Experimental and numerical analysis of the biomechanical characteristics of orthodontic mini-implants

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Meiner lieben Familie gewidmet

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1 Zusammenfassung

Seit einigen Jahren werden verstärkt orthodontische Mini-Schrauben oder Mini-Implantate zur Verankerungsverstärkung eingesetzt. Trotz zahlreicher Vorteile bestehen nach wie vor widersprüchliche Ansichten in Bezug auf Einflussfaktoren, die ihre klinischen Eigenschaften bestimmen. Ziel dieser Untersuchung war es, vier verschiedene Parameter experimentell und theoretisch zu untersuchen, die einen Einfluss auf die Stabilität der Verankerungsschrauben haben könnten. Diese waren: 1) Implantattyp, 2) Implantatlänge, 3) Implantatdurchmesser und 4) Positionierung. Zwei verschiedene Kräfte, eine geringe von 0,5 N und eine höhere von 2,5 N wurden durch eine Nickel-Titan-Zugfeder (NiTi) angelegt.

Das Material bestand aus 90 Mini-Schrauben, die in frische Segmente von Rinderrippen eingesetzt wurden. Jeweils vierzig Aarhus- (American Orthodontics, Wisconsin, USA) und Lomas-Schrauben (Mondeal, Mühlheim, Deutschland) in zwei unterschiedlichen Längen (7 mm, 9 mm) und mit einem Durchmesser von 1,5 mm wurden untersucht. Die Lomas-Schrauben standen in der Länge 7 mm auch mit dem Durchmesser 2 mm zur Verfügung, um den Einfluss des Durchmessers untersuchen zu können. Die Mini-Schrauben wurden mit zwei Winkeln positioniert, jeweils eine Hälfte senkrecht, die andere Hälfte mit einer Angulation von 45° nach mesial.

An den Präparaten wurden Kraft/Auslenkungs-Diagramme im Mobilitäts-Mess-System (MOMS) des Labors der Stiftungsprofessur für Oralmedizinische Technologie, mit Kräften bis 0,5 N und 2,5 N in distaler Richtung gemessen. Die NiTi-Feder wurde auf den Hals des Mini-Implantates an der einen Seite und auf dem mechanischen 3D Kraft/Drehmoment-Sensor an der anderen Seite befestigt. Die Kraftrichtung war parallel zur Knochenoberfläche und zur Horizontalen. Jede Einzelmessung wurde zweimal durchgeführt. Anschließend wurden die Präparate in einem µCT (µCT40, Scanco Medical) gescannt und die Geometrien wurden mit dem speziell für diese Aufgabe entwickelten Programm ADOR-3D rekonstruiert. Die so entwickelten Finite-Elemente(FE)-Modelle wurden im FE-System MSC.Marc/Mentat2007r1 berechnet.

Die Statistik umfasste eine univariante Varianzanalyse (three-way ANOVA) zur Analyse des Einflusses der Parameter Schraubentyp, Länge, Positionierung und Kraft, einen *Studentschen* t-test für die Auswertung des Durchmessers und einen Altman-Bland-Test für

den Vergleich der beiden Messdurchgänge und den Vergleich zwischen den experimentellen und numerischen Ergebnissen. Zusätzlich wurde ein Youden-Plot für den Vergleich der experimentellen und numerischen Ergebnisse angefertigt.

Die Ergebnisse zeigten, dass sich das biomechanische Verhalten der Mini-Schrauben zwischen den zwei Kraftgruppen unterschied. Wenn eine geringe Kraft von 0,5 N angewendet wurde, wurden Auslenkungen des Schraubenkopfes von 4 bis 9 µm in Kraftrichtung gemessen, die Rotationen lagen bei 0,006° bis 0,025°. Die Ergebnisse schwankten zwischen den verschiedenen Mini-Implantaten, die Varianzanalyse zeigte jedoch keine statistisch signifikanten Unterschiede in den Auslenkungen.

Bei der Anwendung der höheren Kraft von 2,5 N konnte festgestellt werden, dass die 9 mm langen Mini-Implantate eine statistisch signifikant kleinere Auslenkung (Mittelwert 10±7 µm) als die 7mm langen (Mittelwert 22±11 µm, $p<0.01$) zeigten, und die 2 mm dicken Implantate auch eine signifikant kleinere (Mittelwert 0.008±0.002mm) als die 1.5 mm dick (Mittelwert 21±1 µm, $p<0.001$) aufwiesen. Die Kraft, bei der sich die Signifikanz in Bezug auf Implantatlänge und Implantatdurchmesser zeigte, wurde mit 1 N ermittelt. Der Insertionswinkel beeinflusste nicht die Stabilität der Mini-Implantate. Die LOMAS Mini-Implantate zeigten für alle Kräfte dagegen statistisch signifikant stärkere Rotationen als die Aarhus Mini-Implantate. Die numerischen Werten zeigten eine zufriedenstellende Korrelation mit den Messergebnissen, die Abweichungen lagen bei maximal 20%. Dies entspricht dem typischen Fehler einer derartigen FE-Simulation.

Zusammenfassend kann festgestellt werden, dass in klinischen Situationen, bei denen die angewandte Kraft kleiner als 1 N ist, wie zum Beispiel bei Zahnintrusionen oder indirekter Verankerung, Mini-Implantate mit kleineren Dimensionen zuverlässig eingesetzt werden können. Bei Einsatz höherer Kräfte sind entsprechende Mini-Implantat-Dimensionen entscheidend für die Primärstabilität. Jedoch sind beim klinischen Einsatz sowohl der Abstand der Zahnwurzeln als auch die anatomische Lage sorgfältig zu bedenken.

2 Introduction and review of the literature

2.1 Introduction

Anchorage in orthodontics is the resistance to unwanted tooth movement. In the field of orthodontics, several methods have been developed to overcome the critical problem of anchorage. Among them, the skeletal anchorage systems gained increasing interest. Starting with the use of vitalium screws [Gainsforth and Higley, 1945], and progressing to conventional osseointegrated implants which have been used as orthodontic anchorage [Roberts et al., 1989], onplants [Block and Hoffman, 1995], palatal implants [Wehrbein, 1996], mini-plates [Jenner and Fitzpatrick, 1985], mini-implants [Kanomi, 1997] and mini-screws [Costa et al., 1998], orthodontic therapy seems to be facilitated in an important way. In 1997, *Kanomi* described a mini-implant specifically made for orthodontic use and in 1998 *Costa et al.* presented a screw with a bracket-like head. There are different terms used describing the orthodontic anchorage implants such as palatal implants, mini-implants, mini-screws, micro-implants and micro-screws. In 2005 by *Carano and Melsen* it was agreed that the word mini-implant should be applied to all these terms. In 2005 *Mah and Bergstrand* agreed to this aspect and they pointed out that mini-implant is more appropriate than micro-implant or screw, because the word “micro” is defined as a magnitude of 10^{-6}.

Mini-implants are mainly preferred among the others, because of their comparatively much smaller size. These small dimensions allow an increase in potential intraoral placement sites, even interdentally between the roots. Due to the small size, their placement and removal are simple and the surgical trauma is restricted to the minimum. This means shorter chair time and less pain and discomfort, whilst low cost and the ability of immediate loading could be considered as additional advantages.

Despite the many advantages they present, their clinical behaviour is still unclear. The failure rates of mini-implants described in the literature are approximately 10%-30% and are still not satisfactory. Retention of mini-implant in bone depends on different influencing factors. Some of them have been reported to be the implant type, the implant dimensions [Fritz et al., 2003; Tseng et al., 2006; Berens et al., 2006], the implant surface characteristics [Kim et al., 2009], the insertion angle [Wilmes et al., 2008b], the drilling hole size [Gantous et al., 1995], the insertion torque [Motoyoshi et al., 2006], the force magnitude [Cheng et al., 2004], the anatomic location [Wiechmann et al., 2007; Tseng et al., 2006], the soft tissue characteristics [Cheng et al., 2004], inflammation of the peri-implant area [Miyawaki et al., 2003] and possible root proximity. All aforementioned factors are still under investigation and studies seem not to have drawn yet specific conclusions.

Besides the parameters that may interfere with mini-implant stability, the generally accepted protocol for successful and predictable placement of mini-implants includes atraumatic surgical technique, short healing period, biocompatible materials, and patient management. An ideal method for achieving stable implants in the initial integration stage has not yet been developed.

Another parameter that is thought to play an essential role on mini-implant retention in bone is its primary stability. We know from dental implantology that it is an implant's primary stability that largely determines how long it is retained. Numerous recent studies have thus dealt with the primary stability of microscrews and the factors affecting their stability. Primary stability is called the mini-implant stability immediately after insertion in the bone [Wilmes et al., 2006]. It is achieved due to a mechanical contact between implant and bone interfaces whereas secondary stability developes because of bone remodeling processes and is the mini-implant stability due to osseointegration. There is a critical period in terms of screw stability between these two phases of primary and secondary stability, namely, a period in which less new secondary stability is provided by bone formation than primary stability has been lost due to bone resorption. This phase of limited stability is referred to as a “stability dip”. This is the period in which implants or microscrews are at particular risk of premature loss. Despite that, it is assumed that adequate overall stability is the result of high primary stability, even during the bone-remodeling proesses induced by insertion. Primary stability is in turn also dependable on some influencing factors. These factors are said to be the implant design [Kim et al., 2008; Lim et al., 2008], bone quality [Motoyoshi et al., 2007], implant site preparation [Okazaki et al., 2008] and insertion angle [Wilmes et al., 2008b].

The primary stability, which is important for mini-implant survival, is measured in most studies by means of the maximum insertion torque or pull out strength. In this study a different experimental method was used. Following the experimental part, a subsequent numerical investigation by using the finite element analysis was performed.

2.2 Review of the literature

2.2.1 Mini-implants

2.2.1.1 Mini-implant types

The main differences between currently used mini-implants are related to their composition, size and design and include 1) the alloy used for their fabrication, 2) the diameter of the threaded portion, 3) the length of the implant and 4) the design of the neck and head [Papadopoulos and Tarawneh 2007]. As for the fabrication, most mini-implants are made of medical type IV or type V titanium alloy.

The thread may be conical or cylindrical. In the conical design, the diameter becomes progressively narrower in the apical region. This shape diminishes the risk of periodontal ligament injure in comparison with the cylindrical ones, since a greater distance between the mini-implant and the root apex is achieved [Kim et al., 2008; Poggio et al., 2006]. The diameters can range from 1.2 to 2.5 mm and the lengths from 4 to 12 mm. The head may be button like, sphere like, hexagonal, cross like, bracket like or with a hook.

The shape of the transgingival part of mini-implants may also vary and can be cylindrical, conical or polygonal. This is called the neck of the implant and it is the part that pierces the soft tissue. The length of the neck to be chosen for usage depends on the patient mucosa thickness. Except the shape and length, also the diameter of the neck varies and can be equal, wider or smaller as the diameter of the head. It is said that the design and diameter of the neck plays an important role on the accumulation of microbial plaque between the head of the mini-implant and the transgingival part.

Over 30 mini-implants are available nowadays in the market. Some representatives are shown in Figure 1 and are the Aarhus mini-implant (American Orthodontics, USA), the LOMAS pin (Mondeal, Germany), the tomas® pin (Dentaurum, Germany), the Absoanchor (Dentos, Korea), the Dual Top (Jeil Medical Corporation, Korea), the Orlus (Masel Orthodontics, PA), the Anchor Plus (KJ Meditech, Korea), the Dentis (KITA, Korea) and the Ortho-C Implant (IMTEC, USA).

Figure 1: Example of different types of orthodontic mini-implants.

2.2.1.2 Insertion methods

The insertion of mini-implants in placement sites is achieved by the predetermined tool of each company. The mini-implants are divided into self-drilling, where immediate insertion of implant in bone is possible, and to non self-drilling ones needing a pilot hole before insertion. In the non self-drilling mini-implants a low-speed contra-angle with a drill of some millimeters (mm) narrower than the mini-screw diameter is normally used for the initial entry into the bone. The drilling hole size is thought to influence stability of mini-implants and is discussed later in this paper. Even in the case of self-drilling mini-screws it is advised that a pre-drilling is required in all regions of high bone density such as the mandible and the palate, in order to diminish the insertion torque in the desired level and avoid mini-implant's metal fatigue and eventual screw fracture.

Preferably mini-implants should be inserted into attached gingival rather than in non-keratinized tissue. This could be explained by the fact that non-keratinized mucosa is movable and prone to plaque accumulation, so mechanical loosening or infection of the implant could be observed. In such cases a surgical flap to place implant under the mucosa with a ligature or hook extension is recommended.

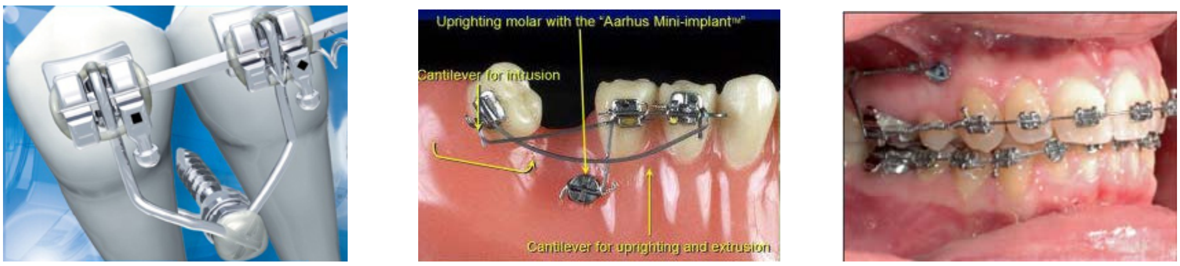
Some degree of angulation to the bone surface during mini-implant insertion, has been proposed by some investigators (ranging from 10° to 70°) to increase the surface contact between the mini-screw and the bone. This will create better mechanical contact between mini-implant and bone and will improve retention, whilst the risk of root damage is diminished as mini-implant tip is distanced from the teeth.

At last, vital tissues neighboring the mini-implant placement site should be treated with caution i.e. as when placing micro-implants in the palate where the greater palatine artery and nerve must always be avoided. Not damaging the roots of adjacent teeth is also of high importance. For that reason, surgical guides are advisable, in order to place the mini-implant in the exact desired place. The construction of the surgical guide can be self-made, from acrylic and wire. A peri-apical radiograph with the surgical guide in place is then needed to identify the accurately the mini-implant placement site.

2.2.1.3 Clinical use

Mini-implants have been utilized as anchorage units in multiple orthodontic problems with very promising results. The anchorage they provide can be characterized as *direct* or *in-direct.* Direct anchorage is the anchorage where force is applied directly on the mini-implant, whereas in indirect anchorage the mini-implant is connected and stabilized with the loaded unit (Figure 2).

Many clinical case reports and clinical studies with mini-implant application have been found in the literature. In one of them *Maino et al.* [2003] introduced the spider screw® for skeletal anchorage for prerestorative treatment in adult patients. *Fritz et al.* [2003] investigated the Dual Top in human by inserting 36 micro-implant in 17 patients and achieved molar up-righting, distalization and/or molar mesialization. Molar up-righting by means of micro implant anchorage was also reported by *Park et al*. [2004a], in patients with mesially tipped second molars, due to the loss of the adjacent first molar.



a b c

Figure 2: Examples: a) indirect anchorage (Dentaurum) b) indirect anchorage (Mondeal), c) direct anchorage in clinical case.

Freudenthaler et al. [2001] inserted 12 titanium bicortical screws horizontally as anchorage for mandibular molar protraction in eight patients. The use of mini-implants has also been reported for anterior open bite treatment [Sherwood et al., 2002; Erverdi et al., 2004; Park et al., 2004b; Kuroda et al., 2004], upper molar distalization [Gelgör et al., 2004], intrusion of maxillary incisors [Ohnishi et al., 2005], maxillary canine retraction [Herman et al., 2006], retraction of anterior segment [Park et al., 2007] and lower molar mesialization [Chung et al. 2007].

2.2.1.4 Risk factors

The long term stability of mini-implants and their retention in bone are important parameters for their success. The failure rates described in the literature are approximately 10%-30% and are still not satisfactory. Risk factors that can jeopardize their clinical performance have been attributed to mechanical and biologic reasons and are mentioned below.

a) Implant design

Differences have been reported between conical and cylindrical shaped mini-implants regarding their retention in bone, with the first ones tending to be in an advantageous position. The conical mini-implants show greater primary stability compared to the cylindrical ones as found in a study of *Wilmes et al.* [2008a]. He compared the Dual Top mini-screw and the tomas® pin and found that despite having the same dimensions the tomas® pin types showed less primary stability than the Dual Top screws. One apparent reason for that is the intraosseous part of the tomas® pin which is cylindrical, which seems inferior to those having a conical shape.

Kim et al. [2008] also showed in his mechanical study that the conical group of mini-implants showed significantly higher maximum insertion torque (MIT) and maximum removal torque (MRT) than the cylindrical group. He concludes that although the conical shaped mini-implant could induce tight contact to the adjacent bone tissue and might produce good primary stability, the conical shape may need modification of the thread structure and insertion technique to reduce the excessive insertion torque while maintaining the high resistance to removal. The same authors in 2009 [b] compared cylindrical, taper shaped and dual thread mini-implants and sates that the cylindrical shape had the lowest MIT and MRT in each length. Although taper shape showed the highest MIT in each length, when the values of insertion and removal angular momentum were analyzed (IAM and RAM), dual-thread shape showed significantly higher MRT and RAM in each length. Dual-thread groups showed a gentle increase of insertion torque and a gentle decrease of removal torque in contrast to the other shape groups. He concluded that dual-thread shape provided better mechanical stability with high removal torque on the broad range than other shapes. However, due to their higher IAM and time of MIT they need improvement to reduce the long insertion time to decrease the stress in the tissues.

b) Implant dimensions

Implant dimensions are referred to mini-implant length and diameter. The influence of these two parameters on mini-implant stability is still under investigation and studies seem to be controversial.

Many clinical studies do not correlate the length of a mini-implant to its survival rate. *Fritz et al.* in 2003 comes to the result that 4 mm long screws offer adequate stability when compared with 6 mm and 8 mm screws. *Miyawaki et al.* [2003] do not associate the length of the screw with its stability if the screw was at least 5 mm long. Also *Cheng et al*. in 2004 and *Park et al*. in 2006 agree with the abovementioned authors. The short screws used for the fixation did not jeopardize their performance, this means that longer implants did not necessarily resulted in greater bone support [Park et al., 2006].

On the contrary in a study of *Tseng et al.* [2006] the length of the inserted mini-implants was an important risk factor. They emphasize that the actual depth of insertion of the mini-implant was more important than its length, the recommended length being at least 6 mm. This is in accordance with dental implantation, where the shorter and smaller diameter implants had lower survival rates than their counterparts [Winkler et al., 2000].

As for the implant diameter, most studies have shown that implant diameter has a great impact on the insertion torque of orthodontic mini-implants. According to Wilmes et al. [2008a] mini-implants with 2 mm diameter showed significantly higher insertion torque when compared with mini-implants with a 1.6 mm diameter. The study of *Lim et al.* in 2008 *Miyawaki et al*. [2003] was in accordance to these findings and thought that the diameter of the mini-implants was significantly associated with their stability. The later reported that 1 year success rate of mini-implants with a 1 mm diameter was significantly less than that of mini-implants with diameters of 1.5 and 2.3 mm. They also found that patients with a high mandibular plane angle showed a significantly lower success rate than those with an average or low angle. This could be attributed to the fact that the thickness of buccal cortical bone in subjects with high mandibular plane angle was thinner than that in subjects with a low angle in the mandibular first molar region. They conclude that the wider implants should be especially placed in patients with vertical facial growth. A study of *Berens et al.* 2006 was in accordance with the previous statements since they found that mini-implants of a diameter of 2 mm in lower jaw increases success rate. They also recommend a mini-screw diameter of at least 1.5 mm in the palatal upper jaw. It has been generally suggested that implants smaller than 1.3 mm should be avoided, especially in the thick cortical bone of the mandible [Carano et al., 2005].

On the contrary *Ohmae et al.* [2001] showed that mini-screws, 1 mm in diameter and 4 mm in length, placed in the mandibular third premolar region of beagle dogs were able to sustain an intrusive force of 1.5 N for 12 to 18 weeks. *Cheng et al*. [2004] states that implant types of identical configuration show no difference in their success.

