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Fracture resistance of orthodontic mini-implants: a biomechanical *in vitro* study

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SUMMARY Sufficient primary stability is of importance for the survival of orthodontic mini-implants. This means that adequate torque has to be achieved during insertion. However, as moments exceeding the fracture resistance of a mini-implant may result in their fracture, the maximum torque load capacity should be known. In this study, the threshold torque values resulting in the fracture of various mini-implant types and diameters were evaluated.

Forty-one different mini-implants with diameters ranging from 1.3 to 2.0 mm (Aarhus screw, Abso Anchor, Ancora, Bone screw, Dual Top, Lomas, MAS, O.S.A.S, Ortho Easy, Spider Screw, and Tomas pin) were inserted in acrylic glass by a robot system. Ten specimens of each mini-implant type were tested. The insertion torque was measured and the maximum torque at the time of mini-implant fracture was evaluated. Significance of the mean value differences was evaluated by Kruskal–Wallis tests.

Fracture moments varied depending on the diameter of the mini-implants. The measured values ranged from 108.9 Nmm (MAS 1.3 × 11 mm) to 640.9 Nmm (Lomas 2.0 × 11 mm). The differences were highly statistically significant ($P < 0.001$).

The risk of mini-implant fracture should be borne in mind at the time of insertion, especially if mini-implants with a small diameter are employed. To minimize the risk of fracture, pre-drilling should be carried out if the mini-implants are to be inserted at a site with a high bone density.

Introduction

Skeletal anchorage, and especially orthodontic mini-implants, have attracted attention in recent years because of their versatility, minimal surgical invasiveness, and low cost (Kanomi, 1997; Costa *et al.*, 1998; Melsen and Costa, 2000; Wilmes 2008). Sufficient primary stability, measured by insertion torque, seems to play a major role in the survival rate during treatment (Melsen and Costa, 2000; Motoyoshi *et al.*, 2006, 2007). An insertion torque of 5–10 Ncm (50–100 Nmm) for mini-implants with a diameter of 1.6 mm is recommended to minimize the risk of failure (Motoyoshi *et al.*, 2006, 2007). Higher values may cause mini-implant fracture (Büchter *et al.*, 2005a; Wilmes *et al.*, 2006; Präger *et al.*, 2008; Reicheneder *et al.*, 2008). Depending on the level of mini-implant fracture, removal can be difficult. In addition, depending on the insertion site, adjacent structures may be damaged at the time of removal of the mini-implant segment. Therefore, the maximum torque load capacity of mini-implants at the time of insertion seems to be crucial. However, this aspect has not been systematically investigated and, to-date, has been addressed mostly in the orthopaedic literature (Collinge *et al.*, 2000; Merk *et al.*, 2001; Perren *et al.*, 2001). Some mini-implant manufacturers provide data on the fracture resistance of their products but without presentation of the specific test protocols.

The aim of the present study was therefore to analyse the threshold torque values resulting in fracture of different mini-implant types and diameters with a validated and standardized measurement appliance.

Materials and methods

Forty-one mini-implant types (all made of titanium grade 5, Ti-6Al-4V) were investigated (Table 1). The implants were inserted into acrylic glass, which was chosen instead of bone due to its high homogeneity. In total, 82 acrylic blocks (1 × 1 × 5 cm) were prepared (two for each mini-implant type). Pre-drilling was performed perpendicular to the surface of the acrylic block using a bench drilling machine (Opti B 14 T; Rexon, Hilden, Germany) at 915 rpm. The pre-drilling diameter was chosen for each mini-implant in such a way that the implant could be manually inserted in the acrylic glass to a depth of 4 mm, where full thread engagement was established. The appropriate pre-drilling diameter was evaluated for every mini-implant diameter before the main study was conducted. With this insertion mode, it was ensured that the implant fractured in every instance when the robot completed the final insertion and torque measurement.

The following drills were used: 1.0 and 1.3 mm: Dual Top system (Jeil Medical Corporation), 1.1 and 1.2 mm: tomas drill (Dentaurum), 1.5 mm: Lomas system (Mondeal

Table 1 Torque fracture values and used pre-drilling diameters for the tested mini-implant types (length and diameter provided by the manufacturers).

