Bone response to unloaded and loaded titanium implants with a sandblasted and acid-etched surface: A histometric study in the canine mandible

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Abstract: Many dental clinical implant studies have focused on the success of endosseous implants with a variety of surface characteristics. Most of the surface alterations have been aimed at achieving greater bone-to-implant contact as determined histometrically at the light microscopic level. A previous investigation in non-oral bone under short-term healing periods (3 and 6 weeks) indicated that a sandblasted and acid-etched titanium (SLA) implant had a greater bone-to-implant contact than did a comparably-shaped implant with a titanium plasma-sprayed (TPS) surface. In this canine mandible study, nonsubmerged implants with a SLA surface were compared to TPS-coated implants under loaded and nonloaded conditions for up to 15 months. Six foxhound dogs had 69 implants placed in an alternating pattern with six implants placed bilaterally in each dog. Gold crowns that mimicked the natural occlusion were fabricated for four dogs. Histometric analysis of bone contact with the implants was made for two dogs after 3 months of healing (unloaded group), 6 months of healing (3 months loaded), and after 15 months of healing (12 months loaded). The SLA implants had a significantly higher ($p < 0.001$) percentage of bone-to-implant contact than did the TPS implants after 3 months of healing (72.33 ± 7.16 versus 52.15 ± 9.19; mean ± SD). After 3 months of loading (6 months of healing) no significant difference was found between the SLA and TPS surfaced implants (68.21 ± 10.44 and 78.18 ± 6.81, respectively). After 12 months of loading (15 months of healing) the SLA implants had a significantly greater percentage ($p < 0.001$) of bone-to-implant contact than did the TPS implants (71.68 ± 6.64 and 58.88 ± 4.62, respectively). No qualitative differences in bone tissue were observed between the two groups of implants nor was there any difference between the implants at the clinical level. These results are consistent with earlier studies on SLA implants and suggest that this surface promotes greater osseous contact at earlier time points compared to TPS-coated implants. © 1998 John Wiley & Sons, Inc. *J Biomed Mater Res,* **40,** 1–11, 1998.

Key words: surface characteristics; titanium implants; osseointegration; bone response; loaded implants

INTRODUCTION

In the past 30 years, the use of endosseous dental implants anchored in the jaws with direct bone-to-implant contacts has become a predictable and widely accepted treatment modality for fully and partially edentulous patients. This progress is based on fundamental studies conducted by Bränemark et al.[1,2] and Schroeder et al.[3–5] using commercially pure titanium implants in animal models. These studies showed that implant anchorage with direct bone contact can be achieved if certain surgical principles are followed. This type of implant anchorage is often termed osseointegration[2] or functional ankylosis.[5]

In recent years, attempts have been made to improve bone anchorage of dental implants. Thomas and Cook (1985)[6] examined the variables that influenced the apposition of bone to an implant surface. Of 12 parameters studied, only surface characteristics had a significant effect on the integration of the implant. This observation has been confirmed in a histometric study by Buser et al. (1991)[7] that showed a positive correlation between the percentage of bone-to-implant contacts and the roughness values of five different titanium surfaces tested. Among those, a sandblasted

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and acid-etched titanium implant revealed the best bone apposition to the implant surface, with 52% and 58% of bone-to-implant contacts after 3 and 6 weeks of healing, respectively. However, this study was carried out in long bones of miniature pigs and evaluated only short-term healing periods of unloaded experimental implants.

The purpose of the present study was to further evaluate this sandblasted and acid-etched titanium implant surface with a histometric study in the canine mandible. Short- and long-term evaluations were performed of implants restored with fixed partial dentures to study this titanium surface under unloaded and loaded conditions. The histometric analysis included similar measurements made on titanium implants with a titanium plasma-sprayed (TPS) surface. This microporous titanium surface, which has been utilized successfully in implant dentistry for more than 20 years, served as a control.[8] The radiographic[9] and soft tissue[10] analyses of these implants were reported in separate publications.