In our study the parameters of length and diameter of mini-implants were examined to find possible correlations with implant stability. The results are described in the corresponding chapter.

c) Insertion procedures

Insertion angle

The angle of mini-implant insertion is proposed by some investigators to be less than 90°, because an oblique rather a straight insertion is thought to increase contact between implant and bone. *Melsen et al.* in 2005 recommends the placement of mini-implants at such an oblique angle both in the maxilla and mandible in an apical direction. The degree of angle proposed varies between authors. A 30° to 40° angulation in the maxilla and a 10° to 20° in the mandible are proposed by *Kyung et al*. [2003]. *Carano et al.* [2005] also suggested an angulation of 30° to 45° in the maxilla. In a study of *Wilmes et al*. [2008b] the angle of mini-implant insertion influenced the measured insertion torque According to them the reason for this may be the longer distance through cortical bone when the implant is inserted in an oblique direction despite the smaller insertion depth. For them an insertion angle ranging from 60° to 70° is advisable.

Drilling

As mentioned above there are two types of mini-implants with regard to the insertion method, the self-drilling and the non self-drilling ones, the latest that require a drilling hole. The drilling hole diameter is found to have a major impact on the measured insertion torques which in turn seems to be an indicator for mini-implants retention in bone. The larger the measured insertion torque, the better the implant stability achieved. In a systematic review [Chen et al., 2009] it was concluded that in studies where a 1.5 mm diameter pilot drill was used for the 2 mm diameter implants, the survival rates were 85% to 100%. According to *Melsen et al.* [2005] the pilot drill should be usually 0.2 to 0.3 mm thinner than the mini–implant diameter. In accordance to them *Park et al*. [2006] used a

0.9 mm diameter drill for 1.2 mm mini-implants for an over 90% overall success rate. Concluding, the larger the pre-drilling diameter, the smaller the insertion torque according also to reports of *Wilmes et al.* [2009].

Comparing self-drilling and non self-drilling (self tapping) mini-implants during implantation *Su et al.* [2009] found that the self-tapping implants typically had a lower insertion torque than the self-drilling implants. Based on the displacements under lateral loading, however, both the self-tapping and self-drilling implants showed similar resistance to lateral forces. Nevertheless, *Wang and Liou* [2008] compared the performance of self-drilling and self-tapping mini-implants under orthodontic force and draw the conclusion that it didn’t differentiate in both mini-implant types inserted both in the maxilla.

d) Insertion torque/pull out strength

Insertion torque (IT) is the result of frictional resistance between screw threads and bone. Axial pull out strength (PS) reflects the magnitude of the PS that the screw bears before bone rupture. Both methods have been used to determine mini-implant retention in the bone. A correlation between IT and PS was found by many authors even though other studies concluded that this correlation does not exist.

Insertion torque is said to determine primary stability [Deguchi et al., 2006; Wilmes et al., 2006]. And as known, a sufficient primary stability measured by insertion torque seems to play a major role for the treatment time survival rate [Motoyoshi et al., 2006]. This is also proven in dental implantology. Insertion torque levels must range between certain limits, since very low or very high values can be critical for mini-implant success. *Motoyoshi et al.* [2006] reported higher loss rates when the insertion torque exceeds 10 Ncm for mini-implants with a diameter of 1.6 mm. A torque value of more than 15 Ncm recorded at the time of insertion appears to be one of the critical variables for mini-implant survival under immediate loading according to *Chaddad et al.* [2008]. The high torque values may result in higher failure rates due to bone compression, local ischemia, necrosis and micro damages [Wawrzinek et al., 2008].

Placement torque correlates directly with cortical bone thickness. Other aspects influencing IT are the bone quality and quantity, the drilling hole, screw characteristics and insertion technique, continuous or intermittent rotation and dry or wet conditions. *O’Sullivan et al*. [2004] reported that insertion torque values differ according to implant type and higher values of insertion torque show higher interfacial stiffness at the implant-bone interface.

e) Load

The time of loading has been investigated in many researches. Many authors support the fact that mini-implants can be loaded immediately, but some allow healing periods of some weeks or even months for a better outcome.

Miyawaki et al. [2003] suggested that immediate loading of a screw-type implant anchor is possible if the applied force is less than 2 N. Immediate loading is probably possible because of successful mechanical integration between the implant anchor and the alveolar bone. This means that if primary stability of mini-implant is adequate it is possible to load it immediately [Kyung et al., 2003]. A finite element analysis found that an immediately loaded implant should be limited to 50 cN of force in a 2 mm diameter mini-implant. Other studies do not correlate immediate loading and mini-implant success rate.

With regard to the magnitude of orthodontic load, it was found that a load in the range of 1 to 2 N could be well sustained by the mini-implants while no significant difference was noted in the magnitude of load between successful and failed implants [Cheng et al., 2004]. *Roberts et al*. [1989] also stated that forces between 1 and 3 N did not affect the implant stability. In a study, *Kyung et al.* [2003] mentioned that even smallest mini-implants can withstand as much as 4.5 N of force, whereas most orthodontic applications need forces of less than 3 N. *Liou et al.* [2004] supplied a 4 N loading on the implants at the zygomatic buttress of the maxilla to create a mass retraction of the anterior teeth and all 32 mini-screws remained stable clinically for 9 months. On the other hand, *Buechter et al.* in 2006 showed that tip forces higher than 600 cN resulted in a high risk for osseointegration loss, agreeing with *Isidor* [1997] who noticed that high forces tend to damage the interface integration.

Duration of force may also contribute to implant stability risk. *Serra et al.* [2008] placed 2 mm wide and 6 mm long mini-implants in rabbits and analyzed interfacial healing 1, 4 and 12 weeks after placement. The immediate 1 N load did not cause significant changes in the fixation of the mini-implants after 1 and 4 weeks of bone healing. Nevertheless, after 12 weeks, the loaded group had significantly lower removal torque (RTT) values than the unloaded group. As for the direction of force, force system generating a moment in the screw in the unscrewing direction is associated with failure as reported from *Costa et al.* [1998], whereas methods of force application do not matter according to *Park et al*. [2006].

f) Anatomic location and bone parameters

Mini-implants can be placed both in maxilla and mandible, but investigators have shown that placement site may influence their performance. Possible sites in the maxilla are the nasal spine, the palate, the infra-zygomatic crest, the maxillary tuberosities and the alveolar process. In mandible insertions have been reported in the symphysis, the alveolar process and the retro-molar area.

Cortical bone thickness (CBT) and density can vary according to the region of placement. Areas with thick cortical bone are considered the most stable for mini-implant placement. Since retention depends essentially on the bone-metal interface, the greater the bone, the better the primary stability. On the other hand, the higher the bone density the greater the bone pressure and bone damage during insertion.

Implants in the posterior maxilla had longer survival than in the posterior mandible. Implants in the posterior versus anterior mandible were also prone to failure. This may be attributed to the higher susceptibility to infection in the posterior mandible, mainly because less attached gingival is available in this region and to the higher bone density where overheating is more likely to occur [Cheng et al., 2004]. *Berens et al*. [2006] warned not to place mini-screws in the lingual side of the lower jaw, due to the technical demand during insertion and the patients tongue interference. *Park et al.* [2006] on 227 screw implants showed higher failure rate in the mandible (13.6% for the mandible and 4% for the maxilla). Other investigators could not identify a difference in failure rates between maxilla (15.9%) and mandible (16.4%) [Miyawaki et al., 2003; Motoyoshi et al., 2006]. In the mandible the safest sites are between first and second molars and premolars [Poggio et al., 2006] and mesial or distal to the first molar [Deguchi et al., 2006].

In maxilla the best insertion sites are in the anterior and apical portion [Poggio et al., 2006]. In the maxilla, *Berens et al.* [2006] observed quite high loss rates on the palatal side of the upper jaw. In this case mucosal thickness came into play. The palatal mucosa is 5 mm thick in some parts which automatically leads to a long lever arm, which is a decisive factor in the loss of the screw. In palate, the midpalate, and 3 to 6 mm to the paramedian region offer sufficient bony support [Bernhart et al., 2000]. *Baumgaertel* [2009] found that CBT decreased from anterior to posterior palate and recommends a placement site in premolar region. The same holds for *Kang et al*. [2007] who found that the midpalatal area within 1 mm of the midsagittal suture had the thickest bone available in the whole palate. The thickness tended to decrease laterally and posterior. So, when a mini-implant could deviate from the midpalatal area by more than 1 mm, they recommend placing it not far posterior or using a shorter mini-implant.

Concluding we could say that there is evidence that cortical bone thickness (CBT) can have strong influence on primary stability of mini-implants. *Motoyoshi et al.* in 2007 and *Motoyoshi et al.* in 2009 (a) found in both studies that success rates in the groups with CBT ≥ 1 mm were significantly higher than those in the groups with CBT ≤ 1 mm. Interdentally cortical bone thickness varies in the upper and lower jaw and a distinct pattern appears to be present. The knowledge of this pattern and the mean values of thickness can aid in mini-implant site selection and preparation.

g) Surface characteristics

The surface of the intra-osseous part of mini-implant is mostly treated mechanically, but there are also cases where sandblasting and acid etching is performed. Mechanical and surface treatments seem to provide better osseointegration and can help to increase their stability. The preference between a large-grit sandblasting and acid etching (SLA) or a mechanical preparation depends on the desired clinical outcome of mini-implants, since the type of surface preparation is seemed to influence the degree of osseointegration.

In a study of *Kim et al*. [2009a] the maximum insertion torque value and insertion angular momentum were significantly lower in the SLA group than in the machined group, but showed higher removal energy, indicating that SLA surface treatment had influenced the osseointegration potential. On the contrary, in a study on the success rates of surface treated mini-implants, surface characteristics did not appear to influence survival rates of immediate loaded mini-implants [Chaddad et al., 2008].

h) Other factors

Root proximity is referred as a critical factor for implant survival. *Kuroda et al.* [2007] classified the inserted screws in their study according to its proximity to the root. In category I, the screw was absolutely separate from the root; category II, the apex of the screw appeared to touch the lamina dura; and category III, the body of the screw was overlaid on the lamina dura. There were significant differences in the success rates between categories I and II, I and III, and II and III. Although screws in all 3 categories in the maxilla and categories I and II in the mandible showed high success rates above 75%, screws in category III in the mandible had a low success rate of 35%. He concludes that the proximity of a mini-screw to the root is a major risk factor for the failure of screw anchorage and this tendency is more obvious in the mandible.

Motoyoshi et al. 2009 in a FE study stimulated four categories as further: the implant touches nothing; the implant touches the surface of the periodontal membrane; part of the screw thread is embedded in the periodontal membrane; and the implant touches the root. Maximum stress on the bone increased when the mini-implant was close to the root. When the implant touched the root, stress increased to 140 MPa or more and bone resorption could be predicted.

Patient-related factors such as age and gender seem not to influence success rates in most publications, although in one study where computed tomography was used measured cortical bone was thinner in females in the attached gingiva mesial to the maxillary first molar.

Physical and dental status such as osteoporosis, uncontrolled diabetes, periodontal disease, smoking and pharmacologic prescriptions such as biophosphonates are considered risk factors for classic dental implants. It is probably wise to avoid the use of mini-implants in these patients [Reynders et al., 2009].

Soft tissue characteristics are also an implant maintenance related factor. The necessity of peri-implant keratinized mucosa for the maintenance of implant health has long been a debatable issue for endosseous dental implants. However, retrospective clinical surveys have failed to reveal major differences in the survival of implants placed in keratinized or non keratinized mucosa. *Warrer et al* [1995] discovered that absence of keratinized mucosa around endosseous implants increased the susceptibility of the peri-implant region to plaque induced tissue destruction. This is in accordance to the findings of *Cheng et al*. [2004] who found that absence of keratinized mucosa around mini-implants significantly increases the risk of infection and failure.

2.2.1.5 Osseointegration or not?

As widely known, osseointegration is not assumed for mini-implants as only the mechanical contact between bone and implant interface is necessary to provide stability. This is the reason of immediate loading ability of mini-implants, since no healing period is awaited. However, osseointegration in mini-implants was found to be present in many studies and these investigators recommend a waiting period prior to force application.

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

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

Because complete osseointegration of mini-implants used in orthodontic therapy is not wanted due to the complications during removal, most of them are manufactured with a smooth surface which impairs the development of bone formation. Despite the amount of osseointegration that may occur it is thought that removal is not difficult since coherence is relatively low as active remodelling and less mineralized bone formation takes place in the bone around the loaded screw part [Serra et al., 2008].

2.2.2 The finite element analysis

2.2.2.1 The method

The finite element method (FEM) is a numerical method from engineering science. It is a computer aided mathematical technique for obtaining approximate numerical solutions to the abstract equations of calculus that predict the response of physical systems subject to external influences [Knox et al., 2000].

In the FEM the body to be analyzed consists of a large number of small elements in a form of a mesh that are sufficient to describe the geometry of the subject (Figure 3 and 4). Elements consist of element edges and element faces and are connected by points called nodes. The nodes are characterized by their global co-ordinates and symbolized by a spot on the screen. According to the number of nodes, the shape of element can be in form of line, triangle, square, or a bended element (Figure 3). The number and volume of elements differ between structures and depend on their mechanical properties (density, Young's modulus of elasticity). The element types (two-dimensional or three dimensional) and their material properties are chosen to represent the properties of the physical model.

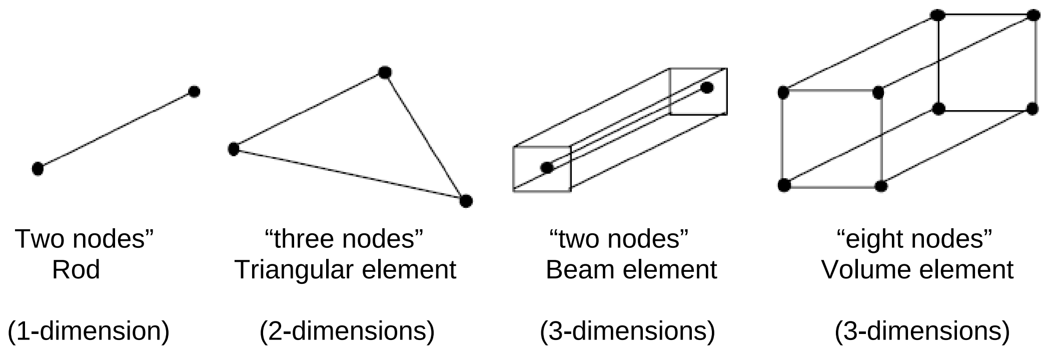


Figure 3: Different types of elements (from MSC Software Corporation, 2007).

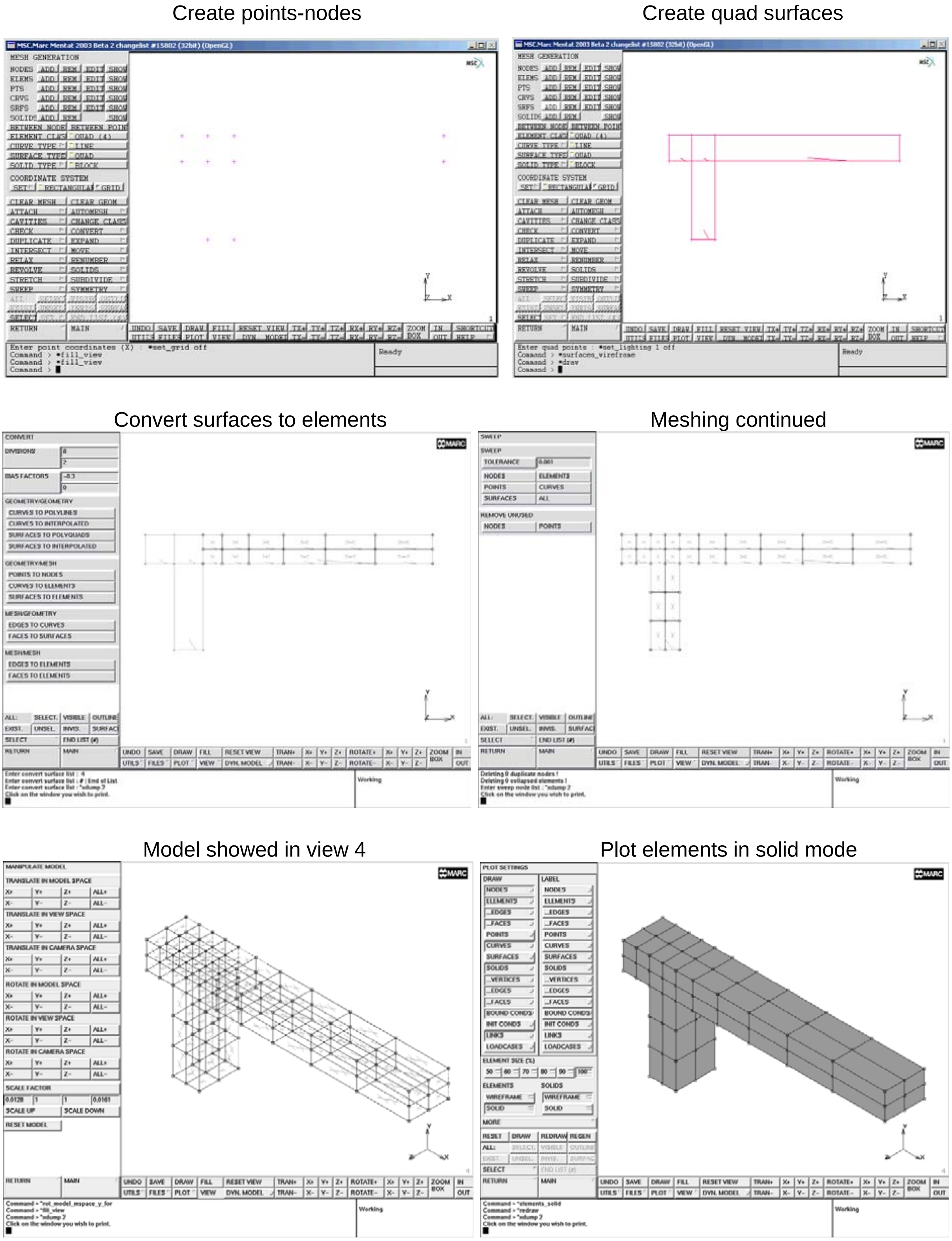


Figure 4: Example of computer aided step-by-step numerical model generation of the finite element analysis (from MSC. Software Corporation, 2007).

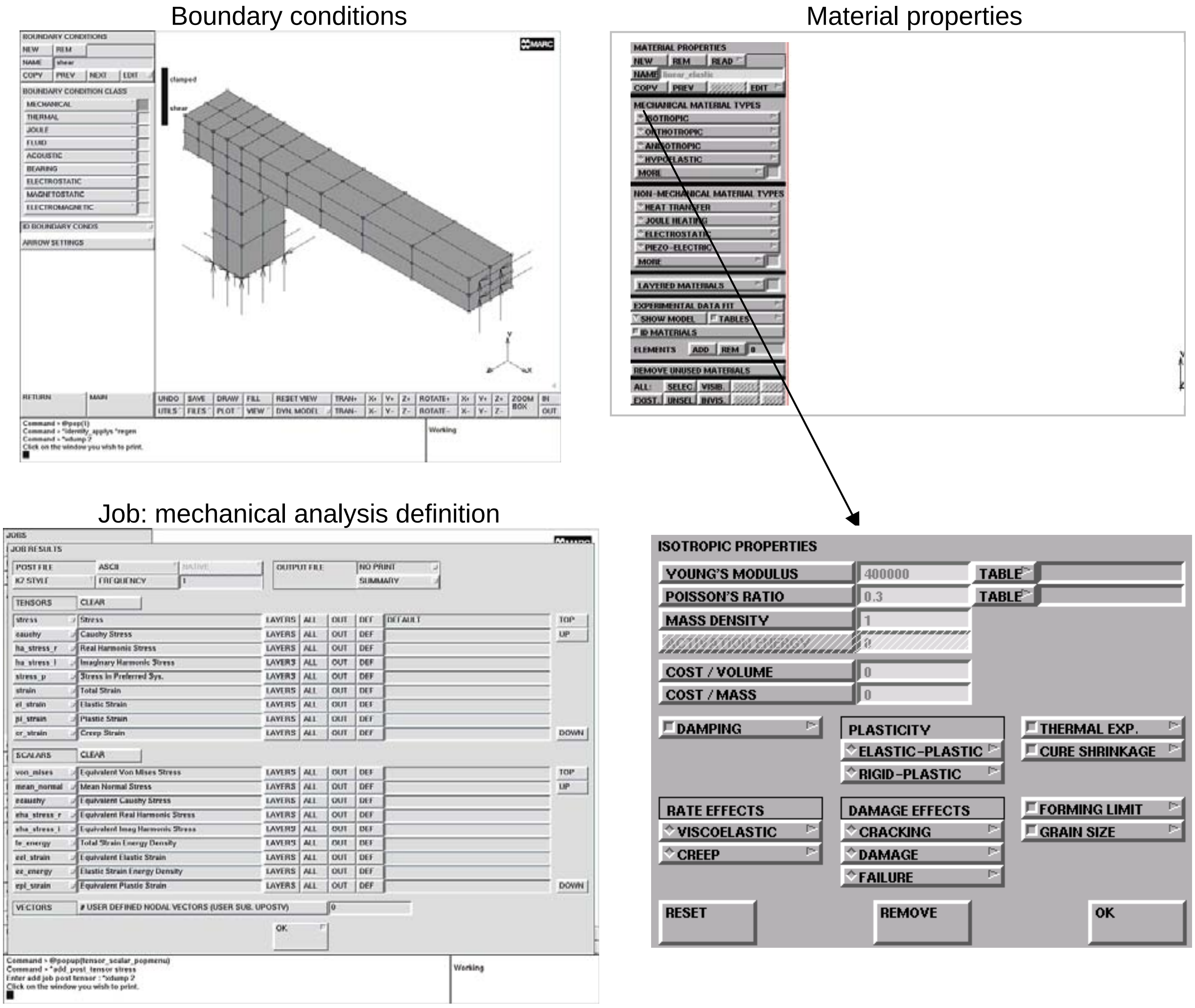


Figure 4: (continued).

