Original article

Laser-treated stainless steel mini-screw implants: 3D surface roughness, bone-implant contact, and fracture resistance analysis

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Summary

Background/Objectives: This study investigated the biomechanical properties and bone-implant intersurface response of machined and laser surface-treated stainless steel (SS) mini-screw implants (MSIs).

Material and Methods: Forty-eight 1.3 mm in diameter and 6 mm long SS MSIs were divided into two groups. The control (machined surface) group received no surface treatment; the laser-treated group received Nd-YAG laser surface treatment. Half in each group was used for examining surface roughness (Sa and Sq), surface texture, and fracture resistance. The remaining MSIs were placed in the maxilla of six skeletally mature male beagle dogs in a randomized split-mouth design. A pair with the same surface treatment was placed on the same side and immediately loaded with 200 g nickel–titanium coil springs for 8 weeks. After killing, the bone-implant contact (BIC) for each MSI was calculated using micro computed tomography. Analysis of variance model and two-sample t test were used for statistical analysis with a significance level of P <0.05.

Results: The mean values of Sa and Sq were significantly higher in the laser-treated group compared with the machined group (P <0.05). There were no significant differences in fracture resistance and BIC between the two groups.

Limitation: animal study

Conclusions/Implications: Laser treatment increased surface roughness without compromising fracture resistance. Despite increasing surface roughness, laser treatment did not improve BIC. Overall, it appears that medical grade SS has the potential to be substituted for titanium alloy MSIs.

Introduction

The achievement of ‘absolute’ anchorage has always been a challenge in orthodontics. The earliest attempt to use a screw or implant as skeletal anchorage to facilitate orthodontic treatment was reported in 1945 (1). Later, Linkow presented a different approach for anchorage reinforcement with an endosseous blade implant (2). Skeletal anchorage, however, did not become widely accepted until the first successful case report in 1983 of a surgical
Vitallium bone screw used to intrude maxillary central incisors (3). Fourteen years later, Kanomi (4) reported that titanium mini-implants (1.2 mm in diameter and 6 mm long) can be used to control vertical and horizontal tooth movement without undesired side effects. Since then, numerous clinical reports and experimental studies have been published showing that mini-screw implants (MSIs) have become a crucial tool in contemporary orthodontic mechanics (5).

Most modern MSIs are fabricated from medical type IV or V titanium alloy with a thread diameter ranging from 1.2 to 2.0 mm and a length from 4.0 to 12.0 mm (6). Their applications were limited because of the anatomic structures of the jaws. Since most MSIs are inserted into inter-radicular areas, the risk of damage to roots, nerves, blood vessels, nasal, and maxillary sinuses, bone, etc. still exists (7). The inter-radicular distances were reported as 1.60–3.46 mm in the maxilla and 1.99–4.25 mm in the mandible with increasing widths from the CEJ to the apex (8). To avoid damaging structures, MSI diameters should be reduced as much as possible. Based on a volumetric tomographic image study, an ideal titanium MSI placed in the inter-radicular spaces is recommended to be a conic-shaped implant, with a cutting thread of 6–8 mm in length, and a maximum diameter of 1.2–1.5 mm (9). A study showed that the optimal length of MSIs with a diameter of 1.3 mm is 5 mm in the maxilla and 6 mm in the mandible (10). However, the decrease in MSI diameter reduces its strength and can possibly result in fracture during insertion (11) and removal (9, 12). When fractured, it is sometimes difficult to remove small MSI tips from bone (13). To avoid fracture, the MSIs material should have adequate strength to resist insertion torque. Alternatively, instead of Ti, 316 stainless steel (SS) can be considered for its low cost, high yield strength, corrosion resistance, and easy processing (6, 14).

