it had a stroke depth of 1.2 mm in a buccal bone thickness (2 mm).

(Figure 26), trauma was conducted at a force level of 75 PSI, stroke depth of 1.26 mm in a buccal bone thickness (0.9 mm).
(Figure 27), trauma was conducted at a force level of 55 PSI, it had a stroke depth of 0.6 mm in a buccal bone thickness (1.2 mm)
(Figure 28), trauma was conducted at a force level of 55 PSI, it had a stroke depth of 0.5 mm in a buccal bone thickness (1.3 mm).

(Figure 29), trauma was conducted at a force level of 35 PSI, it had a stroke depth of 0.6 mm in a buccal bone thickness (1.2 mm)
(Figure 30), trauma was conducted at a force level of 35 PSI, it had a stroke depth of 0.5 mm, 0.4 mm in a buccal bone thickness (0.7 mm, 1.4 mm) respectively.
Figure 31-36: FEM analysis on StSt needle, S1: Maximum tensile stress, S3: Maximum Compressive stress, Usum: Total deformation
Figures 37-42: FEM analysis on Ti needle, S1: Maximum tensile stress, S3: Maximum Compressive stress, Usum: Total deformation
Table 1: Diameter of the needles used to apply trauma to the synthetic bone block at the tip and the base “3mm from the tip”

<table>
<thead>
<tr>
<th></th>
<th>Tip</th>
<th>Base</th>
</tr>
</thead>
<tbody>
<tr>
<td>Small</td>
<td>0.23</td>
<td>0.4</td>
</tr>
<tr>
<td>Medium</td>
<td>0.36</td>
<td>0.75</td>
</tr>
<tr>
<td>Large</td>
<td>0.76</td>
<td>1.2</td>
</tr>
</tbody>
</table>
Table 2: Micro CT scan measurements on the procaine bone. 1,2 were under 75 PSI air pressure, 3-15 was trauma under 35 PSI, 961 - 630 was under trauma of 90 PSI and 1208 - 1021 under trauma of 55 PSI. Slide start & end are the slide number in the scanned sample.
Table 3: Comparison of trauma depth among the four experimental groups; where group 1 is trauma at 75 PSI, group 2 is trauma at 35 PSI, group 3 is trauma at 55 PSI and group 4 is trauma at 90 PSI

<table>
<thead>
<tr>
<th>Groups</th>
<th>N</th>
<th>Minimum</th>
<th>Maximum</th>
<th>Mean ± SD</th>
<th>95% Confidence Interval for Mean</th>
<th>F**</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2</td>
<td>0.0870</td>
<td>0.0947</td>
<td>0.0909 ± 0.0055</td>
<td>0.0418 - 0.1400</td>
<td>1.082</td>
<td>0.378</td>
</tr>
<tr>
<td>2</td>
<td>13</td>
<td>0.0260</td>
<td>0.7590</td>
<td>0.1044 ± 0.1981</td>
<td>-0.011 - 0.2240</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>7</td>
<td>0.0243</td>
<td>0.5525</td>
<td>0.1390 ± 0.2050</td>
<td>-0.0761 - 0.3541</td>
<td></td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>6</td>
<td>0.1142</td>
<td>0.8450</td>
<td>0.3281 ± 0.3487</td>
<td>-0.2267 - 0.8830</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Total</td>
<td>28</td>
<td>0.0243</td>
<td>0.8450</td>
<td>0.1474 ± 0.2234</td>
<td>0.0548 - 0.2400</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 4: Comparison of trauma volume among the four experimental groups; where group 1 is trauma at 75 PSI, group 2 is trauma at 35 PSI, group 3 is trauma at 55 PSI and group 4 is trauma at 90 PSI