The above procedure shown in Figure 4 is only a brief example of the steps needed for a model generation. Depending on the complexity of the examined structure and the desired analysis, the finite element model generation and its numerical calculation can vary from quite simple to very complex and highly sophisticated. Good numerical results depend not only on the familiarization with the chosen finite element program, but also on the good knowledge of the structures under investigation. Material sciences, physics and mathematics are major components for the generation of a finite element model and the conduction of a numerical calculation. Numerical simulations of complex configurations should be made from users with deep knowledge in those fields, in order to have a successful result.

2.2.2.2. The finite element method in dentistry and orthodontics

The FE method has been introduced in dental biomechanical research in 1973 [Farah et al., 1973] and since then has been extensively applied to analyze the stress and strain in the alveolar supporting structures and especially in the periodontal ligament [Kawarizadeh et al., 2003; Natali et al., 2007]. Several researches were also performed to investigate dental materials [Keilig et al 2009] and also dental implants with FEM. *Gallas et al.* [2005] performed a FE model of an endosseous implant and its surrounding osseous structure and found out that the highest stress when the implant is used for orthodontic anchorage was located in the cervical margin.

FEM has already been broadly applied also in orthodontic research. *Yettram et* al. [1977] were amongst the first to employ a two-dimensional finite element model of a maxillary central incisor to determine the instantaneous centre of rotation of this tooth during translation. *Halazonetis* [1996] used a similar two-dimensional model to determine periodontal ligament (PDL) stress distribution following force application at varying distances from the centre of resistance of a maxillary incisor. Using more complex three dimensional models *Tanne et al*. [1987], *McGuinness et al.* [1991] and *Wilson et al.* [1991], have studied moment to force ratios and stress distributions during orthodontic tooth movement. In the field of dentofacial orthopaedics, finite element models have been employed to evaluate the stress distribution induced within the craniofacial complex during the application of protraction headgear [Tanne et al., 1991], orthopaedic chin cup forces [Tanne et al., 1993] and conventional headgear forces [Tanne and Matsubara, 1996]. The finite element method has also been applied to the evaluation of orthodontic attachment. *Ghosh et al.* [1995] have used three dimensional FEM models of ceramic orthodontic bracket designs to determine the stress distribution and cohesive failure within the bracket when a full dimension stainless steel arch wire is engaged within the bracket slot. *Katona* [1994] and *Katona and Moore* [1994] have used a two-dimensional finite element model of the bracket tooth interface to assess the stress distribution in the system when bracket removing forces are applied. Similarly, *Rossouw and Tereblanche* [1995] have used a simplified three dimensional finite element model to evaluate the stress distribution around orthodontic attachments during debonding. In a study of *Reimann et al*. (2009) the purpose was to analyse the biomechanical behaviour of posterior teeth under headgear traction with neighbouring teeth in different eruption stages by using a finite element (FE) model of the right part of a human maxilla. *Bourauel et al.* (2009) described in detail the application of finite element methods in orthodontic biomechanics with the help of several typical examples.

In recent years, interest is focused in the FEM investigation of orthodontic mini-implants regarding their biomechanical performance and the stress distribution in the surrounding bone [Motoyoshi et al., 2005; Motoyoshi et al., 2008; Motoyoshi et al., 2009a, b; Stahl et al., 2008].

3 Aim of the study

In this study two different types of mini-implants were examined to describe possible effects of influencing parameters on primary stability of orthodontic mini-implants. The mini-implants used were the Aarhus mini-implant and the LOMAS pin provided in different lengths and diameters and loaded with two force levels.

The first part of the study included the experimental investigation where mini-implant deflections were registered by a customised biomechanical set-up.

In the second stage the finite element method was used for the numerical analysis of the generated 3D reconstructed models.

The aim of this study was to experimentally and theoretically examine the influence of four different parameters on mini-implant primary stability, by measuring their deflection during orthodontic force application. These parameters were:

1) implant type,
2) implant length,
3) implant diameter and
4) insertion angle.

Two different force levels, a low force of 0.5 N and a high one of 2.5 N were applied.

4 Material and methods

4.1 Material

4.1.1 Mini-implants

A total of 90 conical-shaped titanium mini-implants from two different companies, 40 Aarhus mini-implants (American Orthodontics, Sheboygan, Wisconsin, USA) and 40 LOMAS mini-screws (Mondeal, Mühlheim, Germany) with identical design were available for this study. Each type of mini-implant was available in two different lengths (7 mm and 9 mm) and in the same diameter of 1.5 mm. Ten pieces of LOMAS pins of 7 mm length were also available in a wide diameter of 2 mm, in order to examine the influence of diameter width on implant stability (Figure 5). The final sample consisted of 62 carefully selected preparations (Table 1), since 28 mini-implants were not included in the final data analysis due to several reasons (improper insertion, improper model preparation, fracture, distorted measurements due to external noise effects).

a b c

Figure 5: Tested mini-implants: a) Aarhus 1.5x7 mm and 1.5x9 mm b) LOMAS 1.5x7 mm and 1.5x9 mm c) LOMAS 2x7mm.

Implant type	Implant dimensions	Groups	n	
			F=0.5 N	F=2.5 N
Aarhus	1.5x7 mm	1	7	6
	1.5x9 mm	2	8	8
LOMAS	1.5x7 mm	3	7	6
	1.5x9 mm	4	6	6
	2.0x7 mm	5	5	3

Table 1: Mini-implant types and dimensions under investigation.

4.1.2 Animal bone

The animal bone consisted of fresh segments of bovine ribs. Each bovine rib was segmented in a number of small bone pieces, which served as placement sites of each mini-implant.

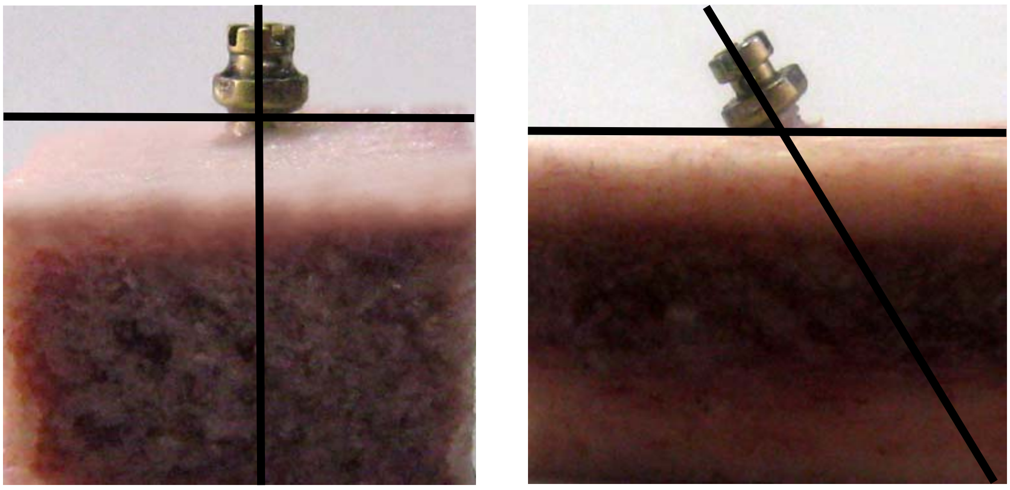
Figure 6: Bone models with inserted mini-implants.

Bovine ribs present the same architectural pattern as the human mandible, with a clear defined cortical and spongious bone. Cortical bone thickness was around 2 mm as clinically measured. Although bone density of bovine ribs is quite higher than that of human mandible, bone quality is not an influencing factor in this study, since only the biomechanical characteristics of mini-implants, placed in the same bone area, were investigated.

4.2 Experimental method

4.2.1 Insertion procedure

The self-drilling mini-implants were inserted into the bone segments using the predetermined tools of the respective company. Half of the mini-implants were inserted straight, the other half with an angulation of 45° to the bone surface (Figure 7a, b). Prior to insertion, the periosteum was removed from each bone piece.



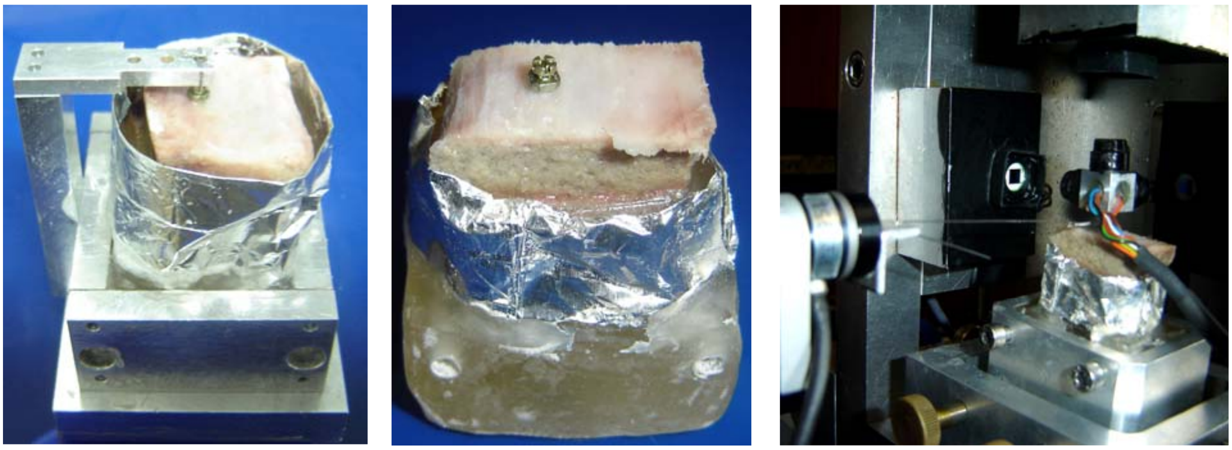
a b

Figure 7: Insertion of mini-implants in two different modes: a) vertical, b) with a 45º of angulation to the bone surface.

4.2.2 Specimen preparation

In order to engage the bone blocks in the measurement system described below two metallic cubes of the same dimensions but with different usage were needed (Figure 8a, b, c). The first one was used to position the mini-implant in certain place in space. This was done by a custom-made grip made of ligature wire, which attached the mini-implant to the lever arm of the metallic cube. The mini-implant was fixed in a certain position, in a way that it would match the desired place in the mechanical testing machine, where it would be transferred. Autopolymerising dental acrylic resin was mixed and poured into the metallic cube. Foil was used to help the support of the acrylic. The acrylic level was limited in the lower third part of the bone piece. So, the most bone elements were free of resin. During the exothermic reaction of the acrylic and also during the whole experiment, the bone was cooled and moisturized with a 0.9% saline solution.

The construction designed was then transferred to the second metallic cube which served as sample holder. The sample holder containing the preparation was then adapted and stabilized to its specific place in the optomechanical system during the experiment. After this procedure the base holder, the sample holder and the preparation became a rigid body and only the movements of mini-implants were to be registered during force application.

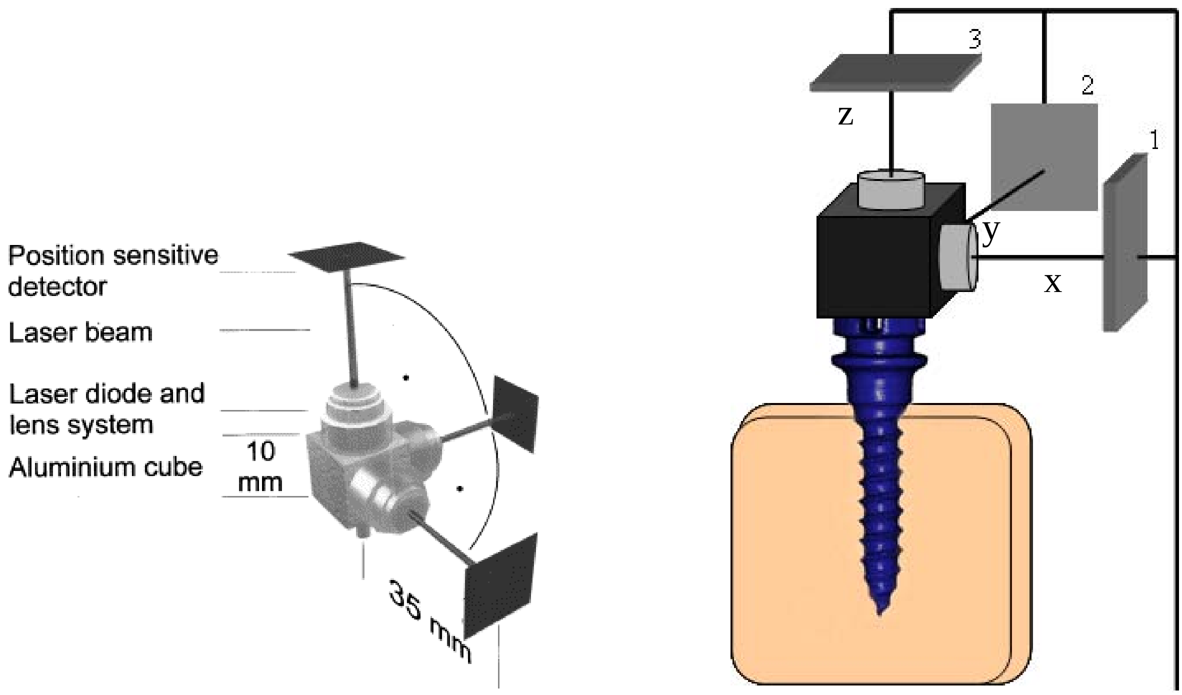


a b c

Figure 8: a) Model orientated in space using the first metallic cube, b) model ready for measurement, c) model in sample holder just before force application.

4.2.3 Optomechanical system

Following implant insertion the preparations were mounted in the 3D mobility measurement system (MOMS) [Hinterkausen et al., 1998]. The MOMS consisted of two components, a mechanical and a laser-optical subsystem. The mechanical system, serving for load application splits up into three basic components: a force/torque transducer (ATI FT Nano 12, SCHUNK GmbH & Co. KG, Lauffen/Neckar, Germany), a stepping motor driven positioning table and a computer running the control software. The laser-optical subsystem registered the implant displacements and rotations non-invasively in all three planes of space (Dx, Dy, Dz, Rx, Ry, Rz). This was achieved by an aluminium cube equipped with three laser diodes on three sides each. The laser beams of the cube were focused on planar positioning sensing detectors (PSD). The data collected were subsequently shown as force/deflection curves. The laser system was fixed on top of each mini screw with an instant adhesive, thus defining a Cartesian rigid body coordinate system (Figure 9a, b; Figure 10). The accuracy of the laser-optical system has been confirmed to be 0.1 mm and 0.2 degrees for registration of tooth or implant mobility [Hinterkausen et al., 1998].



a b

Figure 9: a) Laser system of aluminium cube with laser diodes focusing on the three position sensitive detectors, b) schematic diagram of the configuration.

a) Force application

Force was applied on the mini-implants through closed nickel titanium (NiTi) coil springs (American Orthodontics, USA). The NiTi coil springs were attached on the neck of the mini-implants on one side and on the mechanical 3D force/torque transducer on the other side via wire ligatures. The force axis was parallel to the bone surface and to the true horizontal (Figure 10). Two force levels were applied, a low and a high one: half of the mini-implants were loaded with a force of 0.5 N, the other half with a force up to 2.5 N. Force level was gradually increased from zero to the corresponding maximum point. Maximum load was applied in a total of 10 and 20 incremental steps, respectively. Each preparation was loaded and measured twice in a way that two deflections values were available in the later analysis.

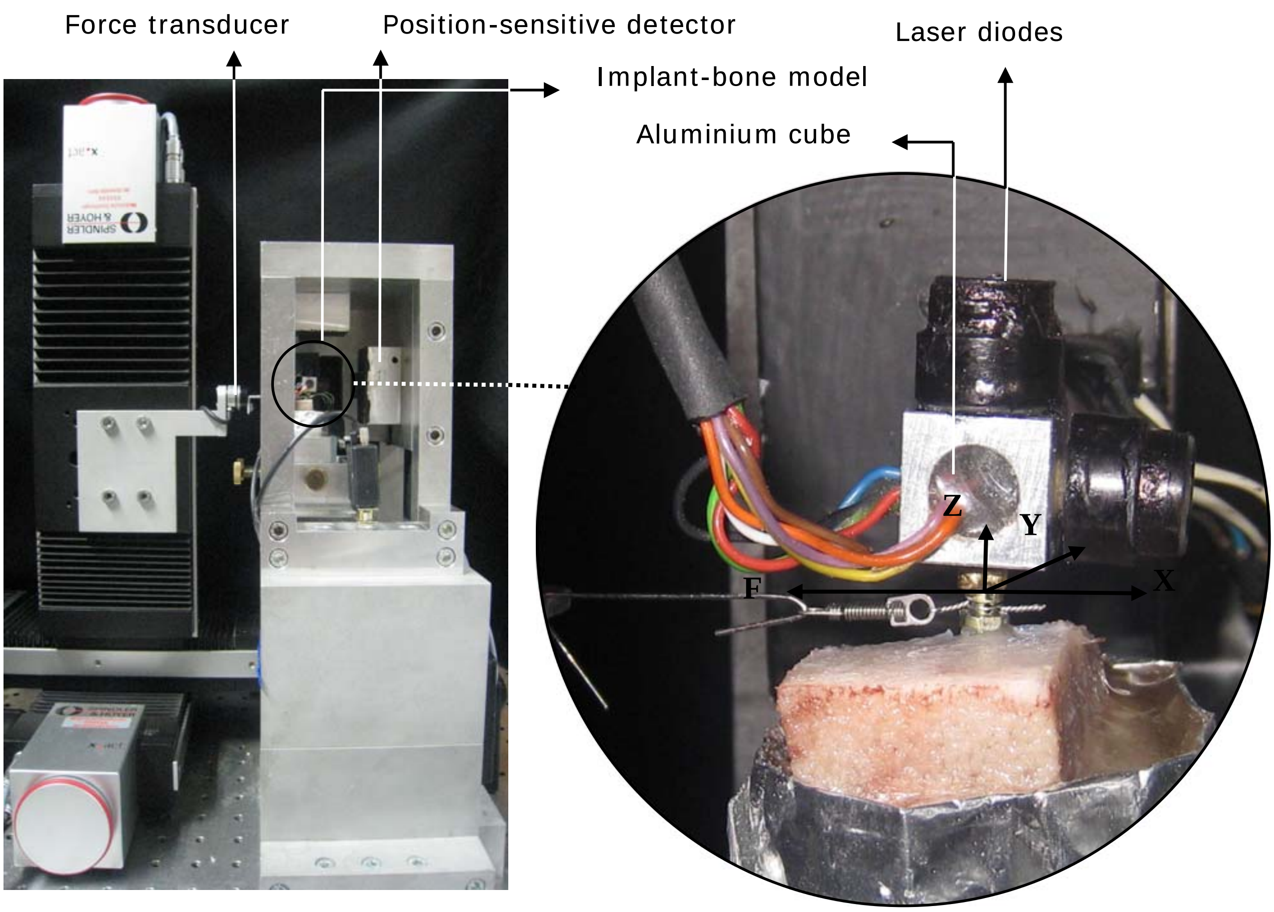


Figure 10: Preparation mounted in the optomechanical system. Force application via closed NiTi coil spring along the x-axis.

b) Implant deflection registration

Displacement and rotation were measured at every step during loading and were available in all three coordinates. The interest on this study was focused on mini-implant displacements (Dx) along the direction of force (x-axis) and on mini-implant rotations around the y-axis (Ry) (Figure 9, 10). Each measurement was repeated twice, to examine possible intra-observer error. The steps of mini-implant loading and mini-implant deflection registration are shown in Figure 11.

