

An *in vitro* investigation of peak insertion torque values of six commercially available mini-implants

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SUMMARY This study compared peak insertion torque values of six commercially available self-drilling mini-implants [Mini Spider® screw (1.5 × 8 mm), Infitas® (1.5 × 9 mm), Vector TAS® (1.4 × 8 mm), Dual Top® (1.6 × 8 mm), Tomas Pin® (1.6 × 8 mm), and Ortho-Easy® (1.7 × 6, 8, and 10 mm)]. Twenty implants each were drilled into acrylic rods at a speed of 8 rpm using a motorized torque measurement stand, and the values were recorded in Newton centimetres (Ncm). A further 20 Ortho-Easy® implants with a length of 6 and 10 mm were tested at 8 rpm; 20 implants of 6 mm length were also tested at 4 rpm. Kaplan–Meier estimates of the peak torque values were compared using the log-rank test with multiple comparisons evaluated by Sidak's test.

There were significant differences in the maximum torque values for different mini-implants with the same length. The Mini Spider® screw and Infitas® showed the lowest average torque values (6.5 and 12.4 Ncm) compared with Vector TAS®, Dual Top®, Tomas Pin®, and Ortho-Easy® (30.9, 29.4, 25.4, and 24.8 Ncm, respectively). There was no correlation between the diameter of the implants and torque values. The Tomas Pin® showed the largest standard deviation (7.7 Ncm) and the Dual Top® implant the smallest (0.6 Ncm). Different insertion speeds did not result in significant differences in peak torque values but the 6 mm mini-implants showed significantly higher torque values than the 8 and 10 mm implants. Using a 'torque limiting' screwdriver or pre-drilling cortical bone to reduce insertion, torque appears justified for some of the tested implants.

Introduction

Over the past decade, temporary anchorage devices (TADs), also known as the cortical anchorage system, have become an important and popular part of orthodontic treatment. TADs summarize a group of adjunctive devices that are inserted intra-orally into bony structures to provide a form of anchorage, which is aimed at preventing unwanted tooth movement (anchorage loss). This has proven to be helpful in patients with inadequate dental anchorage potential and where conventional means of anchorage reinforcement are not applicable. The groups of TADs include screw-type implants (also known as miniscrews, mini-implants, microscrews, or micro-implants), onplants, microplates/miniplates (Bollards), zygoma implants, and palatal implants (Straumann Ortho-Implant®). The difference in nomenclature between micro- and mini- refers to the size of the screws but no definition has been universally accepted (National Institute for Health and Clinical Excellence, 2007).

Various clinical and experimental studies have shown that TADs are a reliable source of maximum anchorage because they are supposed to offer stable intra-oral anchorage in all three dimensions with only little reciprocal effect on other teeth or tooth groups, the latter relying mainly on periodontal support.

Screw-type implants (from now on referred to as mini-implants) form the subgroup of TADs most commonly used. While some variations between manufacturers exist with respect to design, shape, size, and material, they all have three basic features: a head, a trans-mucosal collar (or neck), and a threaded intraosseous body. Their size usually ranges from 4 to 20 mm in length and from 1 to 2.3 mm in diameter and they are usually made of titanium alloy. With this type of TAD, anchorage is achieved by placing the screws buccally or palatally/lingually into the alveolar bone between the roots of the teeth or into the palate.

The great popularity of mini-implants lies in the ability for the orthodontist to place them easily and quickly at the chair-side during routine appointments. This obviates the need for more invasive surgical procedures as are often required for zygoma and palatal implants. Chair-side insertion times from 5–8 (Gelgör *et al.*, 2004) to 10–15 (Chen *et al.*, 2006) minutes have been quoted.

Costa *et al.* (1998) described the use of mini-implants for immediate force application without any healing period: primary stability is used for mechanical retention. There are a number of potential unwanted side-effects associated with the use of mini-implants, including soft tissue damage, damage to roots of adjacent teeth, neurovascular damage, sinus perforation, local emphysema, and implant fracture.