Mini-implant	Minimum (Nmm)	Median (Nmm)	Maximum (Nmm)	SD	Pre-drilling (mm)
Aarhus screw 1.5 × 8 mm (Medicon, Tuttlingen, Germany)	187.89	228.71	272.84	29.32	1.1
Aarhus screw 1.5 × 9.6 mm (Medicon)	174.69	213.26	251.68	29.69	1.1
Aarhus screw 2 × 9.6 mm (Medicon)	378.94	496.52	557.53	66.54	1.5
AbsoAnchor® 1.4 × 8 mm (Dentos, Daegu, Korea)	138.67	183.39	213.26	22.46	1.1
AbsoAnchor® 1.4-1.3 × 8 mm (Dentos)	114.66	155.55	209.05	25.72	1.1
AbsoAnchor® 1.6 × 8 mm (Dentos)	170.04	207.40	233.52	19.26	1.0
AbsoAnchor® 2 × 10 mm (Dentos)	467.18	533.97	578.24	30.18	1.5
Ancora® (20–10) 2 × 10 mm (Serf, Rochecorbon, France)	343.67	391.40	410.46	21.87	1.5
Bone screw 1.7 × 8 mm (Stryker®, Cambridge, Massachusetts, USA)	282.44	324.31	343.07	23.29	1.5
Bone screw 2 × 10 mm (Stryker®)	465.98	554.75	593.55	37.82	1.6
Dual Top™ (G2) 1.6 × 10 mm (Jeil, Seoul, Korea)	259.33	287.69	330.02	27.26	1.3
Dual Top™ (Type G2) 2 × 10 mm (Jeil) with cross shaft	470.35	521.44	589.80	39.53	1.5
Dual Top™ (Type G2) 2 × 10 mm (Jeil) with hexa shaft	467.78	514.91	533.22	22.45	1.5
Dual Top™ (Type JD) 2 × 10 mm (Jeil)	488.19	491.65	505.30	9.05	1.5
Dual Top™ (Type G2) 1.6 × 8 mm (Jeil)	239.37	264.28	287.84	15.96	1.1
Lomas 1.5 × 9 mm (PSM, Tuttlingen, Germany)	181.44	222.56	271.04	25.32	1.0
Lomas 2 × 11 mm (PSM)	529.16	571.86	640.97	32.12	1.5
Lomas 2 × 9 mm (PSM)	425.76	456.90	491.80	19.59	1.5
Micro-anchorage system (MAS) 1.3 × 11 mm (Micerium, Avegno, Italy)	108.95	147.37	172.44	17.81	1.0
Micro-anchorage system (MAS) 1.5 × 11 mm (Micerium)	147.67	194.27	222.11	24.33	1.1
Micro-anchorage system (MAS) 1.5 × 14 mm (Micerium)	182.34	198.02	253.63	22.25	1.0
O.S.A.S 1.6 × 8 mm (Dewimed, Tuttlingen, Germany)	267.13	298.95	336.32	20.46	1.1
Ortho Easy 1.7 × 8 mm (Forestadent, Pforzheim, Germany)	266.53	282.22	294.45	6.98	1.3
Spider Screw® C1 (SLP-1508) 1.5 × 8 mm (HDC, Sarcedo, Italy)	157.58	202.00	239.37	20.86	1.1
Spider Screw® C1 (SSM-1508) 1.5 × 8 mm (HDC)	150.08	224.96	239.37	27.19	1.2
Spider Screw® C2 (SLP-2011) 2 × 11 mm (HDC)	367.53	426.14	463.43	27.75	1.5
Spider Screw® C2 (SSM-2011) 2 × 11 mm (HDC)	108.05	427.49	471.54	105.07	1.5
Spider Screw® K1 (SCL-1508) 1.5 × 8 mm (HDC)	141.67	150.00	164.33	6.48	1.0
Spider Screw® K1 (SCL-1510) 1.5 × 10 mm (HDC)	121.71	144.60	212.36	24.70	1.0
Spider Screw® K1 (SCR-1508) 1.5 × 8 mm (HDC)	235.17	245.45	261.88	8.20	1.1
Spider Screw® K2 (SCL-1909) 1.9 × 9 mm (HDC)	437.02	486.77	519.26	28.85	1.3
Spider Screw® K2 (SCR-1911) 1.9 × 11 mm (HDC)	361.38	375.79	441.52	26.64	1.3
Tomas® (08 AG) 1.6 × 8 mm (Dentaurum, Ispringen, Germany)	296.25	338.79	360.33	19.84	1.1
Tomas® (10 AG) 1.6 × 10 mm (Dentaurum)	323.86	347.05	363.03	11.77	1.3
Tomas® (N08) 1.6 × 8 mm (Dentaurum)	277.34	315.31	348.92	19.49	1.1
Tomas® (N10) 1.6 × 10 mm (Dentaurum)	256.63	316.28	347.42	23.81	1.2
Tomas® (SD; acid etched surface) 1.6 × 8 mm (Dentaurum)	284.54	335.87	411.81	36.42	1.1
Tomas® (SD; sandblasted surface) 1.6 × 8 mm (Dentaurum)	233.37	267.58	305.40	21.06	1.1
Tomas® (SD10) 1.6 × 10 mm (Dentaurum)	291.30	315.98	334.97	16.90	1.3
Tomas® (SD6) 1.6 × 6 mm (Dentaurum)	259.48	323.86	396.65	42.51	1.1
Tomas® (SD8) 1.6 × 8 mm (Dentaurum)	256.33	278.16	296.55	13.85	1.1