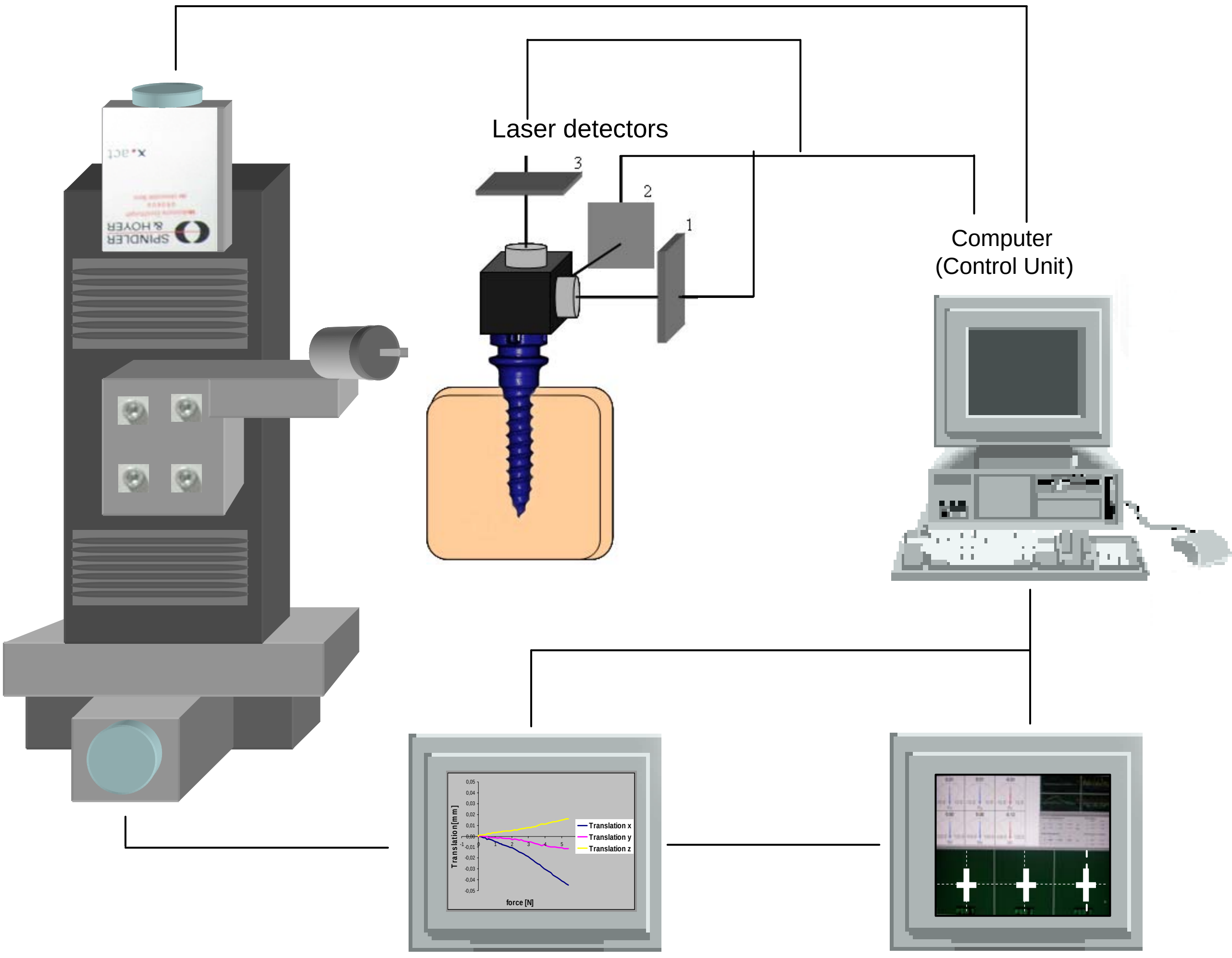


Figure 11: Schematic representation of the experimental set up. Left to right: Mechanical load application system, preparation with pin and laser cube, control computer. Below: Force deflection diagram and control screen.

4.3 Numerical method

4.3.1 Steps of 3D model reconstruction

4.3.1.1 µCT scanning

Following measurements of force/deflection curves, one specimen of each group loaded with the small force only (nine in total, Table 2) was randomly chosen and sent for scanning in a micro-CT scanner (µCT40, SCANCO Medical AG, Brüttisellen, Switzerland; Figure 12). The number of slices in each preparation ranged from 800 to 1035.

Implant type	Implant dimensions	Force	Insertion mode	Groups	n (scanned models)	Numerical model
Aarhus	1.5x7mm	0.5N	straight	1	1	A
		0.5N	angled	2	1	B
	1.5x9mm	0.5N	straight	3	1	C
		0.5N	angled	4	1	D
LOMAS	1.5x7mm	0.5N	straight	5	1	E
		0.5N	angled	6	1	F
	1.5x9mm	0.5N	straight	7	1	G
		0.5N	angled	8	1	H
	2x7mm	0.5N	straight	9	1	I

Table 2: Nine groups of preparations loaded with the small force level were constructed experimentally. One piece of each group was scanned for the FE analysis.

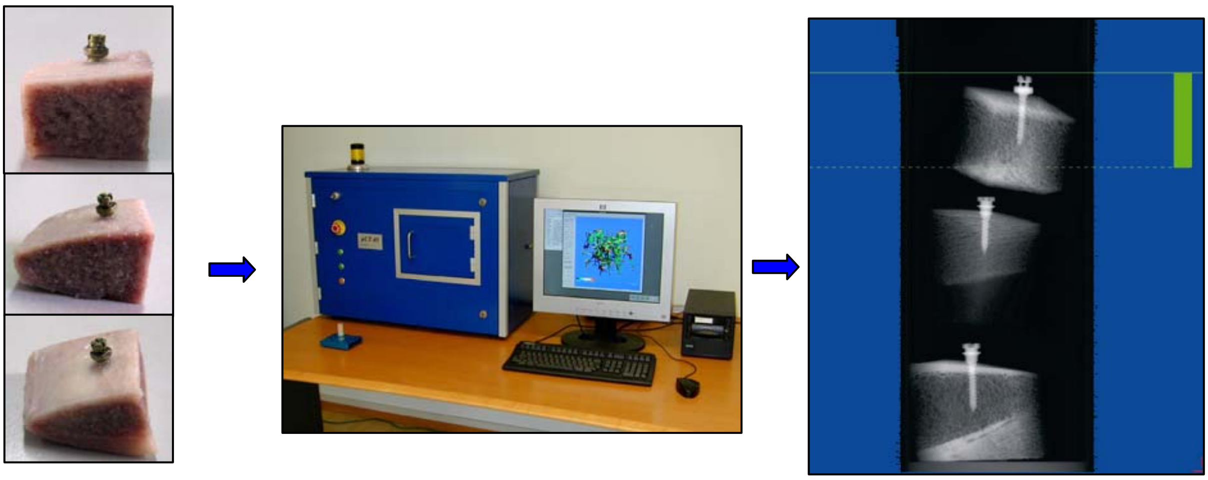


Figure 12: Example of the scanning procedure of three models in the µCT scanner.

4.3.1.2 Surface model generation

Based on the sectional views of the µCT scans a 3D surface reconstruction of the preparations including implant and bone was conducted using the custom-developed software ADOR-3D [Rahimi et al., 2005] (Figure 13). First of all the boundaries of the structures (implant, cortical and spongious bone) were identified and marked in all slices in the cutting plane. Secondly the boundaries were discretized and a 3D surface model was generated.

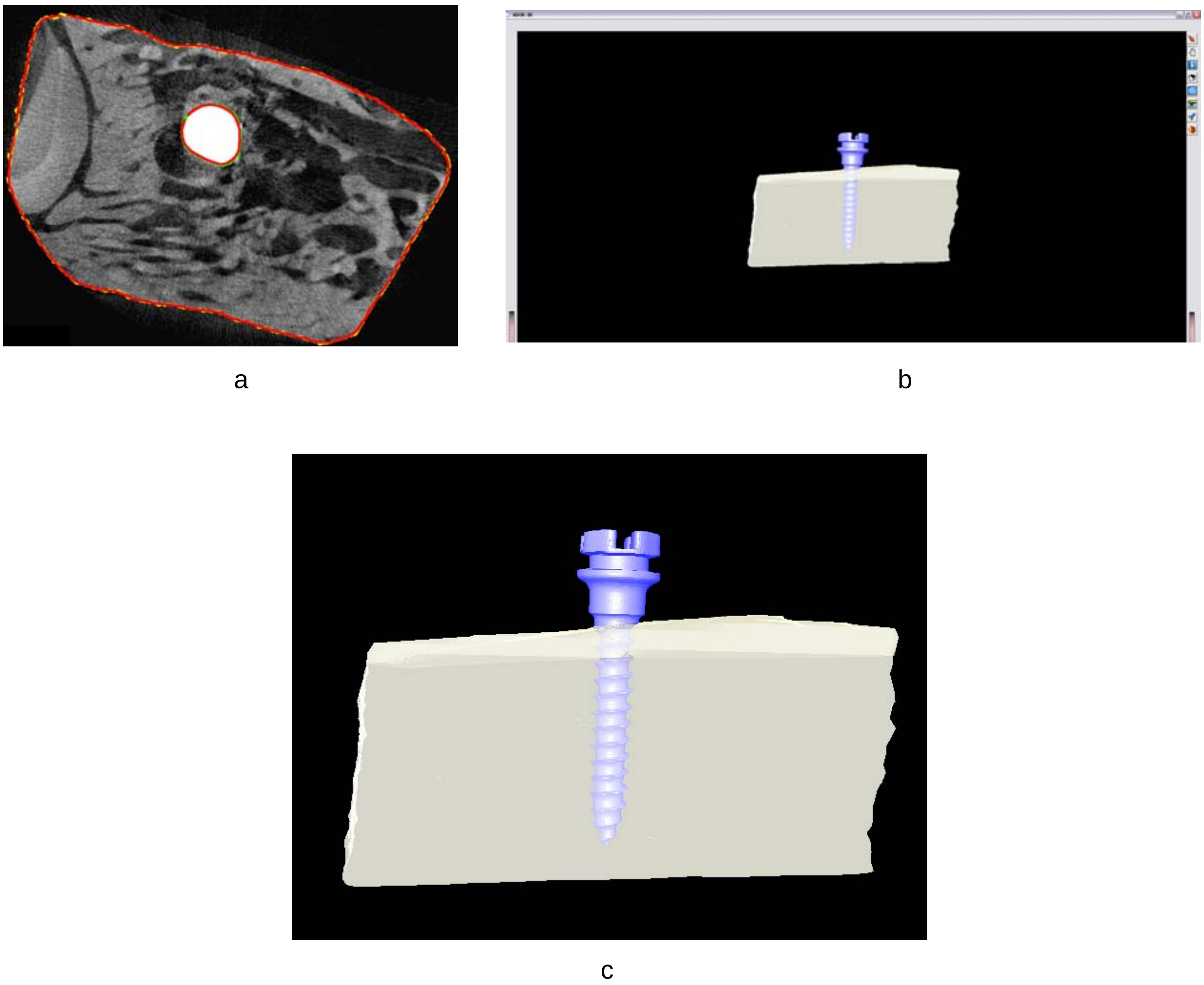


Figure 13: a) Slice view of a specimen, b) 3D surface reconstruction of implant and surrounding bone with ADOR-3D, c) surface model completed.

4.3.1.3 Finite element model generation

Surface models were further processed to generate finite element models (FE-models). For that, the surface models were transferred in the FE package MSC.Marc/Mentat2007r. Tetrahedral elements were used to mesh each model automatically. At the end, five FE mini-implant models (Figure 14) were derived from the scanned preparations.

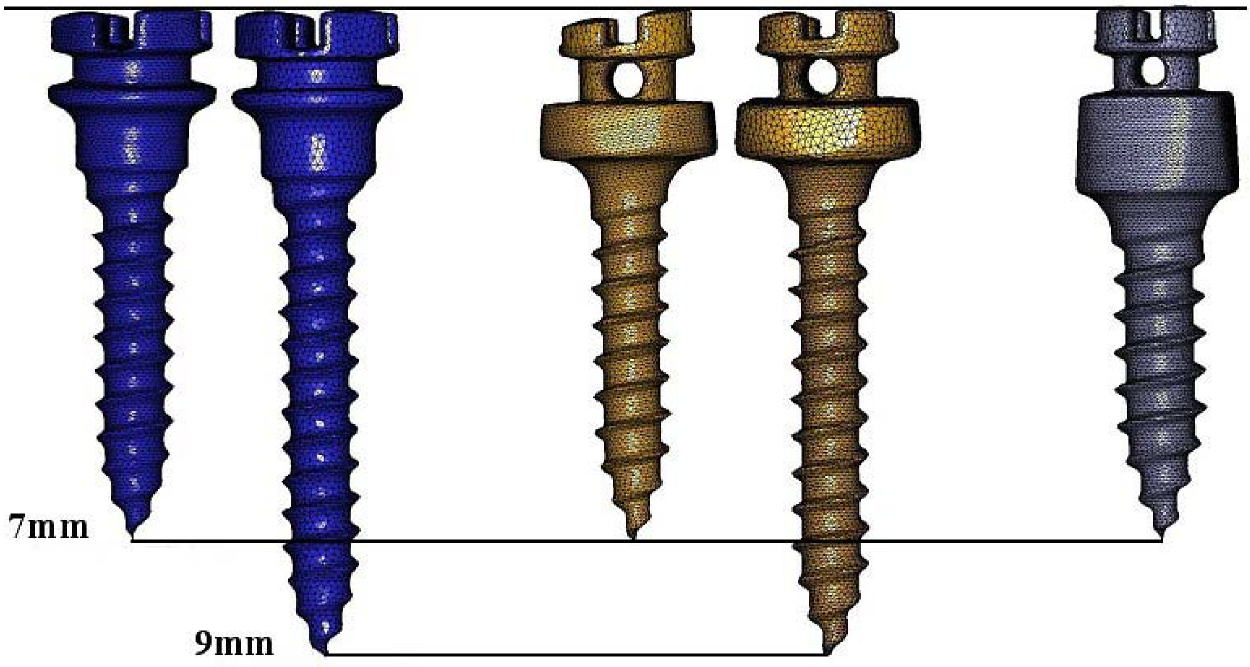


Figure 14: All five types of mini-implants as 3D finite element models. Left to Right: Aarhus 1.5x7 mm, Aarhus 1.5x9 mm, Lomas 1.5x7 mm, Lomas 1.5x9 mm, Lomas 2x7 mm.

After the generation of finite element mini-implant models also the surrounding structures had to be reconstructed. The surrounding structures in our model consisted of the cortical and spongious bone, so the model elements were divided to implant elements, cortical bone elements and spongious bone elements. Critical part in this phase was the connection of the mini-implant elements to the bone elements and the connection between cortical and spongious bone elements. This is the reason why the tetrahedral meshing of the bone elements was made in such a way that the bone elements become smaller and finer toward the implant elements to improve the accuracy of the interfaces and of the numerical calculation. After the tetrahedral meshing, which was done automatically by the computer program, the models were transferred to the FE program 3Matic where a manual element connection in critical areas was done. The number of tetrahedral elements used to mesh each model was ranging from 110.000 to 130.000 and the number of nodes was around 16.000 (Table 4).

In total, nine FE models of mini-implant and surrounding bone were generated (Figure 15-19) representing the nine scanned experimental preparations. Examples of the complete numerical models consisting of the mini-implant and surrounding bone (spongious and cortical) are shown below in figures 15-19.

Implant type	Insertion Angle	Numerical model	Element number		
			Implant	Cortical bone	Spongious Bone
Aarhus 1.5x7 mm	straight	A	43727	14630	51196
	angled	B	39971	41198	33087
Aarhus 1.5x9 mm	straight	C	34335	20840	68637
	angled	D	35111	18445	70481
Lomas 1.5x7 mm	straight	E	43832	27588	40495
	angled	F	42306	27332	27978
Lomas 1.5x9 mm	straight	G	50481	16954	69058
	angled	H	47003	25686	77082
Lomas 1.5x2 mm	straight	I	59128	23144	55215

Table 4: Number of elements of the FE models.

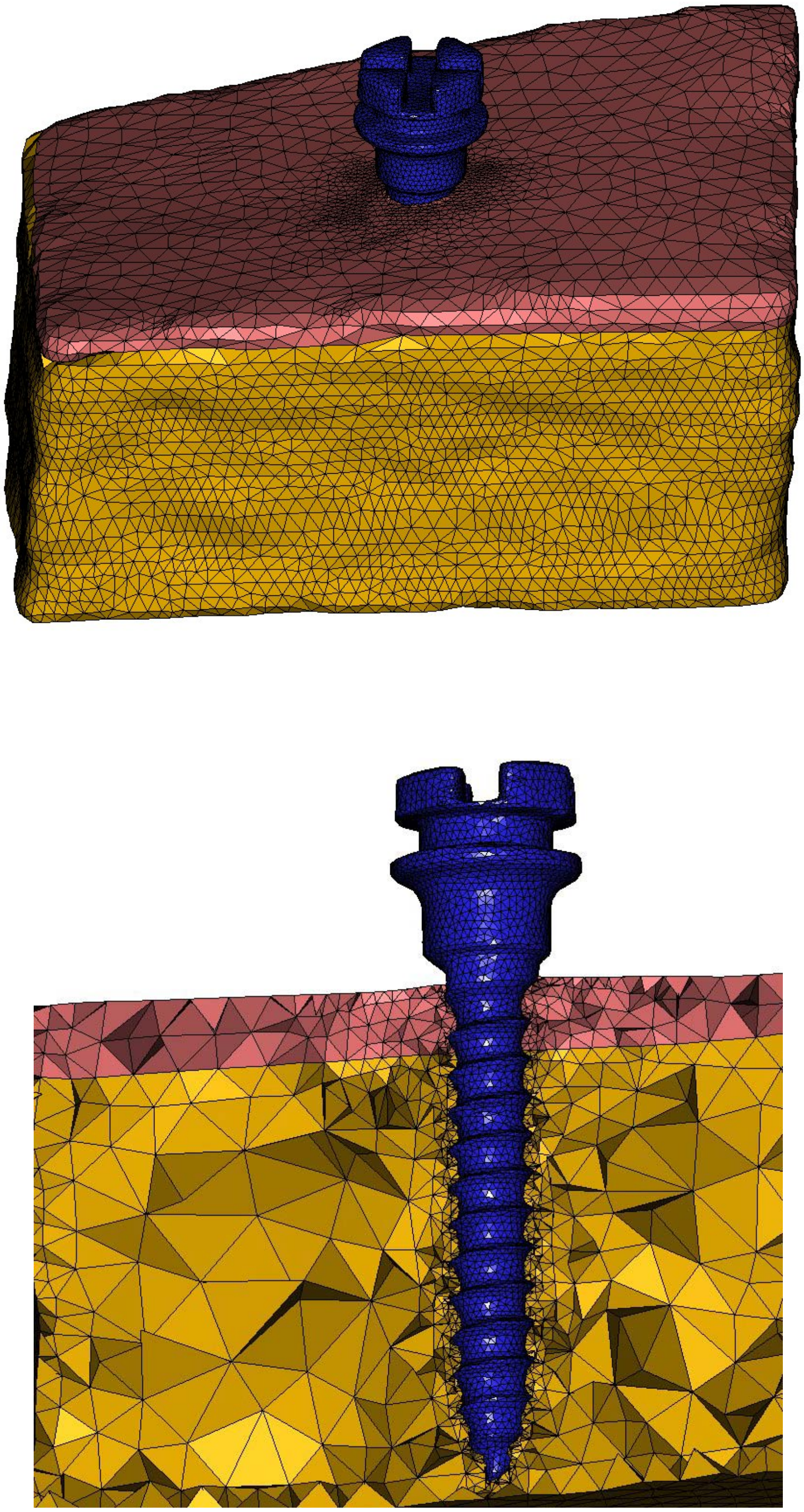
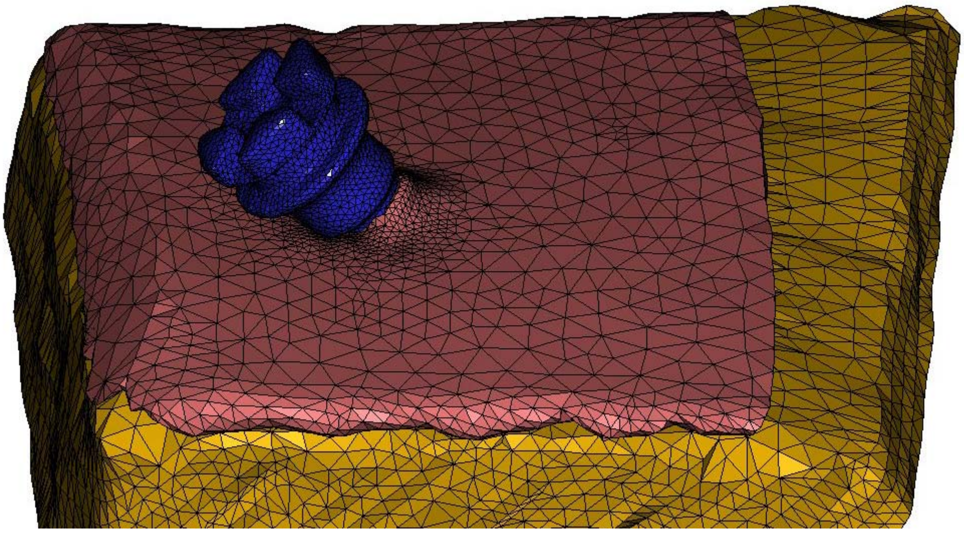


Figure 15: Numerical model 'C' of Aarhus 1.5x9 mm mini-implant, straight insertion. Bone elements become finer close to the mini-implant. Overall view (above) and cut in the plane of the mini-implant (below).