Fracture of implants has only been reported during placement or removal and not during orthodontic force application (Buchter *et al.*, 2005). The most common reason for fracture is exposure to increased torsional stress during placement or removal of the mini-implant (Kravitz and Kusnoto, 2007). Torque has been described as the result of friction resistance between the screw threads and bone. Laboratory-based experiments have shown that torque forces of 23 Ncm and above have to be applied until fracture occurs (Carano *et al.*, 2005; Jolley and Chung, 2007). The incidence of screw fracture during clinical use has been reported to be 3–4 per cent (Chen *et al.*, 2006; Park *et al.*, 2006). The average torque measured at placement has been reported to be between 8.3 Ncm in the maxilla and 10 Ncm in the mandible (Motoyoshi *et al.*, 2006), while other authors describe higher torque values (15 Ncm) for successful insertion (Chaddad *et al.*, 2008). Motoyoshi *et al.* (2007) further suggested that the insertion torque should be higher than 8 Ncm but lower than 10 Ncm for improved long-term success, which is less than half of the torque theoretically necessary to fracture the mini-implants.

Jolley and Chung (2007) subdivided all potential factors, which may lead to implant fracture into the following categories: bone density at insertion site, insertion technique, experience of the clinician, and implant-related factors. Bone density varies within the mandible and maxilla and some investigators have shown that it increases from the anterior to the posterior segments of the jaws. The posterior region of the mandible in particular seems to be formed by denser and thicker cortical bone (Park *et al.*, 2008; Chun and Lim, 2009). The mean insertion torque in a clinical setting is higher in the mandible than in the maxilla, with values of 10.11 and 8.28 Ncm, respectively (Motoyoshi *et al.*, 2009).

Thick cortical bone with a high bone density may constitute a risk for mini-implant fracture especially if a self-drilling mini-implant with a small diameter is used: implant placement resistance correlates positively with bone density and with implant diameter (Friberg *et al.*, 1995).

There are two insertion techniques for mini-implants: self-tapping (pre-drilled) and self-drilling. For the self-tapping technique the implant is placed after a pilot hole is drilled into the bone, using a motorized handpiece. While some systems have a pre-drill for the cortical part of the bone only, others offer pilot drills for nearly the whole length of the screw. However, there are only a few (less than five) manufacturers that offer pre-drilling mini-implants.

The majority of commercially available mini-implants are now delivered with a sharp cutting tip and placed directly into the bone, either manually with a screwdriver or with a slow motorized handpiece (self-drilling technique). The self-drilling technique has the following perceived advantages: 1. higher success rate due to closer implant bone contact and good initial stability (93 versus 86 per

cent; Heidemann *et al.*, 1998; Chen *et al.*, 2008), 2. reduction of root damage due to slower and therefore better controlled insertion and quicker handling as drilling a pilot hole becomes unnecessary. The perceived disadvantage of this technique is the higher risk of mini-implant fracture compared with the pre-drilling technique when the cortical bone is very dense and thick. Increased pressure may be needed to insert a self-drilling mini-implant leading to higher insertion torque (Chen *et al.*, 2008). Those authors suggested that self-drilling implants should not be used in areas with high bone density even though they offer many advantages.

Zipprich *et al.* (2007) measured torque throughout insertion and concluded that the risk of fracture not only depends on the material and design of the mini-implant but also on the experience of the clinician and the insertion technique. Zipprich *et al.* (2007) measured torque variations of up to 45 per cent between clinicians for the same make of implant and concluded that lower more constant forces during insertion were less likely to cause fracture compared with higher forces.

Commercially available mini-implants vary in diameter from 1 to 2.3 mm. While a diameter of a smaller size may be advantageous to reduce the risk of damaging adjacent teeth, some studies have shown that a reduction in mini-implant diameter may decrease the success rate as well as the mechanical stability of these implants.

Jolley and Chung (2007) tested 20 mini-implants from five different manufacturers *in vitro*, by measuring the peak torque force needed to fracture the screws when turning these into a polycarbonate rod. The findings revealed that all mini-implants fractured apart from the Orthoimplant®, which had the greatest diameter. The mean peak torque value at fracture correlated positively with the diameter of the screw (Jolley and Chung, 2007). Other authors (Carano *et al.*, 2004) compared the mechanical properties of two mini-implants from the same manufacturer with different diameters (1.3 and 1.5 mm). The results demonstrated that the implants ($n = 6$) with a diameter of 1.3 mm presented considerably less resistance to bending and torsional strength than 1.5 mm implants, suggesting that the diameter of a mini-implant is directly correlated with mechanical stability and that a reduction of the diameter by 0.2 mm can have a significant effect on the mechanical properties, reducing the force needed for failure by half.