Medical Systems; GmbH, Tuttlinger, Germany), and 1.6 mm: Aarhus screw (Medicon).

After pre-drilling and prior to the measurement of the fracture torque, 10 mini-implants of each type were manually inserted using the handheld screwdriver of the respective mini-implant system (five mini-implants in each acrylic block, Figure 1). Subsequently, the mini-implants were screwed by the robotic measurement system. The central component of the measuring system is a precision robot (RX60; Stäubli Tec-Systems GmbH, Bayreuth, Germany), which was equipped with an angle sensor (Fritz Kübler GmbH, Villingen-Schwenningen, Germany) and a custom-made docking unit for the driver shafts of the respective mini-implant systems. For measurement of the moments, a torque sensor (Burster Präzisionsmesstechnik GmbH, Gernsbach, Germany) was coupled with the custom-made holder of the acrylic glass specimens. The analogue signals delivered by the sensors were digitized by a multichannel measuring device Spider 8 (Hottinger Baldwin Messtechnik GmbH, Darmstadt, Germany) and were stored in a personal computer. The software of the measuring system was programmed such that the robot arm performed five rotations of 360 degrees within 5 seconds (Figures 2 and 3). Due to the solidity of the acrylic glass, the mini-implants fractured during these five rotations. The insertion torque was measured and recorded during complete insertion (Figure 4). All maximum insertion torques (at the time of fracture) were then transferred to a pivot table (Excel 2003, Microsoft) and categorized depending on the mini-implant type. Significance of the mean value differences was evaluated by Kruskal–Wallis tests using the Statistical Package for Social Sciences version 17.0 (SPSS

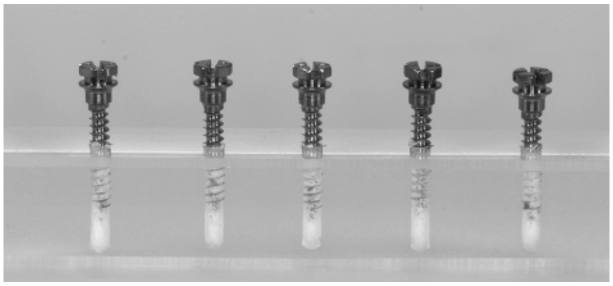

Figure 1 Five Aarhus screw mini-implants manually inserted in an acrylic block before measurement of fracture torque.