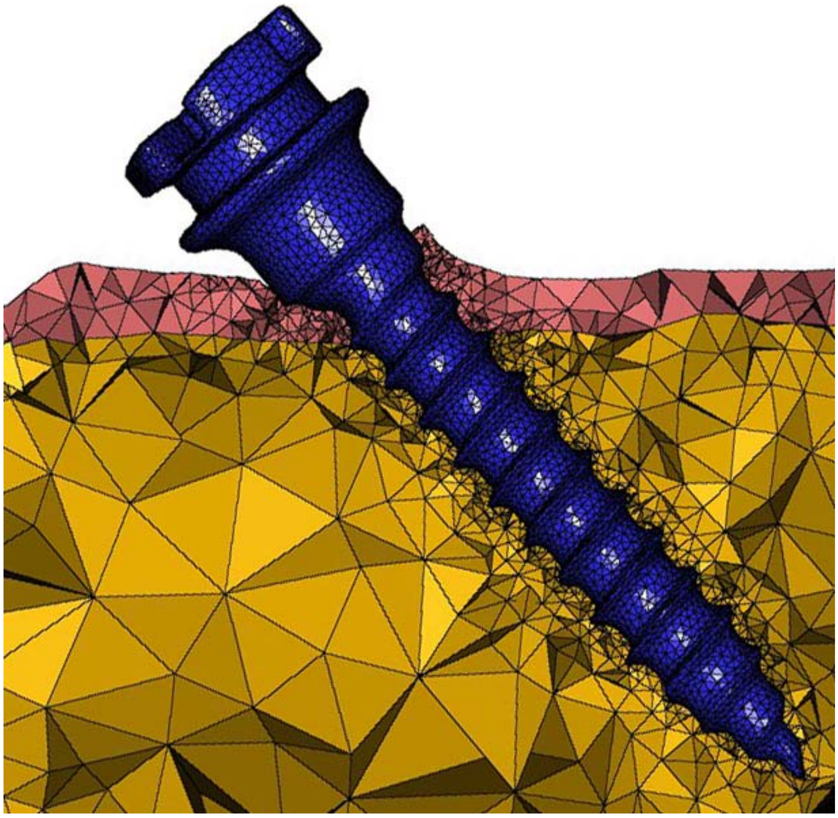


Figure 16: Numerical model 'D' of Aarhus 1.5x9 mm mini-implant, 45° angulation. Overall view (above) and cut in the plane of the mini-implant (below).

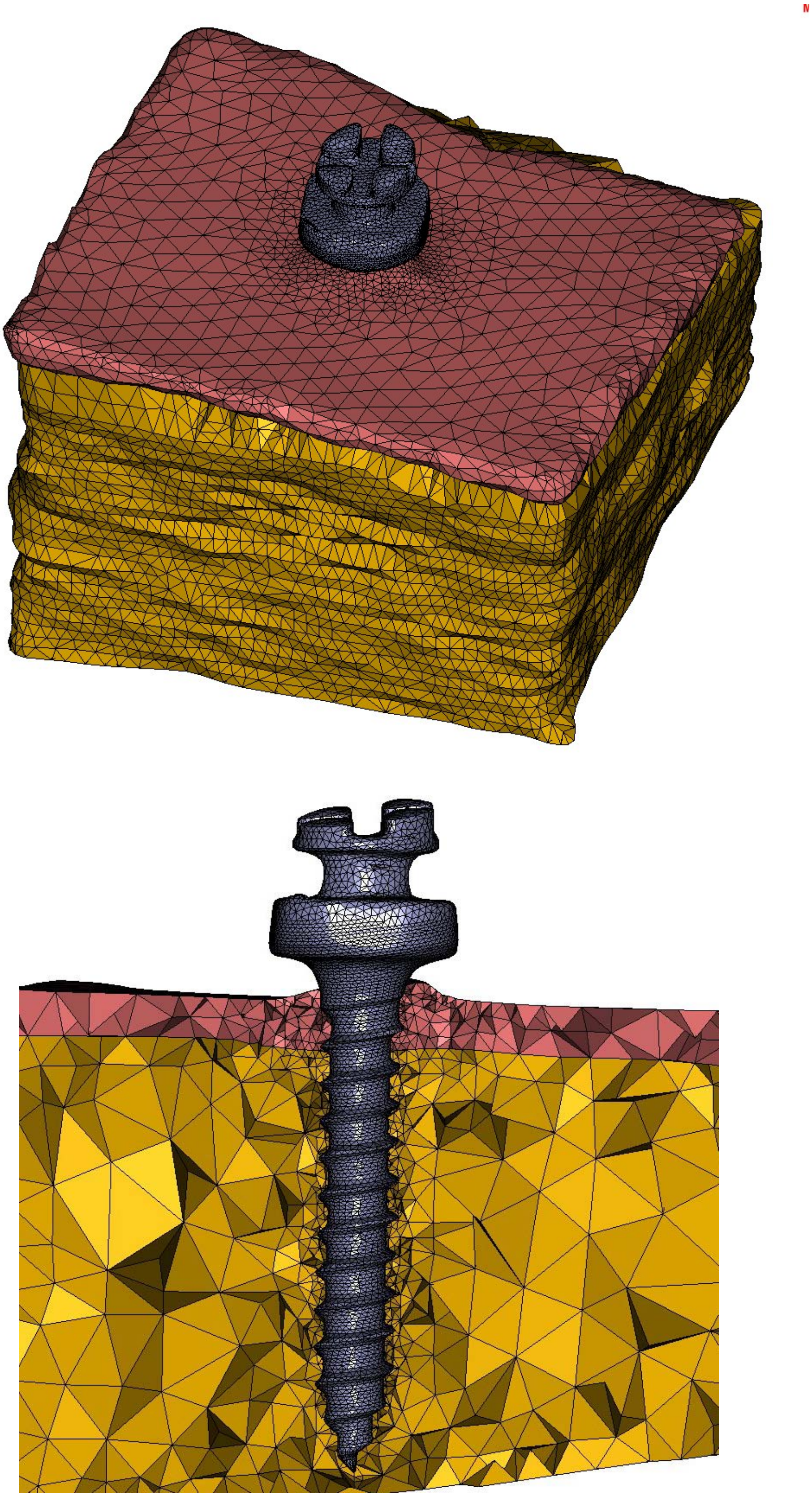


Figure 17: Numerical model 'G' of Lomas 1.5x9 mm mini-implant, straight insertion. Overall view (above) and cut in the plane of the mini-implant (below).

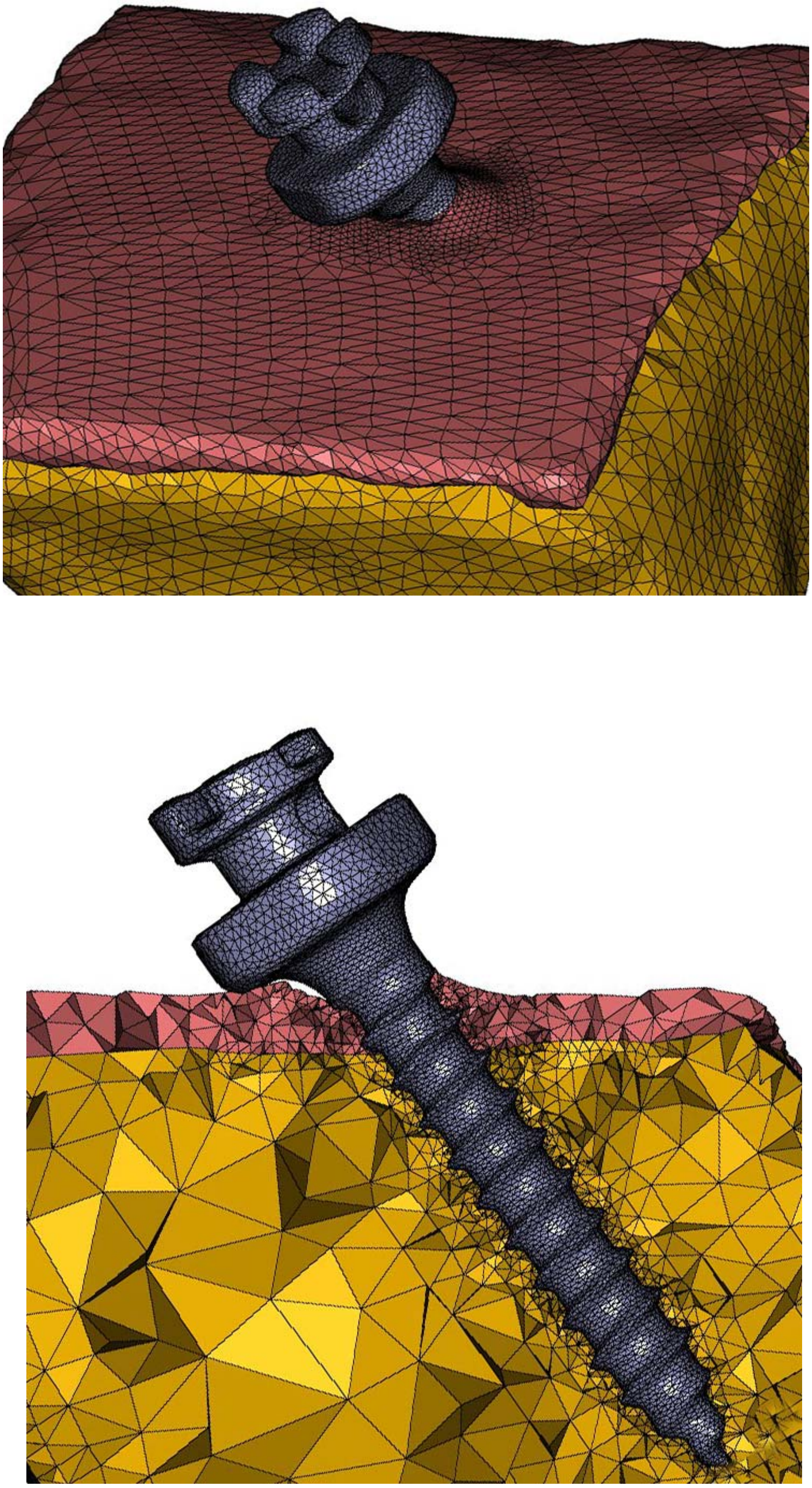


Figure 18: Numerical model 'H' of Lomas 1.5x9 mm mini-implant, 45° angulation. Overall view (above) and cut in the plane of the mini-implant (below).

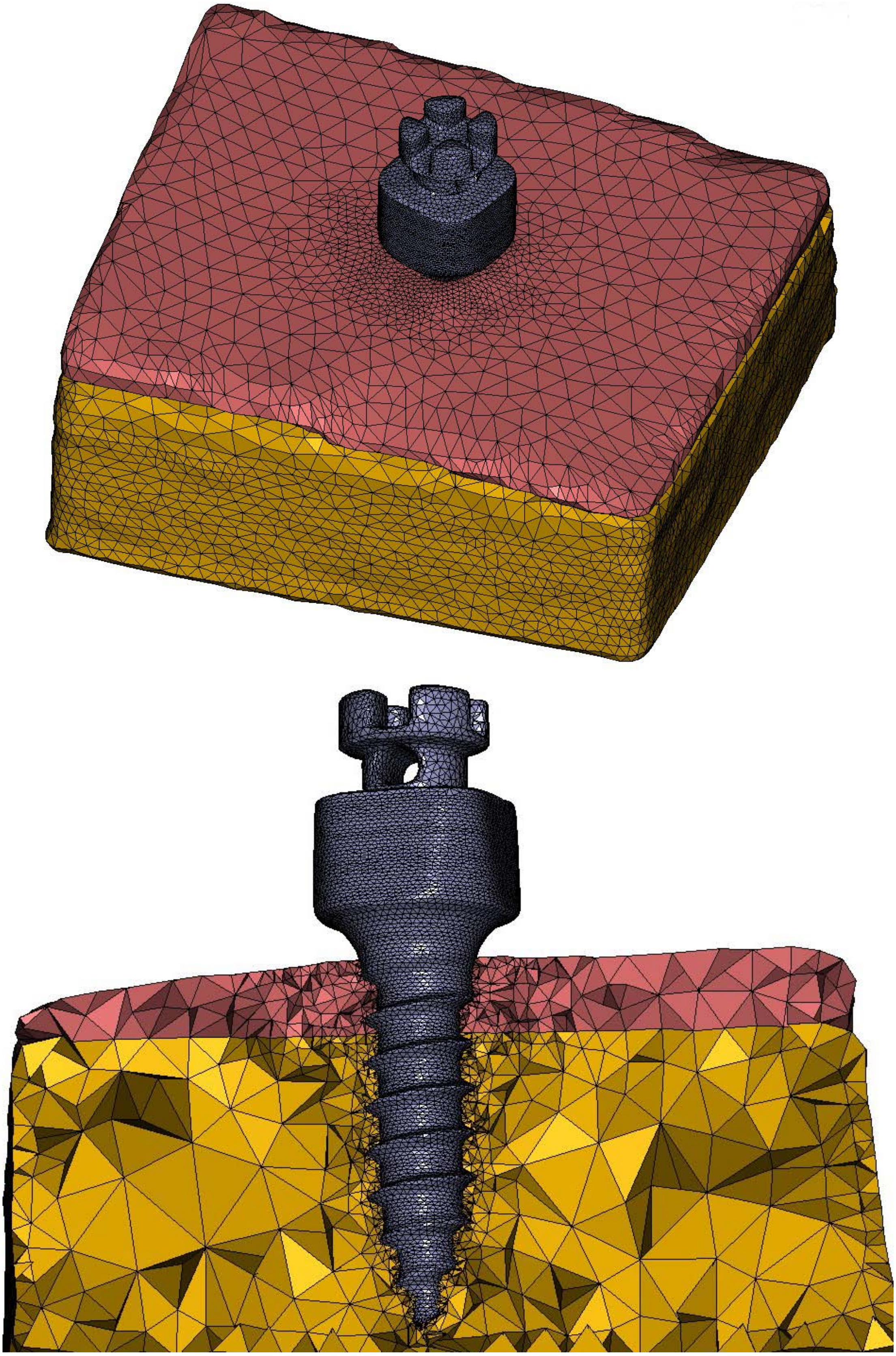


Figure 19: Numerical model 'I' of Lomas 2x9 mm mini-implant, straight insertion. Overall view (above) and cut in the plane of the mini-implant (below).

4.3.2 Concept of combined experimental and numerical studies

Once the 3D finite element model is generated, the loading and boundary conditions and various parameters have to be adjusted in order to resemble as exactly as possible the clinical situation. This involves the direction and amount of force, the material and mechanical parameters, the contact parameters such as frictional coefficient and the contact forces. Example of this concept is presented in figure 20.

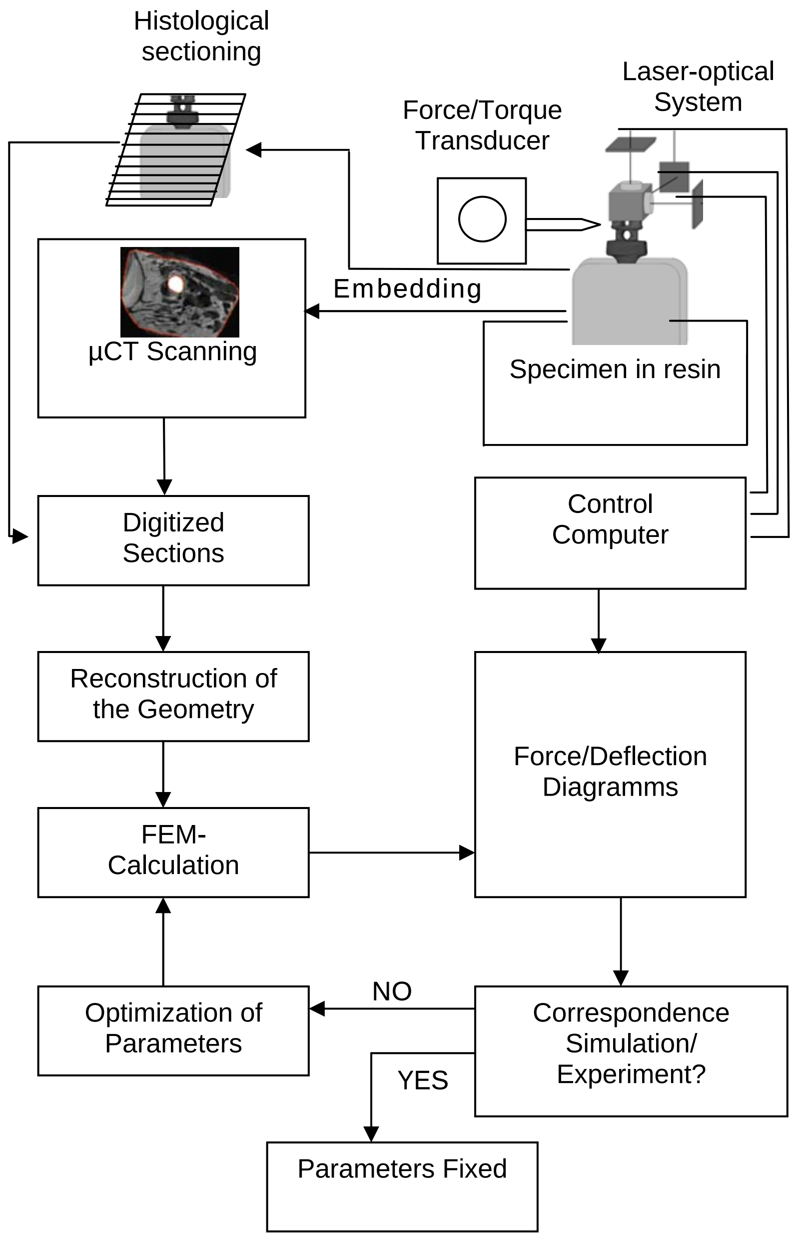


Figure 20: Basic principle of combined experimental and numerical studies. Modified schematic diagram from *Rahimi et al.* [2005].