Mini-implants are available in two shapes: cylindrical (core of constant diameter) and tapered (diameter of core gradually decreases from head to tip). For a tapered implant with an increasing diameter from implant tip to head, the torque value increases rapidly during insertion. The cylindrical shape has a gradual increase of torque (Lim *et al.*, 2008). This suggests that tapered implants may be more prone to fracture. Carano *et al.* (2005) concluded that mini-implants with a cylindrical shape exhibit superior

mechanical qualities to mini-implants with a tapered shape. Interestingly, the contrary was shown by Jolley and Chung (2007). The tapered Orlus® implant showed higher mean torque values at fracture than the cylindrical Tomas Pin®, although both had the same maximum diameter. Jolley and Chung (2007) argued that this was due to the diameter of the tapered implant, which was measured at the midpoint of the screw length, and as the diameter was wider in the area of the neck, the mini-implant possessed superior mechanical properties.

It is yet not fully understood which role implant shape has on fracture resistance and more research on this topic is required. The thread design of a mini-implant also varies widely. It can differ in height, pitch, and depth as well as shape. Some screws have an interrupted thread in order to relieve pressure as the implant is turned into the bone. As the greatest stress is found at the tip of each thread, discontinuation of the thread may help reduce the build-up of stress (Lee *et al.*, 2007).

To date, there are no studies in which the relationship between the mechanical properties of mini-implants and thread design have been compared. At least theoretically, a wider thread and a narrower core diameter should exhibit inferior properties to an implant with the same external diameter but a wider core diameter and a reduced thread depth.

Most commercially available mini-implants are made of titanium or titanium alloys. There are a number of studies comparing the mechanical properties of mini-implants made of different materials. Carano *et al.* (2005) compared three implants of which two were made of titanium and one of stainless steel. The results illustrated that the mini-implant made of stainless steel displayed considerably higher resistance to torsional and bending force, but the authors noted that the amount of mechanical resistance of the titanium implants was sufficient for clinical use.

Iijima *et al.* (2008) reported that implants made of titanium alloys fracture at a significantly higher mean torque value than pure titanium implants. The addition of molybdenum, vanadium, tantalum, niobium, manganese, iron, chromium, cobalt, nickel, copper, or silicon achieve stabilization of the beta phase, which gives the screw greater strength. Iijima *et al.* (2008) used implants where the beta phase was stabilized by the addition of vanadium, iron, and manganese. Adding these elements decreased biocompatibility as vanadium in the titanium–aluminium–vanadium alloy may cause cytotoxic and adverse tissue reactions. However, due to their favourable mechanical properties, titanium alloys are widely used. It appears that the ideal implant material that combines the mechanical strength with a high level of biocompatibility has yet to be manufactured.

There is also a wide variation in the head design of mini-implants. Heads may be shaped as buttons, triangles, spherical heads, with or without holes, or as cross-sectional heads. Different designs allow for attachments of auxiliaries,

such as elastomeric chains, nickel–titanium coil springs, wire ligatures, and archwires (for insertion of power arms).

The authors are not aware of any literature on how the head and neck designs may impact on the mechanical stability of mini-implants. However, it has been reported that mini-implants can fracture at the neck during screw removal (Kravitz and Kusnoto, 2007), so the neck of a mini-implant may be a weak point.

The aim of this study was to investigate the relationship of fracture resistance of six commercially available mini-implants and to try to draw conclusions regarding the variation of their physical characteristics (core versus outer diameter). Two different insertion speeds and three different lengths of mini-implants of one brand were used to determine whether those two parameters had any impact on the above values.

Lastly, as no suitable *in vitro* scenario exists, which can be used to train operators to avoid dangerously high torque values during implant placement, this study also assessed three commercially available synthetic bone substitutes for suitability for training.

Materials and methods

Six commercially available mini-implant systems of different geometries were analysed (Table 1). Twenty samples of each were used to give a total sample size of 160. In subsidiary comparisons, the effect of length was evaluated (6, 8 and 10 mm) as well as the effect of insertion speed (4 or 8 rpm for 6 mm) for the Ortho-Easy® system. Human bone was not considered as an insertion substrate due to the difficulties in obtaining a homogeneous sample that would allow for reproducible testing.

In a pilot study, several substrates were evaluated. Three synthetic bone substitutes (Schmid and Dirr, 1980): two types of Sawbone® mandible (Pacific Research Laboratories, Vashon, Washington, USA) and Synbone®, a synthetic bone material with a 3 mm layer of synthetic cortical bone (Synbone®, Malans, Switzerland). Preformed circular acrylic and Tufnol (RS, Corby, Northants, UK) were tested. Hexagonal heat-cured dental acrylic rods were prepared using a hexagonal Stabilo point 88® (Schwan-Stabilo, Heroldsberg, Germany) pen as a pattern. The ends of the pen were trimmed and blocked with wax for preparation of a plaster mould. The mould was filled with Trevalon/Universal Clear™ (Dentsply, York, Pennsylvania, USA) and cured for 6 hours at 95°C. The finished rods were cut into 18 mm lengths.