<table>
<thead>
<tr>
<th>Groups</th>
<th>N</th>
<th>Minimum</th>
<th>Maximum</th>
<th>Mean ± SD</th>
<th>95% Confidence Interval for Mean</th>
<th>F**</th>
<th>P</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>2</td>
<td>1.0800</td>
<td>1.2000</td>
<td>1.1400 ± 0.0600</td>
<td>0.3776 - 1.9024</td>
<td>5.893</td>
<td>0.004*</td>
</tr>
<tr>
<td>2</td>
<td>13</td>
<td>0.3200</td>
<td>0.7170</td>
<td>0.4796 ± 0.0278</td>
<td>0.4190 - 0.5402</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>7</td>
<td>0.0321</td>
<td>2.2060</td>
<td>0.7873 ± 0.2562</td>
<td>0.1603 - 1.4143</td>
<td></td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>6</td>
<td>0.8440</td>
<td>1.4640</td>
<td>1.1713 ± 0.0960</td>
<td>0.9245 - 1.4182</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Total</td>
<td>28</td>
<td>0.0321</td>
<td>2.2060</td>
<td>0.7519 ± 0.0852</td>
<td>0.5771 - 0.9268</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>
Table 5: Comparison of the Mechanical properties of Surgical Stainless Steel 316L and Titanium(24).

<table>
<thead>
<tr>
<th>Metals</th>
<th>Main alloying composition (wt%)</th>
<th>Mechanical properties*</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>YS (MPa)</td>
</tr>
<tr>
<td>Stainless steel: 316L type (ASTM, 2003)</td>
<td>Fe; 16-18.5Cr; 10-14Ni; 2-3Mo; &lt;2Mn; &lt;1Si; &lt;0.003C</td>
<td>190</td>
</tr>
<tr>
<td>CoCr alloys:</td>
<td></td>
<td></td>
</tr>
<tr>
<td>CoCrWNi (F90) (ASTM, 2007a)</td>
<td>Co; 19-21Cr; 14-16W; 9-11Ni</td>
<td>310</td>
</tr>
<tr>
<td>CoNiCrMo (F562) (ASTM, 2007b)</td>
<td>Co; 33-37Ni; 19-21Cr; 9-10.5Mo</td>
<td>241</td>
</tr>
<tr>
<td>Ti and its alloys:</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pure Ti grade 4 (F67) (ASTM, 2006)</td>
<td>Ti; 0.05N; 0.1C; 0.5Fe; 0.015H; 0.4O</td>
<td>485</td>
</tr>
<tr>
<td>Ti6Al4V (F136) (ASTM, 2008)</td>
<td>Ti; 5.5-6.75Al; 3.5-4.5V; 0.08C; 0.2O</td>
<td>795</td>
</tr>
<tr>
<td>Degradable metals:</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pure iron (Goodfellow, 2007)</td>
<td>99.8Fe</td>
<td>150</td>
</tr>
<tr>
<td>WE43 magnesium alloy (ASTM, 2001)</td>
<td>Mg; 3.7-4.3Y; 2.4-4.4Nd; 0.4-1Zr</td>
<td>150</td>
</tr>
</tbody>
</table>

*under annealed condition except for WE43 which was solution heat-treated and artificially aged (T6). YS = yield strength, UTS = ultimate tensile strength, YM = Young’s modulus.
Table 6: Comparison of the buccal bone and attached gingiva thickness [30].