Stainless steel is one of the most widely used biomaterials and has been used for fixation of bone segments for decades (6). Although the stability of MSIs is expected from the mechanical interlock and osseointegration between screw and bone (15), the formation of a fibrous tissue capsule around SS MSIs may interrupt proper bone-implant contact (BIC) (16). Current evidence suggests that a high percentage of BIC is not desirable for temporary anchorage because MSIs must be easily removed at the end of treatment (13). In addition, some mobility is acceptable with unidirectional light force (12). Deguchi et al. (17) speculated that low BIC could successfully tolerate 200–300 g of orthodontic force. However, a certain amount of osseointegration around SS MSIs may still be required for successful anchorage (18).

Many articles proposed a strong correlation between surface roughness and BIC (19–22). Among additive and subtractive methods used for surface texturing, laser treatment is a new approach (22). Several studies have noted that the removal torque of laser-treated titanium was significantly higher when compared with machined groups (11, 23–26). A tendency for more bone formation on laser-treated surface was also observed (27). Laser roughening enhances hardness, roughness, corrosion resistance, and the degree of purity on modified surfaces (28, 29). Besides, laser ablation creates microporosity in a strictly regular pattern by melting superficial layer of metal surfaces (24, 29, 30). Several studies showed that cells involved in osseointegration had preferable reactions with highly ordered and uniform surface roughness than with random roughness (31, 32). The pattern of surface microstructure can easily be controlled by manipulating the parameters of laser machines including: pulse frequency, output energy, scanning speed, and pulse intervals (33, 34).

Nd:YAG laser (wave length 1064 nm) was commonly used for changing surface texture of titanium implants (11, 23–27) due to its higher absorption on metallic surface compared to other types of lasers (35). It has been shown that Nd:YAG laser irradiation can modify the metallic surfaces and cause extensive melting on titanium disc even at the lowest energy setting (36). This laser roughening process can enhance bone-implant surface interaction of SS MSIs. However, whether laser treatment can affect the strength of SS MSIs and bone reaction around them has not been fully evaluated. The aims of this study were to 1. investigate surface roughness, surface texture, and fracture resistance of SS MSIs with machined surface and Nd:YAG laser-treated surface, 2. explore whether SS MSIs can be suitable substitutes of titanium alloy MSIs for orthodontic loading, and 3. determine whether or not the surface modification by Nd:YAG laser improves the bone response around immediate loaded SS MSIs.

Materials and methods
A total of 48 MSIs (316 SS, 1.2–1.3 mm in diameter, 6 mm long) were customized from medical grade 316 SS (Dentos, Daegu, Korea) and randomly assigned to two surface texture groups (Figure 1). One group received no surface treatment (machined surface); the other was treated with Nd:YAG/passively Q-switched laser (laser-treated surface) according to the manufacturer protocol. Half of the samples (12 machined and 12 laser-treated) were used for surface roughness and strength evaluation and the other half was used in a prospective in vivo experiment to compare BIC.

Surface roughness was measured directly with a non-contact scanning white light interferometer, ZeGage (Zygo Corporation, Middlefield, Connecticut, USA). Samples were placed directly on the stage for measurements on the valley and inclined at approximately 60 degrees on a V-block for measurements on the flank (Figure 1).

![Figure 1. MSIs with machined surface (left) and laser-treated surface (right).](image)
Surface roughness was measured on a standardized area of approximately 200×200 μm on the valley and 160×160 μm on the flank (Figure 2). The Sa (arithmetical mean height, mean surface roughness) and Sq (root mean square height, standard deviation of the height distribution) measurements were repeated three times on each MSI. All images were acquired and analysed with the integrated ZeMaps software in micrometers.

For evaluation of surface characteristics, two SS MSIs from each group were examined under a scanning electron microscope (SEM). Each MSI was set on aluminium SEM specimen mounts by using double stick carbon tape (Electron Microscopy Sciences, Hatfield, Pennsylvania, USA). SEM images were captured digitally on an Amray 1910 Field Emission scanning electron microscope with SEMTech Solutions X-Stream Imaging software (SEMTech solutions, North Billerica, Massachusetts, USA). The surface features were examined on the valley between third and fourth threads with low (500×) and high magnification (5000×).