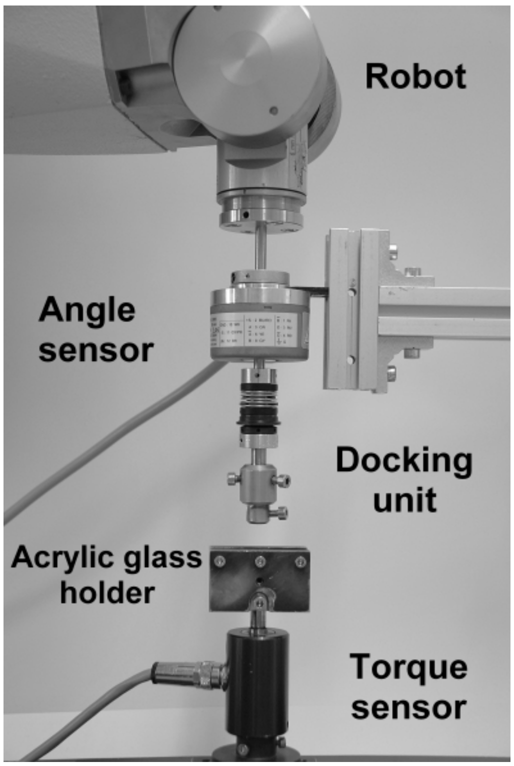

Figure 2 Robotic measurement system.

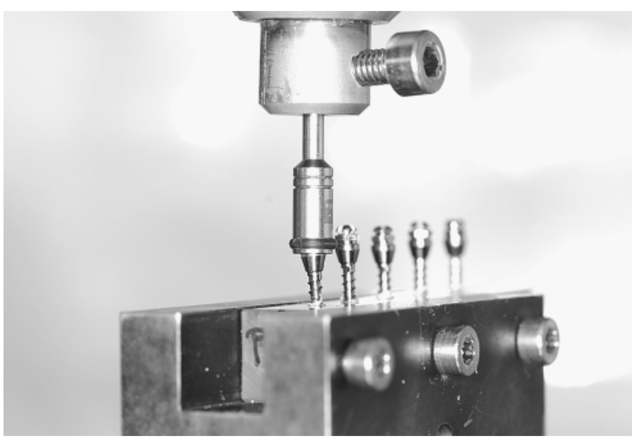

Figure 3 Robot-controlled insertion of a mini-implant (Tomas pin SD) in acrylic glass.

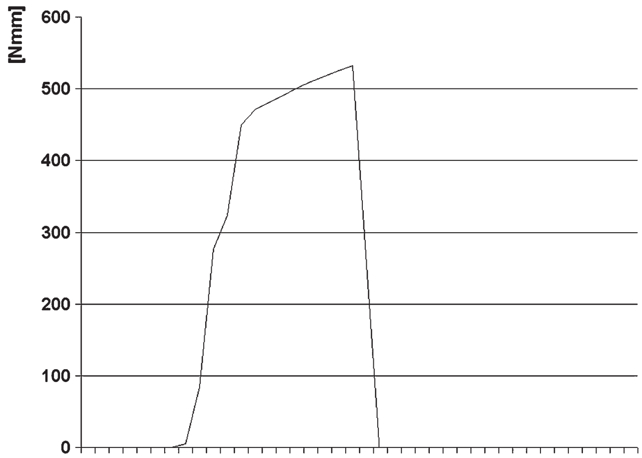

Figure 4 Insertion torque (y-axis) for one mini-implant until fracture (In this case, Bone screw 2 × 10 mm, Stryker). The x-axis represents the insertion procedure.

Inc., Chicago, Illinois, USA). The maximum error was limited to $P < 0.05$.

Results

Fracture torque varied significantly (108.9–640.9 Nmm) depending on the mini-implant type (Figure 5; Table 1). The diameters of the mini-implant had a major impact on fracture torque values: 1.3 mm fractured at median moments of 147.3 (±17.81) Nmm, 1.4 mm at 161.5 (±26.3) Nmm, 1.5 mm at 205.6 (±38.2) Nmm, 1.6 mm at 297.8 (± 48.0) Nmm, 1.7 mm at 287.0 (±23.8) Nmm, 1.9 mm at 438.5 (±56.1) Nmm, and 2.0 mm at 491.6 (±75.3) Nmm (Figure 6). The differences were highly statistically significant ($P < 0.001$).

Depending on the mini-implant type and the employed driver shaft, different fracture patterns were found: almost all mini-implant types fractured at the level of the acrylic block in the region of the thread part mini-implants (Figure 7a). The self-tapping type Tomas pin fractured between the head and the thread of the mini-implant (Figure 7b). Employing the cross driver shaft, the Dual Top Screw (G2) fractured at the interface to the driver.