4.3.3 Numerical simulation in this study

Model loading and boundary conditions were adjusted to perfectly simulate the experimental tests, with the force applied along mesio-distal direction up to a maximum of 0.5N (Figure 21). Since the experimental deflections were derived from the pinhole of the laser cube by the laser beam focusing on the detectors, a same point was constructed for the numerical deflections registration (point A). Both, implant and bone elements (cortical and spongious) were assumed to be homogenous, isotropic and linearly elastic. The mini-implant was assumed to be made of pure titanium. The material properties of each component of the model used in this analysis are summarised in Table 3.

	Young's modulus [GPa]	Poisson's ratio
Mini-implant	110	0.3
Cortical bone	10	0.3
Cancellous bone	0.15	0.3

Table 3: Material properties of FE model components.

In order to further reproduce the experimental condition, the interface between the mini-implants and the bone was not fixed, in a way that friction was considered negligible, as osseointegration was not assumed for the immediately loaded mini-implants. Instead, so-called contact analysis was performed.

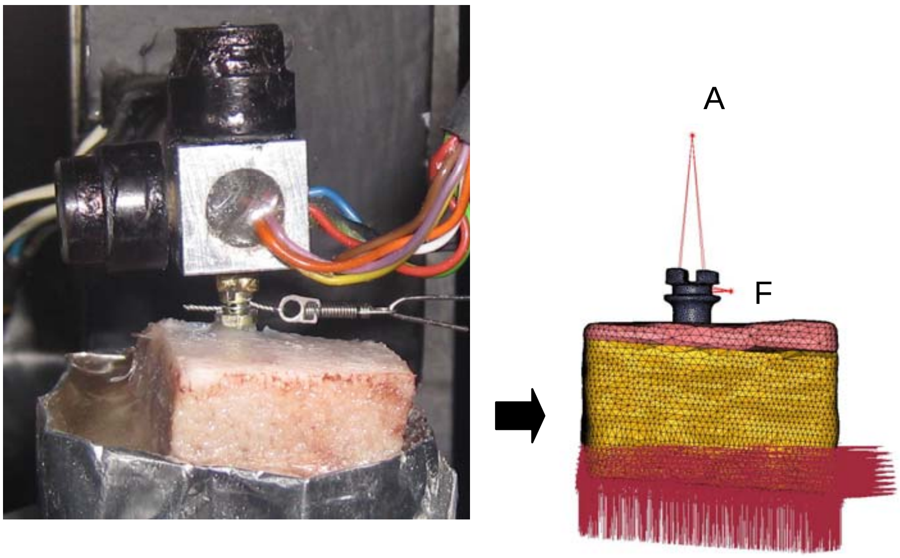


Figure 21: Numerical model of the experimental design in the FE system. The point A in the numerical model on the right coincides with the center of the laser cube in the experimental set up (left). The bone in model is fixed at the basal part to simulate the resin embedding.

4.4 Statistical analysis

Descriptive statistics are expressed as mean ± standard deviation. The experimental error was calculated by testing intra-observer agreement between the first and second measurement of the same preparation using the Altman-Bland test. Difference plots of these two measurements for all implants have also been conducted. To analyze the parameters of implant type, implant length and force level, a Univariate Analysis of Variance (three-way ANOVA) was performed. To examine the effect of implant diameter, only the Lomas mini-implants of the same length (7 mm) and of two different diameters were analyzed, using the independent t-test.

Numerical results were compared to the experimental ones using the Altman-Bland test. Differences were further presented by using difference plots and a graphical technique called Youden plot.

Statistical analysis has been performed with the SPSS v.15 (SPSS Inc., Chicago, Illinois, USA) for ANOVA, with STATA (StataCorp LP, College Station, Texas, USA) software for Altman Bland test and with MedCalc Software (MedCalc Software, Mariakerke, Belgium) for Youden plot. Statistical significance was established at a 0.05 significance level.

5 Results

The results of this study are presented in two parts. In the first part the experimental outcomes are being analyzed. First of all the error of the study by using intra-observer agreement between the first and second measurement of the same preparation is presented. Secondly, the values of mini-implant displacements and rotations are described. At last the statistical results are analyzed. The second part includes the numerical results and their comparison to the experimental ones. Mini-implant displacements along x-axis (Dx) were registered in micrometers (µm) and mini-implant rotations around y-axis (Ry) were measured in degrees (°).

5.1 Experimental results

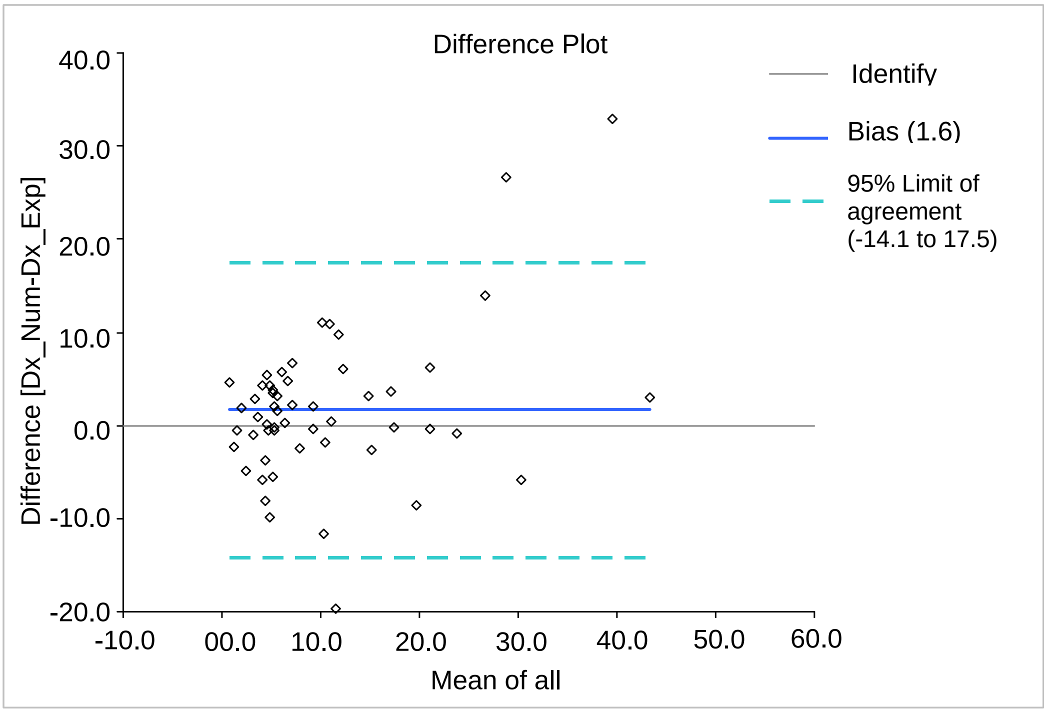
5.1.1 Error of the study

Intra-observer agreement for the whole sample using the Altman-Bland test did not show statistically significant differences in displacement (Dx) and rotation (Ry) values between the first and second measurement of the same preparation (p=NS) (Table 5). The difference plots show that for a 95% level of agreement most values are within limits with only a few showing some deviation (Figure 22). Mean displacement and mean rotation from the two measurements of the same preparation was calculated and this value was used for each mini-implant in further analysis.

Altman-Bland Test	For Dx [µm]	For Ry [°]
n	54	54
95% CI for differences	-5.5 to 3.6	-0.007to 0.014
t statistic	0.43	0.60
DF	53	53
p	0.66 Not significant 95% Limits of agreement	0.54 Not significant 95% Limits of agreement
Lower	-14.1	-0.072
Upper	17.5	0.063

Table 5: Altman-Bland test for displacement (Dx) and rotation (Ry) between first and second measurement of the same implants showed good agreement.

a)



b)

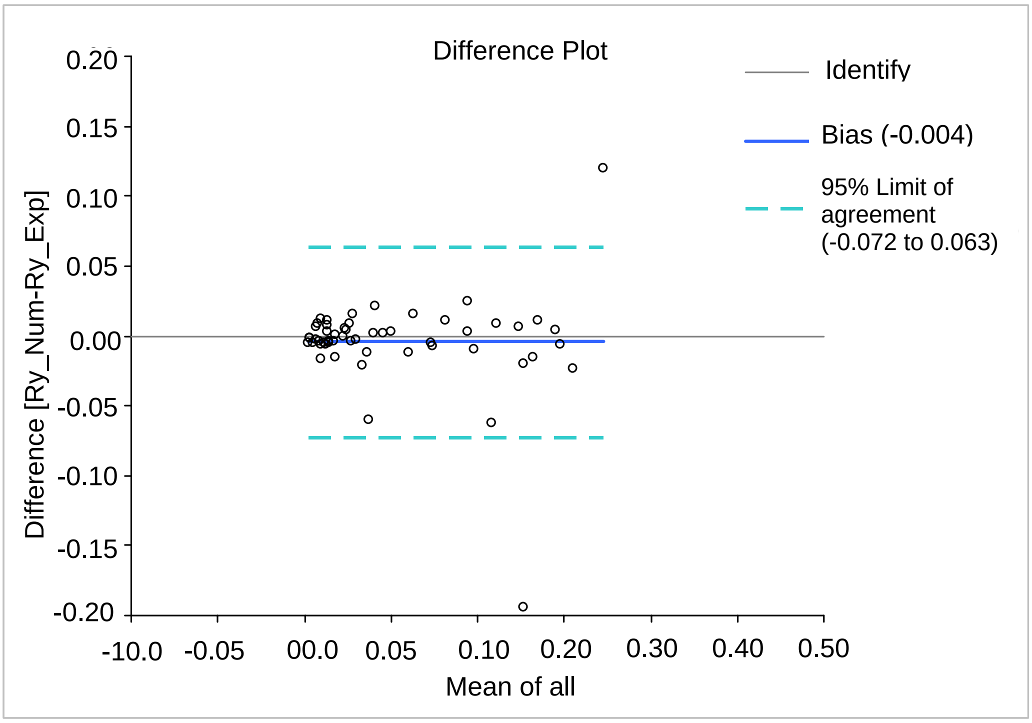


Figure 22: Difference plots show schematically the intraobserver agreement between first and second measurement of the same preparation (p=NS).

5.1.2 Descriptive statistics

Descriptive statistics include the description of the conducted measurements, where lowest and highest values of each group as long as the mean value and standard deviation are being presented. Mini-implant displacements along x-axis (Dx) were registered in micrometers (µm) and mini-implant rotations around y-axis (Ry) were measured in degrees (°).The experimental results are displayed in figures 23-28 as Box-Whisker plots. The following different groups have been arranged for data analysis: a) Individual implant types (Aarhus and LOMAS), differentiated by length and force but insertion mode in common (Figures 23 and 25), b) angle of insertion (straight and 45°angled), differentiated by length and force but both implant types in common (Figures 24 and 26) and c) LOMAS implants of 7 mm in length differentiated by diameter and force (Figures 27 and 28). The descriptive statistics of all implant displacements (µm) and rotations (°) are shown in detail in tables 6 to 9.

5.1.2.1 Small force level group (F=0.5 N)

In the small force group, the mean values of mini-implant displacement (Dx) and rotation (Ry) ranged from 3 to 11 µm (mean 5±2 µm) and from 0.003° to 0.039º (mean 0.016±0.009 º), respectively. The values are presented in Tables 6 and 7.

5.1.2.2. High force level group (F=2.5 N)

When a high force was applied to the implants, mean values of displacement (Dx) and rotation (Ry) were correspondingly higher as expected and ranged from 5 to 43 µm (mean 15±11 µm) and from 0.006º to 0.172º (mean 0.090±0.045º) respectively (Table 8 and 9). By high force application a different biomechanical performance of the groups of mini-implants could be clearly observed and will be discussed in the statistical analysis

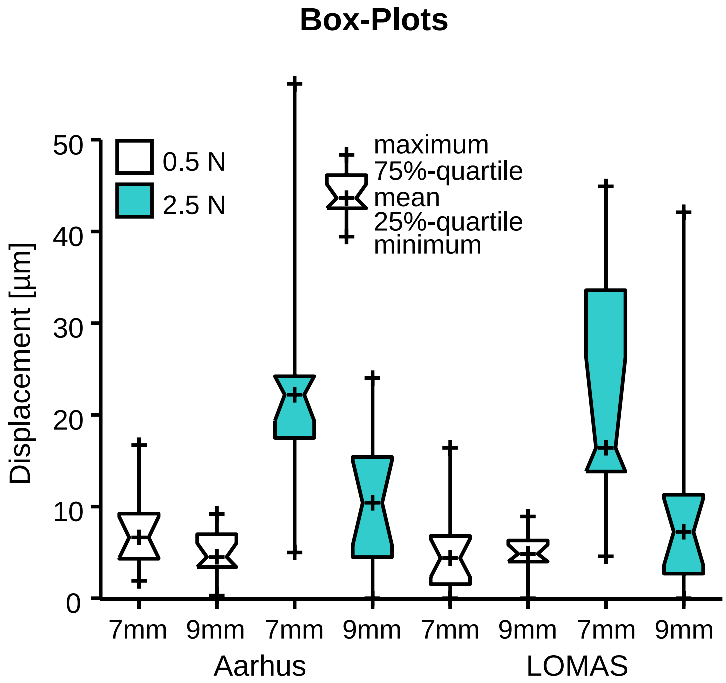


Figure 23: Displacement (Dx) measured for 7 and 9 mm long Aarhus and LOMAS mini-implants, loaded with two force levels. Statistically significant differences were observed in the high force level according to implant length ($p \leq 0.01$). Differences between the two implant types were not significant (p=NS).

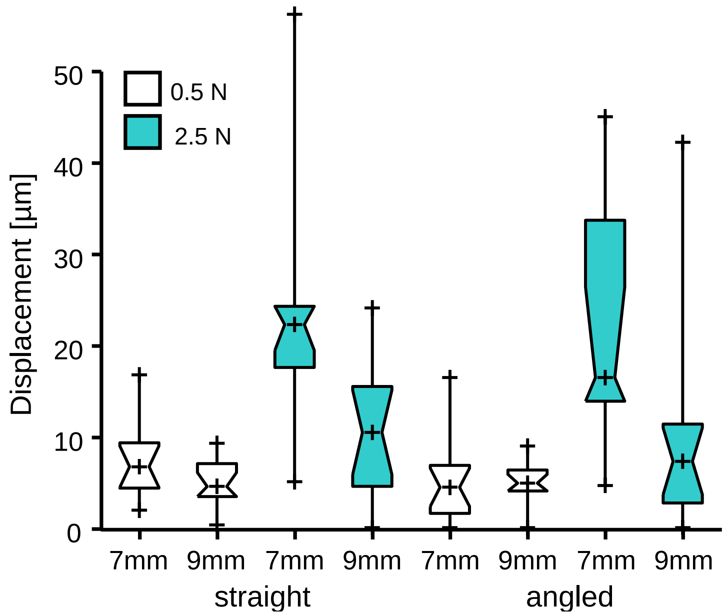


Figure 24: Displacement (Dx) of straight and with 45° of angulation inserted mini-implants loaded with two force levels. No statistically significant differences were observed according to insertion angle (p=NS).

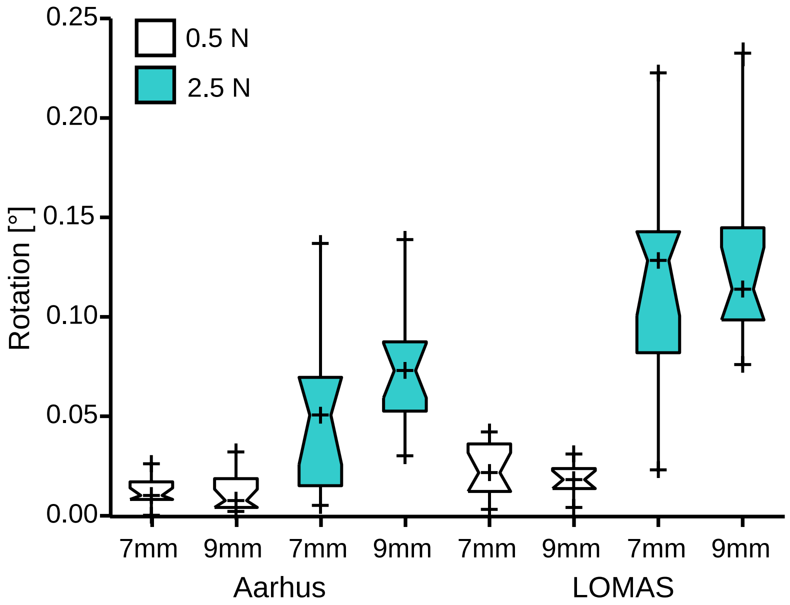


Figure 25: Rotation (Ry) of Aarhus and LOMAS mini-implants loaded with two force levels. Statistically significant differences are observed in both force levels ($p \leq 0.05$).

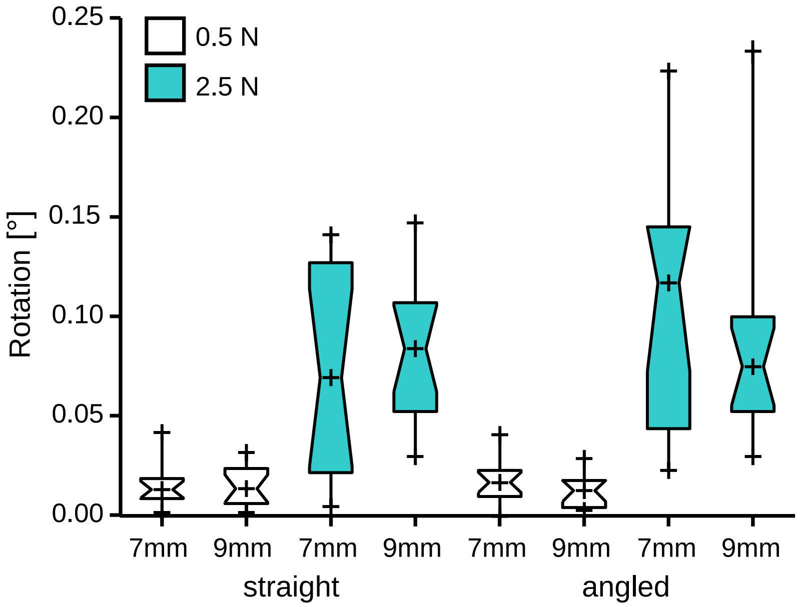


Figure 26: Rotation (Ry) of straight and with 45° inserted mini-implants. No statistically significant differences are observed at any force level (p=NS).

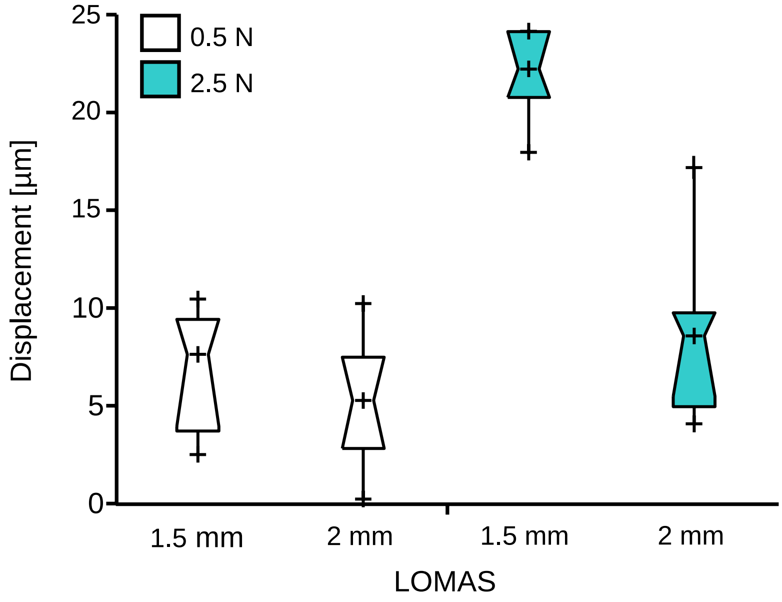


Figure 27: Displacement (Dx) of LOMAS mini-implants of two different diameters (1.5 and 2 mm) loaded with two force levels. Statistically significant differences were observed only in the high force group ($p<0.001$).