Following the topographic measurements and SEM, fracture resistance was assessed as the maximum torque at the time of fracture. Each MSI was inserted into a pilot hole (1.0 mm diameter, 3 mm deep) in homogenous acrylic blocks 1×1×1.8 cm. The MSIs were hand driven half the length of the threads into the acrylic (Figure 3). Once the MSIs were firmly seated in the pilot holes, each sample block was connected to an engine driver attached to a digital torque gauge (Mark-10, Copiague, New York, USA). The block with MSI was secured on the stage and rotated 2 degrees per second until the MSI was fractured. The peak torque value at MSI fracture was indicated as fracture resistance at accuracy of 0.5 Ncm (Figure 3).

Six skeletally mature female beagle dogs (ages 10–15 months; weight, about 9 kg) were selected and approved by Institutional Animal Care and Use Committee. Using a randomized split-mouth design, two MSIs with the same surface texture were placed on one side of the maxilla and immediately loaded with 200 g of force. Similarly, a pair of MSIs with the other type of surface was placed and loaded on the other side of the maxilla. Inter-radicular areas of the third premolars and the first molars were appropriate placement locations according to diagnostic periapical radiographs.

A surgical stent was fabricated with two pieces of SS rectangular wires and light-curing temporary filling material (Spident, Incheon, Korea). Each wire was marked every 3 mm as a gauge (Figure 4). All MSIs were placed perpendicular to the buccal alveolar bone with an implant motor at 15 rpm following initial insertion by a hand driver. Periapical radiographs were taken during placement to verify the clearance with the roots.

The 200 g tension loads were applied with nickel–titanium (NiTi) coil springs (Dentsply GAC International, Bohemia, New York, USA) with activation >3 mm (Figure 4). The length was measured at the first day and final day of experiment to confirm that the 200 g of force was maintained. The appliances were inspected and irrigated with 0.2 per cent chlorhexidine every week. Stabilities were evaluated at insertion and at week 8, and the survival rate was calculated (Figure 5A). Any MSI that showed mobility greater than 1 mm when tested with cotton pliers, or that had pulled out by the spring, was deemed a failure.

Eight weeks after placement, the animals were euthanized and each maxilla was resected en bloc (Figure 5B) and stored in 70 per cent ethanol followed by methylmethacrylate (MMA) embedding prior to sectioning for scanning. Each bone-MSI sample was trimmed parallel to the long axis of the MSI (Figure 5C) and placed into a specimen carrier for scanning (Figure 5D and 5F). Each specimen

Figure 2. 3D scanning images of surface roughness. (A) Valley of machined surface. (B) Flank of machined surface. (C) Valley of laser-treated surface. (D) Flank of laser-treated surface.
was scanned by µCT 35 (Scanco Medical, Basserdorf, Switzerland). The resolution was set at 1 slice every 6 μm and a X-ray energy level of 70 kVp, 114 μA, and 8W was used. The three-dimensional (3D) BIC was calculated as the ratio of bone volume to total volume (BV/TV) in the zone of 18–36 μm from the MSI surface. The zone of 0–18 μm from the MSI surface was excluded to minimize metallic artifacts of the voxel adjacent to the MSI.

After scanning, all bone-MSI samples were ground along the axis of the MSIs to slices of 150–170 μm thick with an Exakt system (EXAKT Technologies, Oklahoma City, Oklahoma, USA) and stained with Toluidine blue. Images of each MSI were captured under 4× magnification using Nikon microscopy and merged using Image Composite Editor (Microsoft Research, Mountain View, California, USA) software (Figure 5E and 5G). The two-dimensional (2D) BIC was evaluated as the length of bone in direct contact with the thread of the MSI divided by the total length of the MSI thread multiplied by 100.