Peak torque values were measured using a motorized torque measurement stand (MTMS)—Mark 10 (Metrology International Ltd, Harrogate, Yorkshire, UK; Figure 1). The substrate was mounted in the lower chuck, which rotated at a controlled speed. The mini-implants were mounted via their appropriate drivers in the upper chuck, which measured the torque values in Ncm.

The mini-implant was lowered until it was in contact with the substrate material and then held to maintain the

Table 1 Six commercial mini-implant systems of different geometries.

Type	Manufacturer	Diameter/mm	Length/mm
Mini Spider® screw	Health Development Company, Sarcedo, Vicenza, Italy	1.5	8
Infitas®	DB Orthodontics, Silsden, West Yorkshire, UK	1.5	9
Vector TAS®	Ormco, Orange, California, USA	1.4	8
Dual Top®	Jeil Medical Corporation, Seoul, Korea	1.6	8
Tomas Pin®	Dentaurum, Pforzheim, Germany	1.6	8
Ortho-Easy®	Forestadent	1.7	6, 8, and 10

**Figure 1** Image of the motorized torque measurement stand.

pressure. The lower chuck was rotated counter clockwise at the required speed until the mini-implant fractured and the corresponding peak torque value was recorded.

Data were analysed using Stata 11 (College Drive, Texas, USA), with significance pre-determined at $\alpha = 0.05$. The null hypothesis was that the mini-implant design had no significant effect on the peak torque value. Kaplan–Meier estimates of survivor function for peak torque values were compared using the log-rank test with multiple comparisons being evaluated by Sidak's test.

Results

In the pilot experiment using Ortho-Easy® mini-implants, none of the synthetic bone was sufficiently resistant to fracture the mini-implants. The implants rotated freely in the synthetic bone after the thread had been 'forced through' the sample. When preformed acrylic rods were used only one mini-implant in five fractured and the remainder deformed on contact with the surface. The preformed Tufnol rods split in their long axis

with no fracture or deformation of the mini-implant. However, as all mini-implants fractured in the heat-cured acrylic rods, these were subsequently used to measure torque values.

Univariate summary statistics and the Kaplan–Meier survival curves are given in Table 2 and Figures 2 and 3. Figure 2 shows the results for the six tested mini-implants. The equality of survival curves was tested using the log-rank test in conjunction with Sidak's test for multiple comparisons. The following mini-implant pairs were not significantly different: Vector TAS®/Tomas Pin® and Dual Top®/Tomas Pin®. There was no significant difference for the 6 mm Ortho-Easy® at 4 or 8 rpm. All other mini-implant pairs (8 mm length at 8 rpm) were significantly different. The Tomas Pin® showed the largest standard deviation (SD) (7.65) while the Dual Top® and Vector TAS® were the most consistent (SD 0.61 and 0.81, respectively).

The Spider® and Infitas® screws fractured at a mean of 6.4 and 12.5 Ncm, respectively, while the Ortho-Easy®, Tomas Pin®, Dual Top®, and Vector TAS® fractured at mean values of 24.8, 25.4, 29.4, and 30.9 Ncm, respectively.

Figure 2 shows that torque values for the Spider® screws decreased steeply around 8 Ncm and for the Infitas® screw at around 12 Ncm. The values for the Tomas Pin® show two mini-implant losses at values below 10 Ncm and a more gradual fracture slope compared with all other mini-implants. The consistency of the material appeared not to be as homogeneous as that of the other manufacturers. One Ortho-Easy® screw failed at 5 Ncm.

Figure 3 shows the two speeds and three lengths tested for the Ortho-Easy® implants. Sidak's test for paired comparisons showed no significant difference for speed (4 and 8 rpm for Ortho-Easy®, 6 mm). There were significant differences for the three lengths 6, 8, and 10 mm. The 6 mm implant was the strongest followed by the 10 mm implant; the 8 mm implant showed the lowest values.

Discussion

Synthetic bone materials are often used in 'hands on' courses as well as for testing of dental implants. These materials are designed to help simulate the tactile sensation of the clinical scenario. The pilot test in this study revealed that the torque values produced on the Sawbone® mandibles (5.7–11.1 Ncm) were similar to those during clinical placement (7.2–13.5 Ncm,

Table 2 Univariate summary statistics for peak torque for each mini-implant: N is the sample size, τ is the mean torque, and SD is its associated standard deviation.