Discussion

In this study, fracture moments for most of the currently used mini-implant types were identified. The diameter of the mini-implant had a major impact on fracture torque values. Thus, it seems advantageous to use mini-implants with a larger diameter that additionally have the advantage of higher primary stability (Wilmes *et al.*, 2006, 2008a,b; Lim *et al.*, 2008) resulting in lower failure rates (Motoyoshi *et al.*, 2006, 2007). However, available space in the alveolar process is limited due to root proximity (Poggio *et al.*, 2006; Präger *et al.*, 2008). As a consequence, the anterior palate

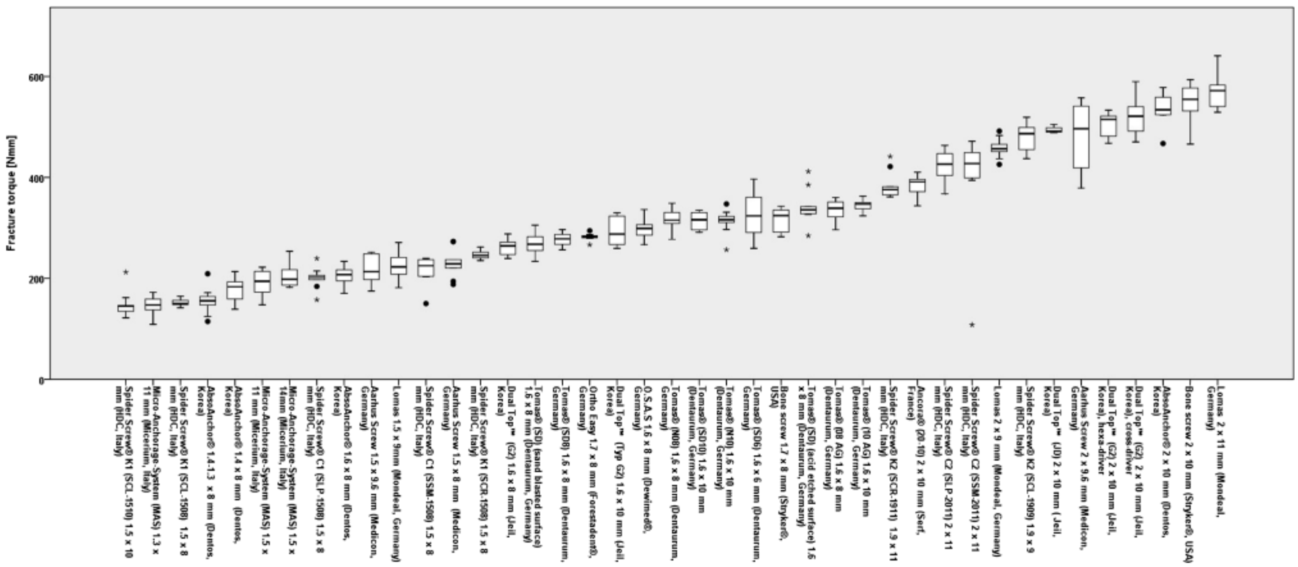

Figure 5 Boxplots of the torque fracture values for the tested mini-implant types.

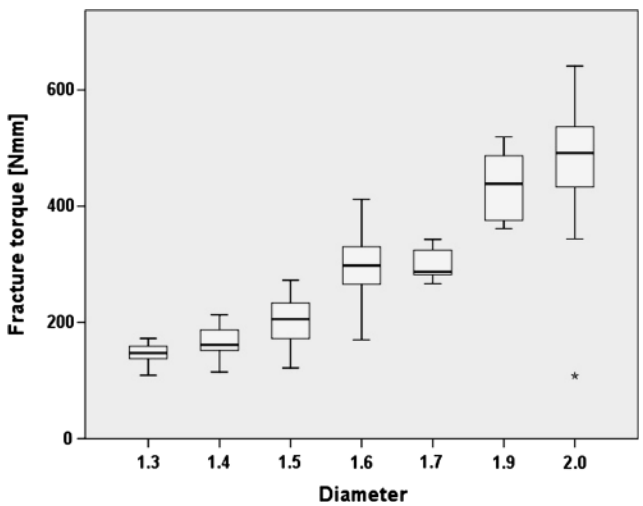

Figure 6 Boxplots of the torque fracture values depending on the diameter of the tested mini-implant. Mini-implants with the same diameter are pooled.