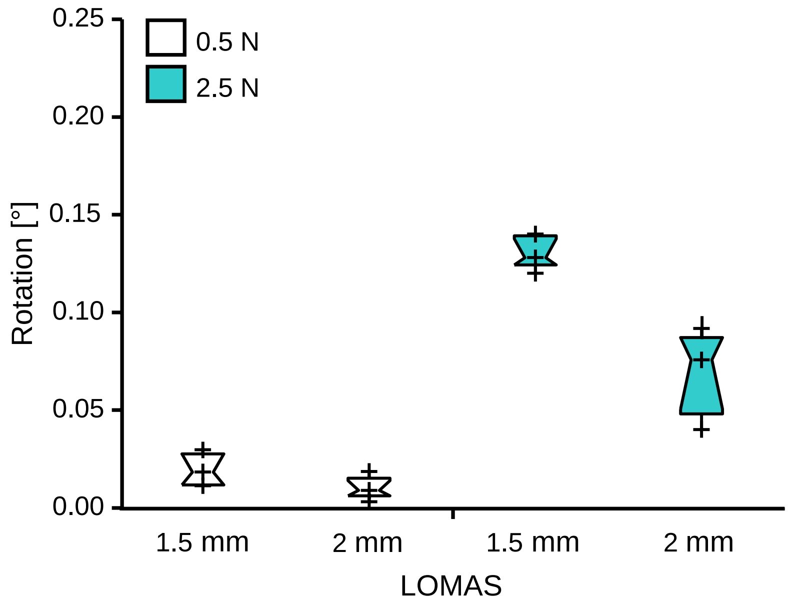


Figure 28: Rotation (Ry) of LOMAS mini-implants of two different diameters (1.5 and 2 mm) loaded with two force levels. Statistically significant differences were again observed only in the high force group ($p<0.05$).

Descriptive Statistics for Dx
F=0.5 N

Dependent Variable: Dx (µm)

Type	*Angle*	*Length*	*Mean*	*SD*	*n*
Aarhus	Straight	7	7.4	4.2	3
		9	5.5	1.7	5
		Total	6.2	2.7	8
	Angled	7	5.2	3.9	4
		9	5.2	0.5	3
		Total	5.2	2.8	7
	Total	7	6.2	3.8	7
		9	5.4	1.3	8
		Total	5.8	2.7	15
LOMAS	Straight	7	6.9	2.4	3
		9	4.6	0.3	2
		Total	6.0	2.1	5
	Angled	7	5.7	3.4	4
		9	5.1	0.6	4
		Total	5.4	2.3	8
	Total	7	6.2	2.8	7
		9	5.0	0.5	6
		Total	5.6	2.1	13
Total	Straight	7	7.1	3.1	6
		9	5.2	1.4	7
		Total	6.1	2.4	13
	Angled	7	5.5	3.4	8
		9	5.2	0.5	7
		Total	5.3	2.4	15
	Total	7	6.2	3.2	14
		9	5.2	1.0	14
		Total	5.7	2.4	28

Table 6: Descriptive statistics for mini-implant displacement (Dx) under force level of 0.5 N.

Descriptive Statistics for Ry
F=0.5 N

Dependent Variable: Ry (°)

Type	*Angle*	*Length*	*Mean*	*SD*	*n*
Aarhus	Straight	7	0.013	0.004	3
		9	0.015	0.011	5
		Total	0.015	0.009	8
	Angled	7	0.011	0.004	4
		9	0.006	0.004	3
		Total	0.009	0.004	7
	Total	7	0.012	0.004	7
		9	0.012	0.010	8
		Total	0.012	0.007	15
LOMAS	Straight	7	0.019	0.015	3
		9	0.020	0.009	2
		Total	0.019	0.012	5
	Angled	7	0.027	0.010	4
		9	0.017	0.008	4
		Total	0.022	0.010	8
	Total	7	0.023	0.012	7
		9	0.018	0.008	6
		Total	0.021	0.010	13
Total	Straight	7	0.016	0.010	6
		9	0.017	0.010	7
		Total	0.016	0.010	13
	Angled	7	0.019	0.011	8
		9	0.012	0.009	7
		Total	0.016	0.010	15
	Total	7	0.018	0.010	14
		9	0.015	0.009	14
		Total	0.016	0.010	28

Table 7: Descriptive statistics for mini-implant rotation (Ry) under force level of 0.5 N.

Descriptive Statistics for Dx
F=2.5 N

Dependent Variable: Dx (µm)

Type	*Angle*	*Length*	*Mean*	*SD*	*n*
Aarhus	Straight	7	23.1	15.0	4
		9	6.6	4.6	4
		Total	14.9	13.5	8
	Angled	7	26.7	23.5	2
		9	13.1	10.9	4
		Total	17.6	15.2	6
	Total	7	24.3	15.8	6
		9	9.9	8.5	8
		Total	16.1	13.7	14
LOMAS	Straight	7	21.9	1.5	3
		9	16.3	3.7	3
		Total	19.1	4.0	6
	Angled	7	18.8	6.7	3
		9	6.2	4.4	3
		Total	12.5	8.6	6
	Total	7	20.4	4.7	6
		9	11.2	6.6	6
		Total	15.8	7.2	12
Total	Straight	7	22.6	10.6	7
		9	10.8	6.4	7
		Total	16.7	10.4	14
	Angled	7	22.0	13.4	5
		9	10.1	8.9	7
		Total	15.1	12.1	12
	Total	7	22.3	11.3	12
		9	10.5	7.5	14
		Total	15.9	11.0	26

Table 8: Descriptive statistics for mini-implant displacement (Dx) under force level of 2.5 N.

Descriptive Statistics for Ry
F=2.5 N

Dependent Variable: Ry (°)

Type	*Angle*	*Length*	*Mean*	*SD*	*n*
Aarhus	Straight	7	0.033	0.024	4
		9	0.055	0.015	4
		Total	0.044	0.022	8
	Angled	7	0.089	0.054	2
		9	0.093	0.014	4
		Total	0.091	0.027	6
	Total	7	0.051	0.042	6
		9	0.074	0.024	8
		Total	0.064	0.034	14
LOMAS	Straight	7	0.130	0.006	3
		9	0.118	0.024	3
		Total	0.124	0.017	6
	Angled	7	0.103	0.061	3
		9	0.136	0.048	3
		Total	0.120	0.052	6
	Total	7	0.116	0.041	6
		9	0.127	0.036	6
		Total	0.122	0.037	12
Total	Straight	7	0.074	0.055	7
		9	0.082	0.038	7
		Total	0.078	0.046	14
	Angled	7	0.097	0.051	5
		9	0.111	0.038	7
		Total	0.105	0.042	12
	Total	7	0.084	0.052	12
		9	0.097	0.039	14
		Total	0.091	0.045	26

Table 9: Descriptive statistics for mini-implant rotation (Ry) under force level of 2.5 N.

Variables: Dx (µm), Ry (º)

Diameter (mm)	Force (N)	Variables	Mean	SD	n
1.5	0.5	Dx	6.9	2.4	3
		Ry	0.012	0.007	3
	2.5	Dx	21.9	1.5	3
		Ry	0.130	0.006	3
2.0	0.5	Dx	5.6	3.5	5
		Ry	0.008	0.008	5
	2.5	Dx	8.8	2.3	3
		Ry	0.070	0.023	3

Dx, mini-implant displacement (µm) along x-axis; Ry, mini-implant rotation (º) around y-axis.

Table 10: Variables Dx and Ry: Descriptive statistics of two implant diameters under two force levels.

5.1.3 Statistical analysis

5.1.3.1 Small force level group (F=0.5N)

When a small force level was applied to the mini-implant the analysis of Variance (ANOVA) did not show statistically significant differences in displacement (Dx) according to implant type, implant length, implant diameter and insertion mode (p=NS, Table 11). In contrast, the rotation (Ry) values showed statistically significant differences between the two implant types, with LOMAS mini-implants tending to rotate significantly more (mean 0.020±0.010º) than Aarhus mini-implants (mean 0.011±0.007º, $p \leq 0.05$, Table 12).

5.1.3.2 High force level group (F=2.5N)

By high force application on mini-implants, the ANOVA test showed a statistically significant difference in displacement (Dx) according to implant length. The 9 mm mini-implants were displaced significantly less (mean 11 ± 8 µm) than the 7 mm mini-implants (mean 22 ± 11 µm, $p \leq 0.01$, Table 13) thus indicating a better primary stability. Apart from implant length, implant diameter was also an influencing factor for implant stability, since the 2 mm wide mini-implants were also displaced and rotated significantly less (mean 9 ± 2 µm and 0.07±0.02°) than the 1.5 mm wide implants (mean 22±2 µm and 0.130±0.005°, $p \leq 0.05$; Table 15). The Ry values showed again that the LOMAS mini-implants rotated significantly more (mean 0.121±0.037º) than the Aarhus mini-implants (mean 0.064±0.033°, $p \leq 0.001$, Table 14). The angle of implant insertion was not found to be an influencing parameter in any force level in this study (p=NS).

F=0.5N

Dependent Variable: Dx

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Corrected Model	1.82E-005[a]	7	2.60E-006	0.364	0.912
Intercept	0.001	1	0.001	121.160	0.000
Type	4.35E-007	1	4.35E-007	0.061	0.808
Angle	3.87E-006	1	3.87E-006	0.542	0.470
Length	9.16E-006	1	9.16E-006	1.282	0.271
Type * Angle	1.29E-006	1	1.29E-006	0.181	0.675
Type * Length	3.84E-007	1	3.84E-007	0.054	0.819
Angle*Length	5.29E-006	1	5.29E-006	0.740	0.400
Type * Angle * Length	1.60E-008	1	1.60E-008	0.002	0.963
Error	0.000	20	7.15E-006		
Total	0.001	28			
Corrected Total	0.000	27			

a Adjusted R Squared = 0.197
p=NS

Table 11: Mini-implant displacement (Dx) versus implant type, implant length and insertion angle in F=0.5N. (three-way ANOVA). No statistically significant differences were observed.

Dependent Variable: Ry

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Corrected Model	0.001[a]	7	0.000	1.715	0.162
Intercept	0.007	1	0.007	81.570	0.000
Type	0.001	1	0.001	7.210	0.014*
Angle	2.33E-005	1	2.33E-005	0.284	0.600
Length	5.17E-005	1	5.17E-005	0.629	0.437
Type * Angle	0.000	1	0.000	1.390	0.252
Type * Length	1.02E-005	1	1.02E-005	0.124	0.728
Angle*Length	0.000	1	0.000	1.742	0.202
Type * Angle * Length	4.45E-006	1	4.45E-006	0.054	0.818
Error	0.002	20	8.21E-005		
Total	0.010	28			
Corrected Total	0.003	27			

a Adjusted R Squared = 0.156
* $p \leq 0.05$

Table 12: Mini-implant rotation (Ry) versus implant type, implant length and insertion angle in F=0.5N (three-way ANOVA). Statistically significant differences were observed according to implant type.

F=2.5N

Dependent Variable: Dx

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Corrected Model	0.001[a]	7	0.000	1.740	0.162
Intercept	0.007	1	0.007	67.868	0.000
Type	1.55E-005	1	1.55E-005	0.154	0.700
Angle	3.77E-006	1	3.77E-006	0.037	0.849
Length	0.001	1	0.001	8.917	0.008**
Type * Angle	0.000	1	0.000	2.085	0.166
Type * Length	5.41E-005	1	5.41E-005	0.535	0.474
Angle*Length	6.71E-006	1	6.71E-006	0.066	0.800
Type * Angle * Length	3.77E-005	1	3.77E-005	0.373	0.549
Error	0.002	18	0.000		
Total	0.010	26			
Corrected Total	0.003	25			

a Adjusted R Squared = 0.172

**p≤0.01

Table 13: Mini-implant displacement (Dx) versus implant type, implant length and insertion angle in F=2.5N (three-way ANOVA). Statistically significant differences were observed according to implant length (p≤0.01).

Dependent Variable: Ry

Source	Type III Sum of Squares	df	Mean Square	F	Sig.
Corrected Model	0.032[a]	7	0.005	4.355	0.006
Intercept	0.221	1	0.221	208.201	0.000
Type	0.019	1	0.019	17.463	0.001***
Angle	0.003	1	0.003	2.597	0.124
Length	0.001	1	0.001	0.839	0.372
Type * Angle	0.004	1	0.004	3.882	0.064
Type * Length	8.41E-006	1	8.41E-006	0.008	0.930
Angle*Length	0.000	1	0.000	0.285	0.600
Type * Angle * Length	0.002	1	0.002	1.522	0.233
Error	0.019	18	0.001		
Total	0.266	26			
Corrected Total	0.052	25			

a Adjusted R Squared= 0.484

***p≤0.001

Table 14: Mini-implant rotation (Ry) versus implant type, implant length and insertion angle in F=2.5N (three-way ANOVA). Statistically significant differences were observed according to implant type (p≤0.001).

Force (N)	Variables (*Equal variances assumed*)	t-test for Equality of means		
		t	df	Sig. (2-tailed)
0.5	Dx	0.558	6	0.597
	Ry	2.102	6	0.080
2.5	Dx	8.035	4	0.001 **
	Ry	4.341	4	0.012 *

* *P*<0.05**,** ** *P*<0.001

Table 15: T-test for the two implant diameters concerning displacement (Dx) and rotation (Ry) versus force level. Significant differences were observed according to implant diameter only in the high force group.

5.1.3.3 Optimum force level

As shown above, the application of two different force levels differentiated some of the out coming results. Since the biomechanical performance of mini-implants was clearly defined when a high force level was applied, the question had risen, which would be the optimum force level, above which the implant length and implant diameter could be influencing parameters for implant primary stability. The incremental steps from 0 to 2.5 N were divided into five 0.5 N intervals and the already previously recorded data were further analyzed. Mean values of displacement (Dx) and rotation (Ry) of mini-implants were calculated again for each force group separately. The same statistical analysis of variance was performed for each group. The results showed that implant length and implant diameter can be statistically significant influencing factors ($p \leq 0.01$) on implant stability when the force level is of 1 N or higher (Tables 16 and 17).

Force	Type			Length (mm)			Angle		
	Dx (µm) (mean ± stdev)								
	AARHUS	LOMAS	p	7mm	9mm	p	Straight	Angled	p
0.71-1.00	5,5 ± 3,4	6,4 ± 2,7	NS	7,3 ± 3,7	4,7 ± 1,7	0,052	5,8 ± 3,1	6,1 ± 3,1	NS
1.01-1.50	7,1 ± 4,5	8,7 ± 3,5	NS	10,3 ± 4,3	5,8 ± 2,4	0,004 < 0,01	7,6 ± 3,4	8,2 ± 4,9	NS
1.51-2.00	9,4 ± 6,4	10,9 ± 3,43	NS	12,9 ± 5,6	7,6 ± 3,4	0,008 < 0,01	9,9 ± 4,1	10,2 ± 6,4	NS
2.01-2.50	12,3 ± 8,6	12,8 ± 4,9	0 NS	16,1 ± 7,8	9,5 ± 4,6	0,013 < 0,05	12,6 ± 5,4	12,5 ± 8,8	NS

Table 16: Mini-implant displacement (Dx) according to force groups. Statistical differentiation for implant length starts at the force level of 1 N.

	Ry (°) (mean ± stdev) *10^{-2}								
Force	Type			Length (mm)			Angle		
	AARHUS	LOMAS	p	7mm	9mm	p	Straight	Angled	p
0.71-1.00	2.34 ± 0.91	3.84 ± 1.27	0.005 < 0.01	2.67 ± 1.16	3.46 ± 1.40	NS	2.89 ± 1.37	3.16 ± 1.30	NS
1.01-1.50	3.16 ± 1.41	5.67 ± 1.89	0.002 < 0.01	4.15 ±2.39	4.47 ± 1.81	NS	3.71 ± 1.87	5.04 ± 2.12	NS
1.51-2.00	4.24 ± 2.08	7.83 ± 2.66	0.002 < 0.01	5.81 ±3.66	5.97 ± 2.32	NS	5.02 ± 2.75	6.92 ± 2.96	NS
2.01-2.50	5.40 ± 2.61	10.32 ± 3.64	0.002 < 0.01	7.69 ±5.06	7.57 ± 2.81	NS	6.55 ± 3.59	8.89 ± 4.06	NS

Table 17: Mini-implant rotation (Ry) according to force groups. Statistical differentiation for implant type is evident at all force levels

5.2 Numerical results

5.2.1 Description

One theoretical model out of the nine reconstructed did not work properly in the simulation due to unknown reasons and was excluded from further analysis (model H). On the eight remaining models, the numerical values of displacement and rotation, registered on the 0.5 N level, were compared to the corresponding mean experimental values. The deflection values in experiment and in the simulation are shown in table 18 and figure 29.

Models		Dx_Exp [µm]	Dx_Num [µm]	Ry_Exp [°]	Ry_Num [°]
Aarhus	A	4.9	4.3	0.011	0.019
	B	6.1	3.0	0.011	0.009
	C	5.1	5.4	0.030	0.023
	D	5.5	5.7	0.036	0.042
LOMAS	E	6.3	5.2	0.014	0.023
	F	4.2	5.2	0.017	0.025
	G	6.3	5.9	0.012	0.024
	I	4.6	4.0	0.010	0.017

Table 18: Comparison between experimental and numerical displacement (Dx) and rotation (Ry) values of the small force level group (p=NS).

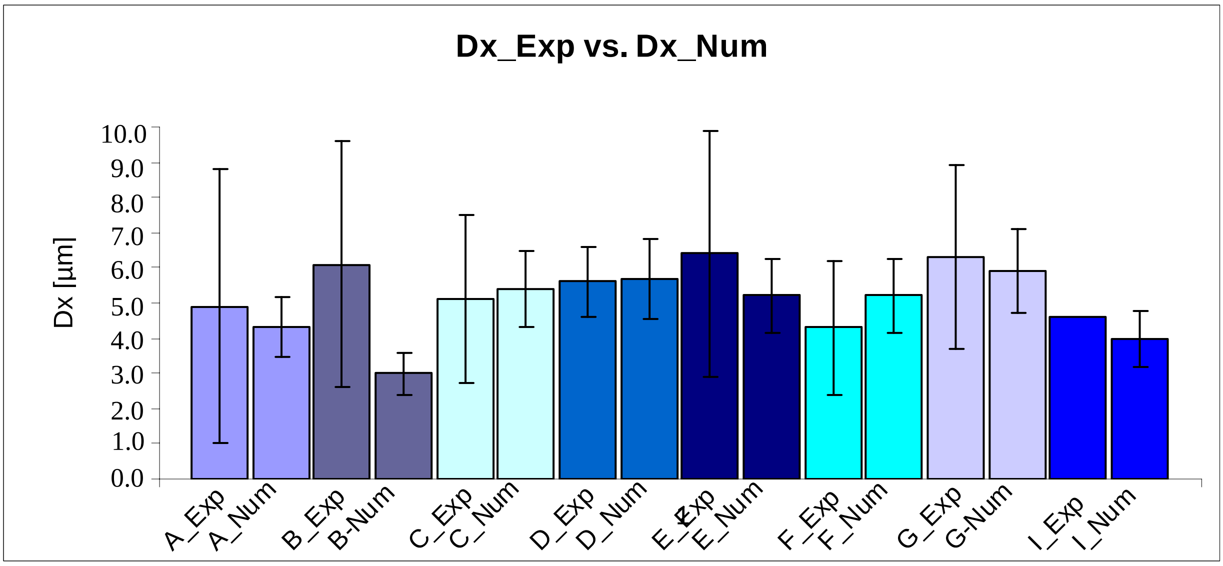


Figure 29: Diagram of experimental and numerical values of the models.