Mini-implant	Length/ mm	Speed/ rpm	N	τ /Ncm	SD/Ncm
Mini Spider® screw	8	8	20	6.59	2.00
Infinitas®	9	8	20	12.39	1.95
Vector TAS®	8	8	20	30.88	0.81
Dual Top®	8	8	20	29.44	0.61
Tomas Pin®	8	8	20	25.42	7.65
Ortho-Easy®	6	4	20	31.96	1.73
	6	8	20	32.19	1.77
	8	8	20	24.83	4.57
	10	8	20	26.95	2.48

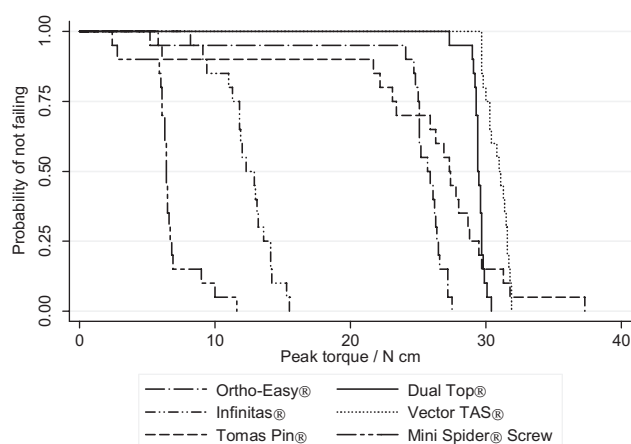


Figure 2 Kaplan-Meier survival estimates of the six tested mini-implants.

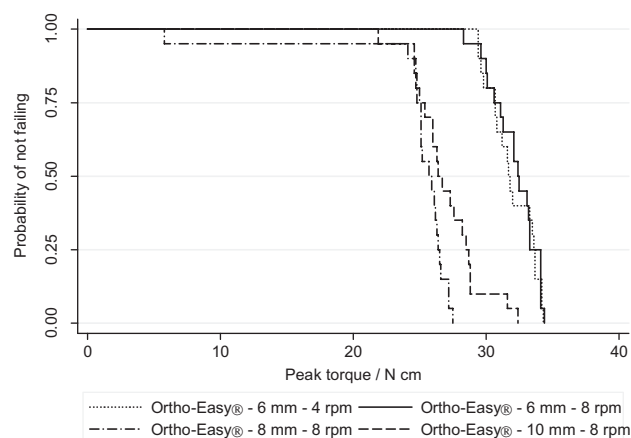


Figure 3 Kaplan-Meier survival estimates for the Ortho-Easy® system at different lengths and speeds.

mean 10 Ncm; Motoyoshi *et al.*, 2006) of human jawbones. Hence, Sawbone® mandibles exhibit sufficient qualities for training purposes regarding the tactile experience for insertion but they are not sufficiently resilient to allow higher torque build-up of stress to allow the mini-implants to fracture.

Bone substitute (Synbone AG), made of polyurethane foam, showed higher torque resistance (15.5–29.2 Ncm) than those recorded for Sawbone®; this material is supposed to resemble the femur and not the mandible and is manufactured with a 3–4 mm thick layer of ‘cortical bone’ replacement. This was still not found to have sufficient resilience for testing mini-implants *in vitro* as none of the mini-implants in the pilot test fractured. Hence, synthetic bone was not found to be the ideal testing material for this study. Instead, heat-cured acrylic rods were chosen similar to materials used by other authors: polycarbonate (Jolley and Chung, 2007) and polyvinyl chloride plates (Heidemann *et al.*, 1998). The heat-cured denture acrylic used in this study was soft enough to allow for the insertion of the mini-implants and at the same time sufficiently resistant to allow the mini-implants to fracture. The objective of this study was to compare the torque values at fracture rather than comparing mini-implants in a clinical setting. Preformed acrylic rods were chosen initially to control the homogeneity of the material; these however did not allow insertion of mini-implants. Therefore, acrylic rods made from the same acrylic mix were used to minimize inconsistencies.