can be chosen as an alternative insertion site since mini-implants with wider diameters could be inserted, if anchorage in the maxilla is required, e.g. for distalization or anchorage of the molars/anterior teeth. Using the anterior palate as the anchorage unit, mini-implants with exchangeable abutments are recommended to establish safe coupling with the appliance (Wilmes and Drescher, 2008).

The measurement system used in the present research was appropriate to evaluate the fracture torques in a standardized way. Even though acrylic glass differs from human bone, it was chosen for its homogeneity, thus providing reproducibility and comparability of the measurements. Acrylic glass was preferred since it is harder compared with bone resulting in fracture even of mini-implants with large diameters. It should be noted that the pre-drilling diameters employed in this study were chosen in order to achieve correct thread engagement prior to measurement. As such, they were only of technical relevance and have no clinical implications. More specific information regarding the selection of pre-drill diameters has been reported (Wilmes and Drescher, 2009).

Only a few studies have been conducted to evaluate the fracture torque of orthodontic mini-implants. Carano *et al.* (2004, 2005) measured a mean value of resistance to breakage in torsion at 487 Nmm for mini-implants with a diameter of 1.5 mm. All tested screws with a diameter of 1.5 mm (MAS, Absoanchor, and Leone) could withstand 400 Nmm. Interestingly, this is more than twice that for the mini-implants in the present study with a diameter of 1.5 mm: 206 (±38) Nmm (Figure 5; Table 1). According to Carano *et al.* (2004, 2005), the mean value of resistance to breakage in torsion is 234 Nmm for mini-implants of 1.3 diameter. The current findings were 147 ± 18 Nmm, which again is much lower. The reason for these differences may be due to the measurement set-up. Additionally, the calibration and accuracy of the torque measurement device were not specified in the studies of Carano *et al.* (2004, 2005).

Lietz (2008) itemized fracture torques of different mini-implant types derived from the manufacturers' product data sheets. For mini-implants with diameters from 1.3 to 2.5 mm, they reported fracture torque values from 210 to 900 Nmm. These results are comparable with the present measured data.

To reduce the risk of mini-implant fracture, stainless steel mini-implants that are resistant to higher moments compared

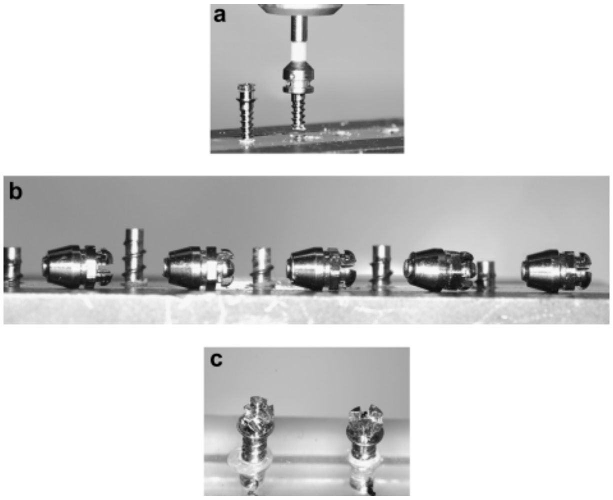

Figure 7 Mini-implant fracture at (a) the level of the acrylic block (Dual Top 2 × 10 mm) (b) between the head and the thread of the mini-implant (self-tapping type Tomas pin), and (c) at the interface to the driver [employing the cross driver shaft, Dual Top Screw (G2)].

with titanium mini-implants should be used (Carano *et al.*, 2005; Lietz, 2008). Due to their stainless steel surface, osseointegration is not possible (Lietz, 2008), which may have an impact on failure rates. However, a correlation between the alloy of the mini-implants and their failure rates has not been investigated yet. This should be evaluated in future clinical or animal studies.