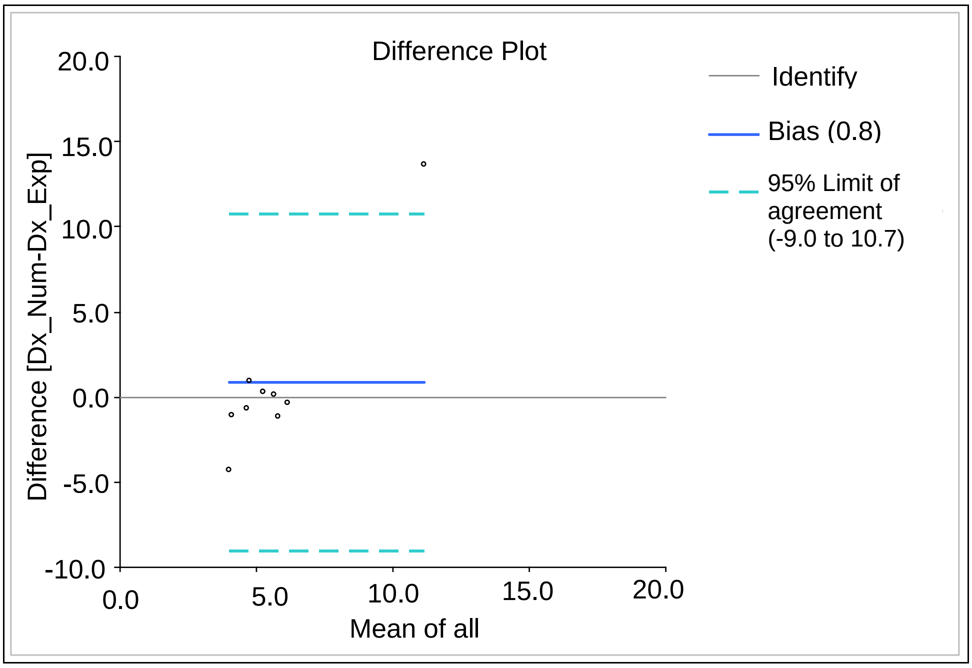
5.2.2 Comparison between experimental and numerical results

Intra-observer agreement between experimental and numerical values did not show statistically significant differences (p=NS, Table 19). A schematic representation is shown with difference plots (Figure 30).

Altman Bland Test	For Dx [µm]	For Ry [°]
n	9	9
95% CI for differences	-3.0 to 4.7	-0.003 to 0.034
t statistic	0.50	1.97
DF	8	8
p	0.63 Not significant	0.08 Marginally Not significant
	95% Limits of agreement	95% Limits of agreement
Lower	-9.0	-0.031
Upper	10.7	0.063

Table 19: Altman Bland tests between experimental and numerical measurements for displacement (Dx) and rotation (Ry) showed good agreement.

a)



b)

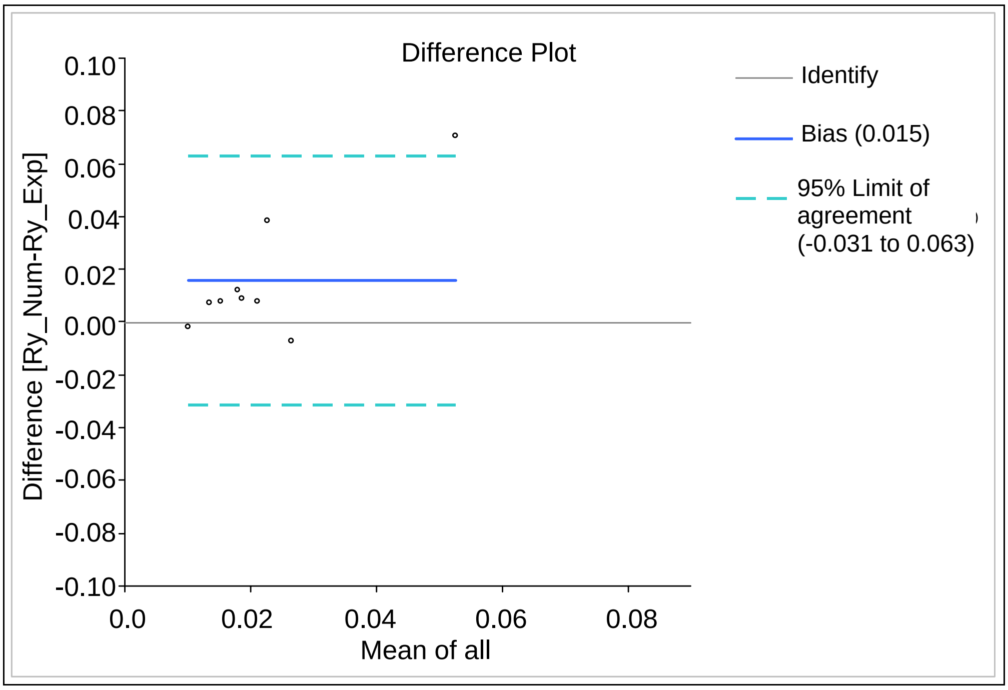


Figure 30: Difference plots for displacement Dx (a) and rotation Ry (b) between experimental and numerical values.

A more detailed analysis using the Youden plot showed a quite good coincidence between experimental and numerical values, with only two values showing slight deviation (Figure 31). For that, a horizontal median line was drawn parallel to the x-axis so that there should be as many points above the line as there were below it. A second median line was drawn parallel to the y-axis so that there should be as many points on the left as there were on the right of this line. Outliers are not used in determining the position of the median lines. The intersection of the two median lines is called the Manhattan median. The circle drawn should include 95 % of the experimental observations, if the individual constant errors could be eliminated. A 45-degree reference line was drawn through the Manhattan median so that observations far from the line and outside of the circle indicated a systematic and random error.

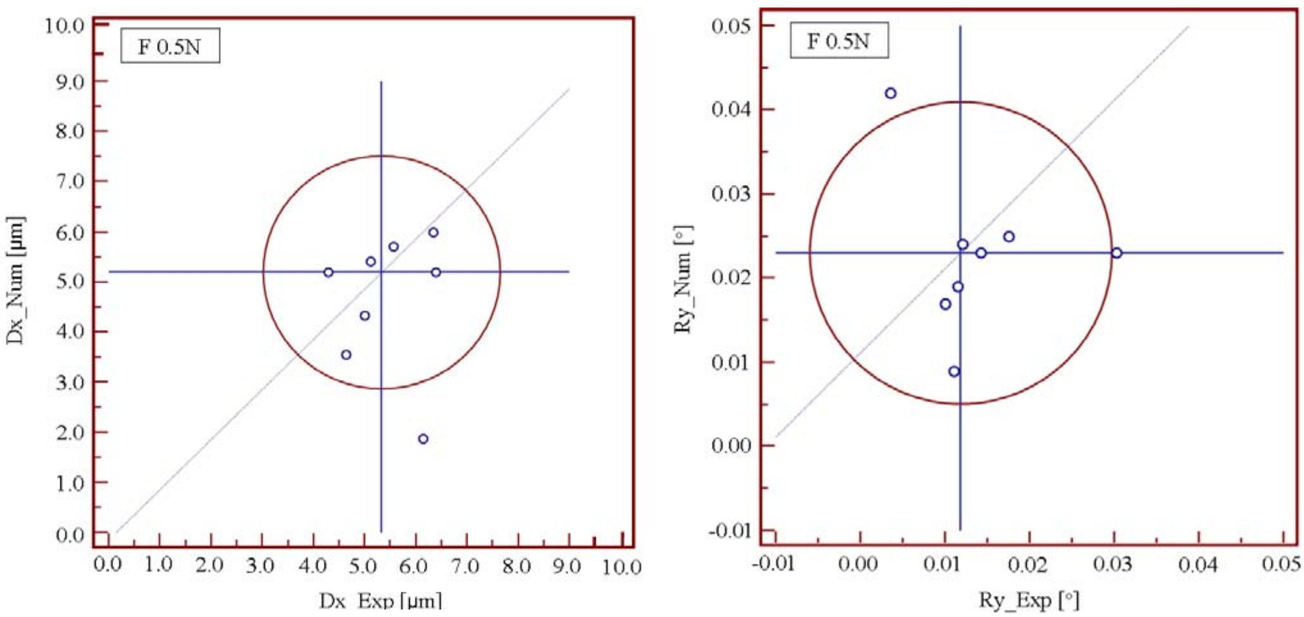


Figure 31: Youden plot: Comparison between experimental and numerical values. a) Displacement Dx_Exp to Dx_Num, b) Rotation Rx_Exp to Rx_Num.

6 Discussion

6.1 Discussion in the light of literature

The review of the literature has revealed different studies, underlining correlations between various parameters and mini-implant stability. In this study conical mini-implant types have been selected instead of cylindrical ones due to their better primary stability [Wilmes et al., 2006; Wilmes et al., 2008]. Insertion of mini-implants with the self drilling method, as proposed by the manufacturers, was selected to exclude a pilot drilling hole, since pilot hole size and depth has been found to influence primary stability of mini-implants [Gantous et al., 1995; Wilmes et al., 2006]. Cortical bone thickness (CBT) and bone quality were not influencing parameters in this study, since they were almost equal in all preparations (CBT around 2 mm). According to *Motoyoshi et al*. [2008] cortical bone thickness at implant site should be greater than 1.0 mm to improve success rates of mini-implants.

In this study, measuring the implants primary stability according to different influencing parameters was the main purpose. As mentioned before, primary stability is the implants stability immediately after insertion. Until know, primary stability was only measured by recording the value of maximum insertion torque (IS). In this study a different method was used.

Our results showed that implant length and implant diameter have a great impact on mini-implant primary stability when the force level was of 1 N or higher. This interaction of force level, implant dimensions and implants displacements was for the first time shown in this study. The literature has revealed contradictory results about the effect of the parameters length and diameter on mini-implant stability. Differences lie mainly in the method and sample used. For example, experimental findings should be compared to clinical studies with caution, since *in vitro* measurements tend to more accurately describe the variable tested; however, they are far from simulating the actual clinical conditions. Clinical studies on the other hand, may report clinically applicable data, but do not provide an insight into the specific details of the research hypothesis. In a clinical study, *Miyawaki et al.* [2003] did not associate the length of the screw with its stability if the screw was at least 5 mm long. *Fritz et al.* in 2003 reported that 4 mm long screws offer adequate stability when compared with 6 mm and 8 mm screws. Also *Cheng et al.*

[2004] did not found implant length to have a significant correlation to implant failure clinically, but in their study the length was only determined by the transmucosal depth rather than by the depth of bone available for anchorage. The differences of those studies compared to this one can be attributed to the criteria used, meaning that mini-implant stability was mostly determined only by implant mobility or complete exfoliation, whereas in our study mini-implant displacements and rotations were recorded accurately in a level of micrometres and degrees. The same goes for a clinical study of Park *et al.* [2006], where implant-related factors (type, length, and diameter) were investigated and did not show any statistically significant differences in success rates among them. The short screws they used for the fixation in the study did not jeopardize their performance. Differences in results can also be explained by the applied force level. We found differentiation between the implant groups to be present when a high force of 2.5 N was applied. When the data were further analyzed the level of 1 N could be defined as threshold for differentiation. In the aforementioned studies the load was less or equal to 2 N so maybe no clear discrimination between groups could be observed clinically.

In an experimental design of *Wilmes et al.* [2006] it was also found that the length of the mini-implants does not have significant effects on their stability as measured by their insertion torque. At this point it should be noticed that the terms primary stability and insertion torque seem to be used interchangeably in many publications. However, this may not be correct, since there is still a controversy over the appropriateness of the use of maximum insertion torque as a measure of implant stability.

Studies have also examined the correlation between IT and axial pull-out strength (PS) to determine whether IT can predict screw retention in bone tissue and results are not unanimous. As Freiberg et al. [1999] reported for the correlation between implant placement resistance and bone density values, if the bone density value is almost the same the stability of the implant becomes almost equal although the initial insertion torque is different.

In contrast to the above-mentioned studies, *Tseng et al*. [2006] found the length of the inserted mini-implant to be an important risk factor, agreeing with the results of this study. These authors emphasized that the actual depth of insertion of mini-implant was more important than its length, the recommended length being at least 6 mm. This is in accordance with the general findings in the field of dental implants, where the shorter

and smaller diameter implants had lower survival rates than their counterparts [Winkler et al., 2005].
As for the implant diameter, it has been dominantly found to have great impact on mini-implant stability according to experimental and also clinical studies. *Miyawaki et al*. [2003] thought that the diameter of the mini-implants was significantly associated with their stability. They reported that 1 year success rate of implants with a 1.5 and 2.3 mm diameter was significantly higher than that of implants with diameters of 1 mm. They also found that patients with high mandibular plane angle showed a significantly lower success rate than those with an average or low angle, due to the thinner cortical bone in molar region. They concluded that the wider implants should be especially placed in patients with vertical facial growth. A study of *Berens et al.* [2006] was in accordance with the previous statements since they conclude that mini-implants of 2 mm in lower jaw increases success rates. They also recommend a mini-screw diameter of at least 1.5 mm in the palatal upper jaw. It has been suggested that implants smaller than 1.3 mm should be avoided, especially in the thick cortical bone of the mandible [Carano *et al.*, 2005]. Nevertheless, *Ohmae et al*. [2001] showed that mini-screws of 1 mm in diameter and 4 mm in length placed in the mandibular third premolar region of beagle dogs were able to sustain an intrusive force of 1.5 N for 12 to 18 weeks.
As far as implant type is concerned, the Aarhus mini-implant showed less rotation tendency than the LOMAS mini-implant at all force levels, despite of same dimensions and same conical design, thus indicating a better mechanical contact between the intra-osseous part of the Aarhus mini-implant and the bone. This may be attributed to slight differences in the shaft design.
The last parameter studied in this research, was the angle of implant insertion which was not found to be an influencing factor on primary stability at any force level. This is not in agreement with other studies, proposing a degree of angulation during insertion to increase implant to bone contact [Lim et al., 2008; Kyung et al., 2003; Carano et al., 2005]. The reason for this may be the method used. *Wilmes et al.* [2008] have found insertion angle to influence success rates by measuring the maximum insertion torque. According to them a 60° to 70° insertion angle would be advisable. The reason for the greater insertion torque found in these insertion angles may be due to the longer distance through the cortical bone when the implant is inserted in an oblique direction despite the smaller

insertion depth. But, as already discussed, insertion torque may not be the best representative of primary stability.

It should be emphasized that the main difference of this study, compared to previous ones, was that primary stability of mini-implants was examined by a direct, non-invasive laser-optical measurement of its deflections, instead of the indirect measurement of the insertion torque or pull out strength used before.

If we consider the multifactor aetiology of mini-implant failure, we could say that adequate implant dimensions may have co-synergic role on its stability and so such ones should be used where possible, especially when high forces are to be applied. Based on the results of this study, it could be suggested that when the applied force magnitude is less than 1 N, as during tooth intrusion or indirect anchorage, mini-implants with small dimensions can be safely used. When higher force levels are to be applied it seems that adequate mini-implant dimensions are essential for its primary stability and hence for its survival rate. However, clinical guidelines such as inter-root space and anatomic location should be taken into consideration.

6.2 Discussion of methods

6.2.1 Experimental

The research topic of this study was selected because of the high scientific interest on mini-implants in the fields of orthodontics during the last years and the lack of unanimous experimental data. The experimental protocol of this study was decided according to the biomechanical and clinical guidelines of mini-implants and bone. Mini-implants and animal bone, mini-implant insertion procedure, force level and force applications were carefully selected to represent as closely as possible the clinical situation. As mentioned above, the main difference of this study compared to other ones was that the primary stability of mini-implants was examined by a direct, non-invasive laser-optical measurement of its deflections, instead of the indirect measurement of the insertion torque or pull out strength used before.

The results of this study differentiated between the groups according to force levels. By low force application we could see the values of implant deflections to be approximately at the same level between the groups. At high force levels we could observe a differentiation in stability according to implant length and implant diameter. This may be, for example, attributed to the accuracy of the optomechanical system. The accuracy of the laser-optical system has been confirmed to be 0.1 mm and 0.2 degrees for registration of tooth or implant mobility [Hinterkausen et al., 1998]. The displacements measured especially during low force application in this study were far below this limit, so possible differentiation may have not been identified. Continuing, one could assume that the higher the force level applied, the greater the implant deflections and thus the more accurate the measurement registration and the observation of differences between groups. This may also be the reason for the larger standard deviation in the high force level groups presented in the tables and box-plots showed before.

Nevertheless we should notice that the greater rotation tendency of LOMAS compared to Aarhus mini-implants was identified at all force levels, meaning that the accuracy of the optomechanical system was high even at small value registration. This may mean that force level indeed differentiated the biomechanical performance of the various groups of mini-implants i.e. the 9 mm long and 2 mm wide mini-implants were actually displaced significantly less than the 7mm long and 1.5 wide ones.

6.2.2 Numerical

As far as the theoretical part of this study is concerned, it could be concluded that using the finite element method was a confirmation to our experimental data. Finite element mini-implant and bone models were accurately reconstructed from the original models by µCT scanning and the numerical results had a good agreement with the experimental ones. It seems that the FEM is developing to a useful and reliable tool for many sciences with many applications in the field of dentistry.

Apart from the benefits of the finite element analysis, the limitations of the method should also been taken into consideration. Starting from the first numerical models generated which were quite simple it was understood that the value of FE analysis entirely depends on element density and the accuracy of geometry, material properties and loading conditions of the examined structure. The more complex the model under investigation the more difficult the finite element analysis is. Problems can occur at every step during 3D model generation starting even from the µCT scanning. Improper position in the scan or improper scanning parameters can lead to model defects. At next stages, during 3D sur-

face model and 3D finite element model generation, problems can occur due to improper visualization, improper method and improper parameter installation (i.e. identification of structures in the slices, number of slices discretized, smoothing factor, material and mechanical parameters, boundary conditions), which may cause model distortion. The result of these may be the loss of information leading to poor quality of the reconstruction of the original structure or to a non functional model. As mentioned above, in this study one out of nine numerical models, named model ‘H’ and representing the LOMAS 1.5x9 mm mini-implant, did not work properly in the simulation due to unknown reasons and was excluded from further analysis.
Despite the limitations, commercial program packages offering the possibility of FE analysis undergo a constant improvement. Improved FE software, sophisticated programs and the computer technology lead day by day to more complex tooth models and model assumptions. Nowadays finite element methods are a very promising tool in dentistry and in orthodontics to understand a number of tissue structure and biomechanical problems.

6.3 Conclusions

- At low force levels (0.5 N) no statistically significant difference in displacement according to implant length and implant diameter could be observed.
- At high force levels (2.5 N), the 9 mm long mini-implants displaced significantly less than the 7 mm ones and the 2 mm wide mini-implants displaced significantly less than their 1.5 mm wide counterparts.
- The force level above which the implant length and implant diameter are statistically significant influencing parameters on implant stability was found to be the 1 N.
- The rotation of LOMAS mini-implants was significantly higher than that of the Aarhus mini-implants at all force levels. Aarhus mini-implants may present a better implant to bone interface.
- Implant insertion angle was not found to be an influencing parameter in this study at any force level.
- Numerical results showed a good agreement with the experimental ones.

Based on the results of this study, it could be suggested that when the applied force magnitude is less than 1 N, as during tooth intrusion or indirect anchorage, mini-implants with small dimensions can be safely used. When higher force levels are to be applied it seems that adequate mini-implant dimensions are essential for its primary stability and hence for its survival rate. However, clinical guidelines such as inter-root space and anatomic location should be taken into consideration.

7 Abstract

Objective: Mini-implants are being utilised as anchorage units in orthodontic treatment. Nevertheless, there seem to be influencing factors that interfere with their clinical performance. The aim of this study was to experimentally and theoretically examine four different parameters, which may have an influence on the primary stability of orthodontic mini-implants. These were 1) implant type, 2) implant length, 3) implant diameter and 4) insertion angle.

Material and Methods: A total of 90 mini-implants were inserted in fresh segments of bovine ribs. Implants were of two types, the Aarhus and the LOMAS mini-implant, of two lengths (7 mm and 9 mm) and of two diameters (1.5 mm and 2 mm, LOMAS only). A closed NiTi coil-spring was attached to each mini-screw. Half of the preparations were loaded with a low force of 0.5 N, the other half with a high force of 2.5 N. Mini-implant deflections during force application were non-invasively registered using a 3D laser-optical system. A subsequent finite element analysis of the applied force systems and the resulting mini-screw deflections was performed.

Results: In the small force group, implant displacements showed no statistically significant difference according to the investigated parameters. In the high force group the 9mm mini-implants displaced significantly less (mean 11±8 µm) than the 7 mm long (mean 22±11 µm, $p<0.01$), and the 2 mm wide significantly less (mean 9±2 µm) than the 1.5 mm ones (mean 22±2 µm, $p<0.001$). The force level where significance occurs was found to be 1 N. LOMAS mini-implants rotated significantly more than the Aarhus mini-implants at all force levels. Intra-observer agreement showed good correlation between experimental and numerical findings.

Conclusion: Implant length and implant diameter become statistically significant influencing parameters on implant stability only when a high force level is applied. Numerical results showed a good correlation to the experimental ones.

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