<table>
<thead>
<tr>
<th></th>
<th>Bone Minimum</th>
<th>Bone Maximum</th>
<th>Attached gingiva minimum</th>
<th>Attached gingiva maximum</th>
<th>Minimum sum</th>
<th>Maximum sum</th>
<th>Proposed needle length</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anterior Maxillary</td>
<td>0.75</td>
<td>1.17</td>
<td>1.2</td>
<td>1.84</td>
<td>1.95</td>
<td>3.01</td>
<td>3-4 mm</td>
</tr>
<tr>
<td>Posterior maxillary</td>
<td>1</td>
<td>1.39</td>
<td>1.05</td>
<td>1.25</td>
<td>2.05</td>
<td>2.64</td>
<td>3-4 mm</td>
</tr>
<tr>
<td>Anterior Mandibular</td>
<td>0.82</td>
<td>1.25</td>
<td>1.07</td>
<td>1.27</td>
<td>1.89</td>
<td>2.52</td>
<td>3-4 mm</td>
</tr>
<tr>
<td>Posterior Mandibular</td>
<td>1</td>
<td>2.9</td>
<td>1.02</td>
<td>1.09</td>
<td>2.02</td>
<td>3.99</td>
<td>3-4 mm</td>
</tr>
</tbody>
</table>
Table 7: Comparison of the buccal bone thickness in different maxillary regions [33].

<table>
<thead>
<tr>
<th></th>
<th>Bone Minimum (right side)</th>
<th>Bone Maximum (right side)</th>
<th>Bone Minimum (left side)</th>
<th>Bone Maximum (left side)</th>
<th>Bone Minimum (Average)</th>
<th>Bone Maximum (Average)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Canine to PM</td>
<td>0.00</td>
<td>1.25</td>
<td>0.25</td>
<td>2.25</td>
<td>0.13</td>
<td>1.75</td>
</tr>
<tr>
<td>PM to PM</td>
<td>0.38</td>
<td>1.00</td>
<td>0.25</td>
<td>1.25</td>
<td>0.32</td>
<td>1.13</td>
</tr>
<tr>
<td>PM to M1</td>
<td>0.25</td>
<td>1.25</td>
<td>0.50</td>
<td>2.25</td>
<td>0.38</td>
<td>1.75</td>
</tr>
<tr>
<td>M1 to M2</td>
<td>0.25</td>
<td>1.38</td>
<td>0.38</td>
<td>1.63</td>
<td>0.32</td>
<td>1.51</td>
</tr>
<tr>
<td>M2 to M3</td>
<td>0.25</td>
<td>1.00</td>
<td>0.38</td>
<td>1.63</td>
<td>0.32</td>
<td>1.32</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th>Bone Minimum (right side)</th>
<th>Bone Maximum (right side)</th>
<th>Bone Minimum (left side)</th>
<th>Bone Maximum (left side)</th>
<th>Bone Minimum (Average)</th>
<th>Bone Maximum (Average)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Canine to PM</td>
<td>0.5</td>
<td>1.5</td>
<td>0.38</td>
<td>1.63</td>
<td>0.44</td>
<td>1.57</td>
</tr>
<tr>
<td>PM to PM</td>
<td>0.5</td>
<td>1.5</td>
<td>0.38</td>
<td>1.88</td>
<td>0.44</td>
<td>1.69</td>
</tr>
<tr>
<td>PM to M1</td>
<td>0.38</td>
<td>1.75</td>
<td>0.5</td>
<td>1.88</td>
<td>0.44</td>
<td>1.82</td>
</tr>
<tr>
<td>M1 to M2</td>
<td>0.5</td>
<td>2.5</td>
<td>0.38</td>
<td>2.38</td>
<td>0.44</td>
<td>2.44</td>
</tr>
<tr>
<td>M2 to M3</td>
<td>0.38</td>
<td>1.25</td>
<td>0.25</td>
<td>1.88</td>
<td>0.32</td>
<td>1.57</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th></th>
<th>Bone Minimum (right side)</th>
<th>Bone Maximum (right side)</th>
<th>Bone Minimum (left side)</th>
<th>Bone Maximum (left side)</th>
<th>Bone Minimum (Average)</th>
<th>Bone Maximum (Average)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Canine to PM</td>
<td>0.5</td>
<td>1.88</td>
<td>0.63</td>
<td>2</td>
<td>0.57</td>
<td>1.94</td>
</tr>
<tr>
<td>PM to PM</td>
<td>0.38</td>
<td>2.25</td>
<td>0.63</td>
<td>2</td>
<td>0.51</td>
<td>2.13</td>
</tr>
<tr>
<td>PM to M1</td>
<td>0.38</td>
<td>2.5</td>
<td>0.63</td>
<td>2.88</td>
<td>0.51</td>
<td>2.69</td>
</tr>
<tr>
<td>M1 to M2</td>
<td>0.38</td>
<td>3.13</td>
<td>0.63</td>
<td>2.5</td>
<td>0.51</td>
<td>2.82</td>
</tr>
<tr>
<td>M2 to M3</td>
<td>0.5</td>
<td>1.88</td>
<td>0.5</td>
<td>2.5</td>
<td>0.50</td>
<td>2.19</td>
</tr>
</tbody>
</table>
Table 8: Comparison of the buccal bone thickness in different mandibular regions [33].