Statistical methods
Surface roughness, peak fracture torque, and 3D and 2D BIC of each group were summarized as mean and standard deviation. ANOVA model with repeated measurements was used to test the effect of groups (control and treated) for each outcome (Sa and Sq separately for the flank and the valley. The same statistics was performed to compare the mean values of Sa and Sq on the flank and the valley in the same group. For comparison of peak fracture torque, ANOVA model was used. Two sample t-test was used to evaluate the effects of surface treatment (machined, laser-treated) on 3D and 2D BIC. The overall significance level was set at 5 per cent.

Results
Three SS MSIs demonstrated mobility of less than 1 mm. They were all machined type, located at the first molars, but firm enough for clinical use. Thus, consistent with the above defined criteria, there were no failures. The three coil springs associated with the mobile MSIs became shorter, but they remained stretched more than 3 mm at week 8, meaning that all MSI loads were maintained at 200 g of force throughout the experiment. Most MSIs at the third premolar demonstrated bicortical anchorage with the perforation of the maxillary sinus wall (Figure 5). The mean values of Sa and Sq were significant higher in the laser-treated group (Sa: 0.36 ± 0.076 μm; Sq: 0.54 ± 0.199 μm) than the machined group (Sa: 0.10 ± 0.051 μm; Sq: 0.17 ± 0.149 μm). Laser treatment increased surface roughness more than 3-fold (Table 1).

The mean values of Sa and Sq on the valley (Sa: 0.13 ± 0.02 μm; Sq: 0.21 ± 0.03 μm) were significantly higher than on the flank in the control group, but not in the laser-treated group (valley: 0.38 ± 0.03 μm;
flank: 0.36 ± 0.08 μm) (Table 1). Under SEM, laser-modified surface texture was characterized with spherical elevations and extensive melting (Figure 6). There was no significant difference in the mean values of peak torque at fracture between machined (13.42 ± 0.63 Ncm) and laser-modified (13.08 ± 0.85 Ncm) MSIs (Table 2). No significant differences were found for the mean values of 3D and 2D BIC between two groups (Table 3).

**Discussion**

Several parameters can be used for measuring surface roughness, but Sa is most commonly used for 3D studies (37, 38). In addition, Sq, another 3D parameter of surface roughness, was used for roughness measurement because Sq is more sensitive to extreme values and statistically more powerful than Sa (39, 40). Laser treatment increased Sa and Sq more than 3-fold (Table 1). This indicates that Nd-YAG laser treatment improved surface roughness by increasing both averaged surface peak heights and deviations of the surface peak heights with a uniform surface texture on the surface of the thread. In the machined surface control group, Sa and Sq were higher on the valley than on the flank areas (Table 1). In the fabricating process, a MSI is usually ‘cut’ from a metal rod. The relatively smooth surface on the flank area is due to the machining process when cutting the metal rod (41). Our results suggest that laser surface treatment increases surface roughness on both valley and flank areas of the MSI thread.

Consistent with surface roughness measurement, SEM images also show that surface texture was modified to become rougher by laser treatment. In previous studies, a ring structure with central holes and melting pearls was identified on laser-roughened CP Ti implant (23, 24, 30). The 200 μm pores created on the surface of Ti6Al4V implants by laser texturing was suggested as the optimal size for new bone formation (33, 34). Compared to machined surface (Figure 6A and 6B), the roughness was obviously increased on laser-treated surface (Figure 6C and 6D). Round projections and extensive melting were observed (Figure 6D), but a regular pattern of laser work usually seen on laser-treated titanium implants was not prominent in this study. This is probably because the periodic striation pattern is attenuated by extensive melting due to lower melting temperature of SS (42). Although linear scratches and surf-like pattern crossing over the scratches were apparent on the machined surface (Figure 6A), the machined surface looked smooth even at high magnification (Figure 6B).
The average peak torque at fracture was not statistically different between machined and laser-treated groups (Table 2), indicating that laser treatment did not alter the strength of the MSIs. Based on the data released from the manufacturer, the average torque strengths of these 1.3 mm in diameter × 6 mm long MSIs are not higher than machined titanium alloy MSIs in identical shape and size. It seems to be in conflict with the current knowledge that SS has higher Young modulus compared with titanium (43). However,
the superiority of mechanical strength between SS and Ti MSIs cannot be simply concluded due to fracture torques measurements performed by different operating settings. Carano et al. (44) showed that SS MSIs started bending at lower level of forces regardless of the values of failure load twice higher than Ti MSIs. The original properties of SS probably were altered by heat and pressure created during cutting process. The screw geometry and materials compositions can also change the screw’s mechanical strength (44). Therefore, further studies regarding how the screw design and manufacturing process affects MSI mechanical properties are still needed.