A second approach to reduce fractures is pre-drilling, if mini-implants are inserted in a region with a high bone quality such as the mandible and anterior palate (Wilmes *et al.*, 2006; Wilmes and Drescher, 2009). This holds true especially in adults where a higher bone quality is encountered compared with children (Wilmes *et al.*, 2006). Furthermore, very high insertion torques may result in higher failure rates due to a distinctive bone compression with microdamage (Wawrzinek *et al.*, 2008). This theory has been discussed both for dental (Büchter *et al.*, 2005b) and mini-implants (Motoyoshi *et al.*, 2006, 2007; Wawrzinek *et al.*, 2008).

A third approach is the use of torque-controlled drivers or ratchets, offered by some of the mini-implant manufacturers. Unfortunately, the accuracy of some of these devices is not well known. A more reliable alternative is a dental surgical unit with electronic torque control. If the programmed torque moment is reached, the engine stops. If this occurs, the mini-implant should be unscrewed or left in place if the position is clinically acceptable. If there is no pre-drilling before mini-implant insertion, pre-drilling should be conducted before a second attempt at insertion. If there is pre-drilling before mini-implant insertion, pre-drilling with a larger diameter should be conducted before the second attempt at insertion (Wilmes and Drescher, 2009).

It is recommended to adjust the insertion torque limit to a value lower than the lowest fracture value in the current study, depending on the mini-implant-type (minimum; Table 1). In addition, tipping or flexion during insertion of the mini-implant should be avoided because this increases the risk of facture (Reicheneder *et al.*, 2008).

There are no studies available reporting how often mini-implant fracture occurs in clinical situations. Since at the Department of Orthodontics, University of Düsseldorf, only one out of more than 1500 mini-implants has fractured, this complication can be rated as very rare. The reason for the very low fracture rate may be the fact that pre-drilling is performed in almost all patients. Büchter *et al.* (2005a) reported that eight of 200 mini-implants fractured in an animal study at the time of insertion and two at the time of removal. This seems to be a very high rate and was probably caused by the high density of pig bone. However, some clinicians have reported mini-implant fractures even at the time of explanation, especially when pure titanium mini-implants were used. For this reason, mini-implants made of pure titanium are no longer available. Nowadays, a titanium alloy (titanium grade 5, Ti-6Al-4V) is employed (Lietz, 2008).

Roughening of the mini-implant surface was found to increase the risk of mini-implant fracture. The sandblasted Tomas pins SD (manufactured for a study on osseointegration) had lower fracture torques compared with the polished regular Tomas pins (Figure 5; Table 1).

If a mini-implant fracture occurs during insertion, should the fragment be removed? As mentioned above, three different fracture types were identified depending on the mini-implant type and the driver shaft employed:

1. Most of the mini-implants fractured at the level of the acrylic block (Figure 7a), which clinically represents the surface of the cortical bone. Since the soft tissues and surrounding bone have to be removed, extraction of the intraosseous fragment is difficult. It should be unscrewed (e.g. by Weingart pliers), if there is no risk of damaging the adjacent roots at the time of exposure of the fragment by a bur. If easy removal of the fractured segment is not possible, leaving the fragment in the bone is an alternative due to a high tissue tolerance to titanium. The patient should be informed about this issue.
2. The self-tapping type of the Tomas pin fractured between the head and the thread of the mini-implant (Figure 7b). Depending on the depth of insertion, removal of the fractured segment seems easier. However, the self-tapping Tomas pin will probably seldom fracture in clinical situations due to very low insertion moments caused by its thread design (Wilmes *et al.*, 2006; Su *et al.*, 2009).
3. Employing the cross driver shaft, the Dual Top Screw (G2) fractured at the interface to the driver (Figure 7c). If ligation of a segmental archwire (indirect anchorage) is no longer possible, the mini-implant can be used by attaching a power chain/coil spring (direct anchorage). This means that orthodontic use is still possible.

Conclusions

1. The diameters of the mini-implants had a major impact on fracture torque values. Thus, it seems advantageous to use mini-implants with a larger diameter.
2. If mini-implants are inserted at a site with high bone quality, pre-drilling seems reasonable even for self-drilling mini-implants to minimize the risk of mini-implant fracture.

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