A slow insertion speed has been postulated by some authors to allow bone to adapt to the stresses generated, the latter possibly contributing to mini-implant fracture (Kravitz and Kusnoto, 2007). In this study, no significant differences in torque fracture values were shown when comparing slow (4 rpm) with fast (8 rpm) insertion speeds. However, it is difficult to determine whether the chosen speed levels can be actually referred to as slow and fast as no speed levels are suggested in the current literature for manual insertion. The torque measuring assembly used was the limiting factor regarding the speed levels tested; 8.5 rpm was the maximum. Speeds of 8 and 4 rpm were chosen for this study, which were in the range of speed levels used in previous *in vitro* studies: 0.5–6 rpm (Jolley and Chung, 2007; Song *et al.*, 2007; Lim *et al.*, 2008).

From clinical experience, an average manual insertion speed of up to 12 rpm assuming that one complete rotation is achieved by two manual turns, which takes about 5 seconds. Ludwig *et al.* (2008) recommended an insertion speed of 25 rpm with the motorized insertion technique.

Comparison of the six mini-implants showed significant differences in the torque values between most of the tested groups. Only the combinations Vector TAS®/Tomas Pin® and Dual Top®/Tomas Pin® were not significantly different from each other. Interestingly, the Tomas Pin® had the largest standard deviation and the Vector TAS® and Dual Top® the smallest standard deviations.

All mini-implants tested fractured at the intraosseous part rather than in the region of the head and neck, and it is hence unlikely that head and neck designs have an impact on the mechanical properties leading to different peak torque values. Even though the incidence of fracture has been reported in the literature, no author has consistently described the fracture location during implant placement. Kravitz and

Kusnoto (2007) are the only authors who mentioned fracture of the screw in the area of the neck during implant removal.

The data from this investigation did not show good correlation of implant diameter and torque resistance of individual designs. The Spider® and Infinitas® screws mainly fractured at the lower third at significantly lower torque values than the mini-implants of other manufacturers; the diameter of those screws was 1.5 and 1.6 mm, respectively. The Vector TAS® implant however showed the highest mean torque values although the latter had the smallest diameter (1.4 mm) of the implants tested. The low torque values obtained for the Spider® and Infinitas® screws are not surprising as the shape of these implants is different to the other makes: they have a more tapered shape of the intraosseous screw compared with all the other implants, which are cylindrical for most of the length. It would be consistent that lower torque resistance is due to the thinner tip: diameter values given by the manufacturers are usually measured at the middle of the screw. The Vector TAS® implant has a very shallow thread with a relatively wide core and this may explain their high torque values. The findings of investigation are in contrast to the results of Jolley and Chung (2007) but confirm the findings of Carano *et al.* (2005). The former authors implied that an implant with a tapered shape may exhibit superior mechanical properties as the diameter of the upper third of the screw is larger than the stated outer diameter. Jolley and Chung (2007) however did not take into account that the tip of the screw consists of a smaller diameter and can therefore fracture more easily at the tip, as demonstrated in this study. The reason for the difference in the mechanical properties found in the study of Jolley and Chung (2007) may not have been due to implant shape but to other factors, such as material composition and method of production.

A direct relationship between screw diameter and torque value at fracture has been shown in a number of studies (Jolley and Chung, 2007; Song *et al.*, 2007; Lim *et al.*, 2008). According to the result of this investigation, however, it appears that core diameter does not solely determine the torsional resistance of a mini-implant.

The correlation between core diameter and mechanical strength has been described for screws used for bone fixation in oral and maxillofacial surgery (Perren, 1976). Mini-implants are derived from surgical screws and hence, similarities in their mechanical properties can be expected.

While the width of the core diameter gives a useful explanation for the significant differences in mean torque values at fracture, this does not explain the high consistency of values for the Vector TAS® and Dual Top® implants and the high variation of values for the Tomas Pin® and Ortho-Easy® screws. Other possible explanations for the differences may lie in the composition of the titanium alloys used: pure titanium compositions have been shown to be more brittle and hence, most of the mini-implants are manufactured with a high-grade titanium alloy (Ludwig *et al.*, 2008). Even though all the tested mini-implants were

made of titanium alloy grade V, there may be slight differences in purity/composition and this may have had an impact on the obtained values. Differences in implant production may also play a part in the variation found in this study as implants are manufactured in various ways. They can either be cut into shape from a titanium rod or be rolled and twisted. Details of how the various implants were made could not be obtained for all the makes used in this study.

The difference in the quality of the connection between the head and the driver can also play a part in the differences found in this study. Even though the head and neck designs of the screws may not influence the mechanical properties of the mini-implants, the connection of the implant head to the driver may well do. Some of the implants allowed for a firm connection between the screw head and the driver, whereas others showed some 'play'. This can lead to tipping of the screw during insertion thus causing fracture at lower torque values as the screw is turned into the acrylic with an off-centre rotation. While this may help explain the very low torque values, it will not explain the high values obtained for some mini-implants.