Ure et al. (15) demonstrated that primary and secondary stability affected clinical stability of MSIs. Maximal primary stability was obtained immediately by mechanical retention mainly from cortical bone and gradually reduced while bone remodelling occurred around MSIs. Secondary stability formed by osseointegration increased over time and compensated the loss of primary stability starting at week 3 for maintaining clinical stability of MSIs. Secondary stability is usually indicated by BIC. Although laser treatment significantly increased roughness, it did not improve 3D BIC. Albrektsson and Wennerberg suggested that the optimal bone response can be induced by the moderate roughness (Sa of 1–2 μm) of the majority of commercial dental implants (37, 38). However, in this study, even the laser-treated surface would be considered as a smooth surface (Sa less than 0.5 μm) (37), which explains why laser-treatment did not increase BIC compared with the relatively smoother machined surface.

In this study, the ratio of BV/TV, the relative amount of bone per unit of volume, was used to represent BIC in the 3D analysis. Because of the radiographic scatter induced by the metal implant thread (45), it is impossible to assess BV/TV immediately adjacent to the surface of MSIs. The zone of 18–36 μm from the implant surface has been recommended as the best estimate for BV/TV (46). Ikeda et al reported that 3D BIC (BV/TV) ratios of medullary bone ranged from 15 to 36 per cent in the zone of

Table 2. Comparison of the mean values of peak fracture torque.

<table>
<thead>
<tr>
<th>Group</th>
<th>N</th>
<th>Mean</th>
<th>SD</th>
<th>Minimum</th>
<th>Maximum</th>
<th>P value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Machined type</td>
<td>12</td>
<td>13.42</td>
<td>0.63</td>
<td>12.50</td>
<td>14.50</td>
<td>0.2873</td>
</tr>
<tr>
<td>Laser-treated type</td>
<td>12</td>
<td>13.08</td>
<td>0.85</td>
<td>12.00</td>
<td>14.50</td>
<td></td>
</tr>
</tbody>
</table>

SD indicates standard deviation.

Table 3. Comparison of the mean values of three-dimensional bone-implant contact (BIC), indicated by bone volume/total volume (BV/TV) and two-dimensional BIC.

<table>
<thead>
<tr>
<th>Group</th>
<th>3D BIC</th>
<th>2D BIC</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>N</td>
<td>Mean (%)</td>
</tr>
<tr>
<td>Machined type</td>
<td>12</td>
<td>46.2</td>
</tr>
<tr>
<td>Laser-treated type</td>
<td>12</td>
<td>46.0</td>
</tr>
</tbody>
</table>

SD indicates standard deviation.
Conclusion

1. Surface roughness was increased more than three times due to melting and solidification on laser-treated surface. A typical regular striation pattern of microporosity was not noticed on SEM images due to extensive melting.

2. Despite significantly increased roughness of laser-treated MSIs, compared with machined ones, laser treatment does not increase peak torque at fracture. Therefore, improvement of strength of MSI is not the indication for laser treatment of MSIs.

3. For secondary stability of SS MSIs, there was no significant difference between machined and laser-treated MSIs. Despite the mild mobility of some machined SS MSIs, all SS MSIs survived for 8 weeks under constant force (200 g) application. The BV/TV of 3D analysis can be considered a useful measurement to determine the bone-implant attachment of MSIs.

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