<table>
<thead>
<tr>
<th>Region</th>
<th>Bone Thickness at 6mm from CEJ</th>
<th>Bone Thickness at 9mm from CEJ</th>
<th>Bone Thickness at 12mm from CEJ</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Bone Minimum (right side)</td>
<td>Bone Maximum (right side)</td>
<td>Bone Minimum (left side)</td>
</tr>
<tr>
<td>Canine to PM</td>
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<td>1.13</td>
<td>0.5</td>
</tr>
<tr>
<td>PM to PM</td>
<td>0.13</td>
<td>1.88</td>
<td>0.5</td>
</tr>
<tr>
<td>PM to M1</td>
<td>0.38</td>
<td>2.63</td>
<td>0.38</td>
</tr>
<tr>
<td>M1 to M2</td>
<td>0.63</td>
<td>5.75</td>
<td>0.63</td>
</tr>
<tr>
<td>M2 to M3</td>
<td>1.5</td>
<td>9</td>
<td>1.88</td>
</tr>
</tbody>
</table>
Table 9: Comparison of the buccal bone thickness in different maxillary and Mandibular areas [31].

<table>
<thead>
<tr>
<th>Cortical Bone Thickness in the Maxilla</th>
<th>Cortical Bone Thickness in the Mandible</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Buccal</td>
</tr>
<tr>
<td></td>
<td>1.8</td>
</tr>
<tr>
<td>Premolar to M1 (occlusal)</td>
<td>1.6</td>
</tr>
<tr>
<td></td>
<td>1.6</td>
</tr>
<tr>
<td>Premolar to M1 (apical)</td>
<td>NA</td>
</tr>
<tr>
<td>M1 to M2 (occlusal)</td>
<td>1.5</td>
</tr>
<tr>
<td>M1 to M2 (apical)</td>
<td>1.6</td>
</tr>
<tr>
<td>M2 (occlusal)</td>
<td>1.3</td>
</tr>
</tbody>
</table>

Table 10: Comparison of the cortical bone thickness in different maxillary intra-oral regions [27]

<table>
<thead>
<tr>
<th>Maxillary Cortical Bone Thickness</th>
<th>Bone Minimum</th>
<th>Bone Maximum</th>
</tr>
</thead>
<tbody>
<tr>
<td>Anterior</td>
<td>1.3</td>
<td>2.2</td>
</tr>
<tr>
<td>Canine</td>
<td>1.6</td>
<td>2.4</td>
</tr>
<tr>
<td>M1 to M2</td>
<td>0.9</td>
<td>2.1</td>
</tr>
<tr>
<td>M2 to M3</td>
<td>0.6</td>
<td>1.2</td>
</tr>
</tbody>
</table>
Graphs 1, 2: Instron machine result when trauma was applied with the small needle on the synthetic bone block. It was found that needle can penetrate bone with forces ranging from 10-17 N with an average of 15 N.
Graphs 3,4: Instron machine result when trauma was applied with the medium needle on the synthetic bone block. It was found that needle can penetrate bone with forces ranging from 25-42 N with an average of 34 N
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