The testing procedure used in this study required constant pressure of the screw to the acrylic rod, which was regulated manually by moving a handle attached to the machine. Other investigators used a weight to standardize pressure on the implants (Jolley and Chung, 2007; Song *et al.*, 2007; Lim *et al.*, 2008). This possible inconsistency should however have affected all samples in a similar way. Further investigations that consider the above possibilities in greater detail may be useful, particularly as testing of 20 Tomas Pins®, using a pre-drilling technique however, did not exhibit such a wide range of torque values (Jolley and Chung, 2007).

According to Motoyoshi *et al.* (2006), mini-implants with a diameter of 1.6 mm and a length of 8 mm should ideally sustain a placement torque between 7.2 and 13.5 Ncm. The torque values obtained for most of the implants tested in this study were well above those values: Dual Top® and Vector TAS® screws failed well above the minimum values as did most of the Tomas Pin® and Ortho-Easy® implants: apart from two and one outlier, respectively. All the Spider® screws fractured at torque values which fell into the range of the average placement torque or fractured below that value. The Infinitas® screws fractured at just above the expected maximum clinical torque values, around 12 Ncm. In a clinical setting, the Spider® screw in particular and the Infinitas® screw to a lesser extent may be at risk of fracture compared with all other implants tested in this study. This is particularly interesting as Motoyoshi *et al.* (2006) measured placement torque after pre-drilling a pilot hole. It has to be assumed that the clinical placement torque for self-drilling implants is even higher.

There are two main strategies to avoid high insertion torque values, which could lead to implant fracture during insertion: using a 'torque limiting' screwdriver and pre-drilling pilot holes in areas where thick compact bone can be expected (such as the palatal bone of the maxilla). Both these should be considered for some of the mini-implants tested in this study.

Conclusions

Overall, the torque resistance of the materials tested resembles that of human bone. None of the synthetic bone materials investigated were sufficient to fracture the mini-implants and these materials are not designed for training surgeons to avoid dangerously high insertion torques.

Significant differences were found for peak torque values between the different implant manufacturers: the Mini Spider® and Infinitas® screws showed the lowest average torque values (6.5 and 12.4 Ncm) compared with the other implants: Vector TAS®, Dual Top®, Tomas Pin®, and Ortho-Easy® (30.9, 29.4, 25.4, and 24.8 Ncm, respectively). The Tomas Pin® showed the largest standard deviation (7.7) and the Dual Top® the lowest (SD 0.6). The more tapered Spider® and Infinitas® screws had the lowest mean torque values; however, the Vector TAS® implant with the smallest overall diameter had the highest average values.

This study failed to demonstrate an inverse correlation between the diameter of the mini-implants and their peak torque values. It hence appears that factors such as material composition, production technique, and the ratio between core and thread play an important role in determining the torque resistance of mini-implants.

The head design of a mini-implant does not appear to have a significant influence on the peak torque values as none of the tested mini-implants fractured at this level. There was no significant difference for peak torque values for the two different insertion speeds tested for the 6 mm Ortho-Easy® implants (4 and 8 rpm). The shorter Ortho-Easy® implants (6 mm) were significantly more torque resistant than longer ones (8 and 10 mm). Using a torque limiting screwdriver and/or pre-drilling cortical bone to reduce insertion torque appear justified for some of the tested implants to reduce the risk of fracture.

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References

- Buchter A, Wiechmann D, Koerdt S, Wiesmann H P, Piffko J, Meyer U 2005 Load-related implant reaction of mini-implants used for orthodontic anchorage. *Clinical Oral Implants Research* 16: 473–479
- Carano A, Lonardo P, Velo S, Incorvati C 2005 Mechanical properties of three different commercially available miniscrews for skeletal anchorage. *Progress in Orthodontics* 6: 82–97
- Carano A, Velo S, Incorvati C, Poggio P 2004 Clinical applications of the mini-screw-anchorage-system (M.A.S.) in the maxillary alveolar bone. *Progress in Orthodontics* 5: 212–235
- Chaddad K, Ferreira A F, Geurs N, Reddy M S 2008 Influence of surface characteristics on survival rates of mini-implants. *Angle Orthodontist* 78: 107–113
- Chen C H *et al.* 2006 The use of microimplants in orthodontic anchorage. *Journal of Oral and Maxillofacial Surgery* 64: 1209–1213
- Chen Y, Shin H I, Kyung H M 2008 Biomechanical and histological comparison of self-drilling and self-tapping orthodontic microimplants in dogs. *American Journal of Orthodontics and Dentofacial Orthopedics* 133: 44–50
- Chun Y S, Lim W H 2009 Bone density at interradicular sites: implications for orthodontic mini-implant placement. *Orthodontics and Craniofacial Research* 12: 25–32
- Costa A, Raffaini M, Melsen B 1998 Miniscrews as orthodontic anchorage: a preliminary report. *International Journal of Adult Orthodontics and Orthognathic Surgery* 13: 201–209
- Friberg B, Sennerby L, Roos J, Lekholm U 1995 Identification of bone quality in conjunction with insertion of titanium implants. A pilot study in jaw autopsy specimens. *Clinical Oral Implants Research* 6: 213–219
- Gelgör I E, Büyükyılmaz T, Karaman A I, Dolanmaz D, Kalayci A 2004 Intraosseous screw-supported upper molar distalization. *Angle Orthodontist* 74: 838–850
- Heidemann W, Gerlach K L, Grobel K H, Kollner H G 1998 Drill free screws: a new form of osteosynthesis screw. *Journal of Cranio-Maxillo-Facial Surgery* 26: 163–168
- Iijima M, Muguruma T, Brantley W A, Okayama M, Yuasa T, Mizoguchi I 2008 Torsional properties and microstructures of miniscrew implants. *American Journal of Orthodontics and Dentofacial Orthopedics* 134: 333.e1–e6
- Jolley T H, Chung C H 2007 Peak torque values at fracture of orthodontic miniscrews. *Journal of Clinical Orthodontics* 41: 326–328
- Kravitz N D, Kusnoto B 2007 Risks and complications of orthodontic miniscrews. *American Journal of Orthodontics and Dentofacial Orthopedics* 131: S43–S51
- Lee J, Kim J, Park Y-C, Vanarsdall R J J 2007 Application of orthodontic mini-implants. Quintessence Publishing Co, Inc., Chicago, Chapter 3, pp. 29–50
- Lim S A, Cha J Y, Hwang C J 2008 Insertion torque of orthodontic miniscrews according to changes in shape, diameter and length. *Angle Orthodontist* 78: 234–240
- Ludwig B *et al.* (eds) 2008 Mini-implants in orthodontics: innovative anchorage concepts. Quintessence Publishing Co., Ltd, London
- Motoyoshi M, Hirabayashi M, Uemura M, Shimizu N 2006 Recommended placement torque when tightening an orthodontic mini-implant. *Clinical Oral Implants Research* 17: 109–114
- Motoyoshi M, Inaba M, Ono A, Ueno S, Shimizu N 2009 The effect of cortical bone thickness on the stability of orthodontic mini-implants and on the stress distribution in surrounding bone. *International Journal of Oral and Maxillofacial Surgery* 38: 13–18
- Motoyoshi M, Yoshida T, Ono A, Shimizu N 2007 Effect of cortical bone thickness and implant placement torque on stability of orthodontic mini-implants. *International Journal of Oral and Maxillofacial Implants* 22: 779–784
- National Institute for Health and Clinical Excellence 2007 Interventional procedure overview of mini/micro screw implantation for orthodontic anchorage. National Institute For Health and Clinical Excellence, London
- Park H S, Jeong S H, Kwon O W 2006 Factors affecting the clinical success of screw implants used as orthodontic anchorage. *American Journal of Orthodontics and Dentofacial Orthopedics* 130: 18–25
- Park H S, Lee Y J, Jeong S H, Kwon T G 2008 Density of the alveolar and basal bones of the maxilla and the mandible. *American Journal of Orthodontics and Dentofacial Orthopedics* 133: 30–37
- Perren S M 1976 Force measurements in screw fixation. *Journal of Biomechanics* 9: 669–675
- Schmid F, Dirr B 1980 [Measurements of the transfer of force by bone screws]. *Deutsche Zahnärztliche Zeitschrift* 35: 28–31
- Song Y Y, Cha J Y, Hwang C J 2007 Mechanical characteristics of various orthodontic mini-screws in relation to artificial cortical bone thickness. *Angle Orthodontist* 77: 979–985
- Zipprich H, Glasl B, Ludwig B, Lietz T 2007 Risikopotenzial und Lösungsstrategien. In: Ludwig B (ed). *Mini-Implantate in der Kieferorthopädie*, Quintessence, Berlin, pp. 121–133