MATERIALS AND METHODS

Implant design and surfaces

Nonsubmerged commercially pure titanium implants with a hollow-screw design were utilized in this study (Institut Straumann AG, Waldenburg/BL, Switzerland). The outer diameter was 4.1 mm, and the total length measured 9 mm. The endosseous portion of the implants was 6 mm in length and had a large-grit (250–500 μm corundum grit) sandblasted and HCl/H_2SO_4 acid-etched surface (SLA). The SLA implant has a proprietary surface with the following treatment steps. The surface to be treated is sandblasted with 250–500 micron corundum, washed in an ultrasonic deionized water bath, and dried. The surface then is acid etched in a hot hydrochloric acid–sulfuric acid mixture, followed by thorough rinsing in deionized water before drying in hot air. The sandblasting produces a macroroughness onto which the acid-etch process superimposes a microroughness (Fig. 1). The control was an implant with the same shape, but it had a titanium plasma-sprayed (TPS) surface with typical roughness and porosity values of 30–50 μm (Fig. 2).

Surgical procedures: Extraction

Details of the different surgical and prosthetic procedures were described in a previous publication reporting radiographic results.[9] Therefore the clinical steps are only summarized here. Tooth extractions were performed in six male, laboratory-bred American foxhounds. They were approximately 2 years of age and had a body weight of about 30–35 kg. All four premolars (P1–P4) and the first molar (M1) carefully were extracted. Adaptation of the wound margins was achieved with interrupted sutures.

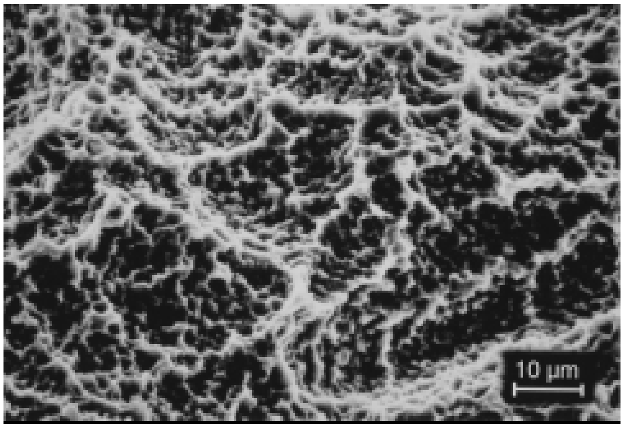

Figure 1. Electron micrograph of sandblasted and acid etched surface (SLA). A roughness value of 2–4 μm is superimposed on a roughness of 18–23 μm. Original magnification X1,700; bar = 10 μm.

Surgical procedures: Implant placement

Endosseous, nonsubmerged titanium implants were inserted after a healing period of 3 months [Fig. 3(A)]. A crestal incision was made, maximizing keratinized mucosa on each side of the incision. Mucoperiosteal flaps were elevated on the lingual and buccal aspects of the alveolar ridges and sutures were used to retract the flaps. Following careful flattening of the alveolar crest, six implants were inserted on each side of the mandible. According to a randomized starting selection, three test and three control implants were placed in an alternating manner per side. Healing screws were placed on top of the implants. In this fashion, no implant type had a biased position in the arch. Due to the narrowing of the ridge in the mesial area of the edentulous ridge, three out of the possible 72 implants could not

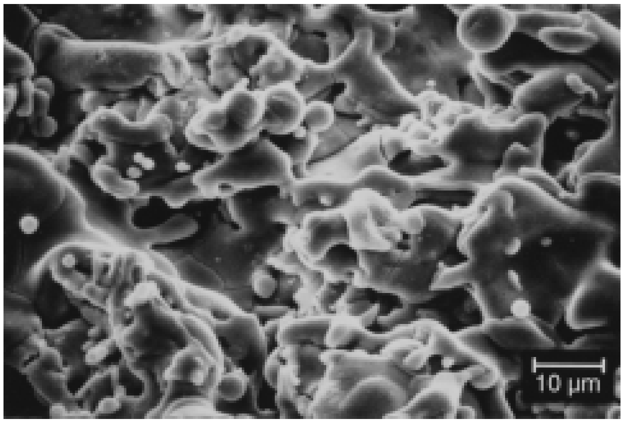

Figure 2. Electron micrograph of titanium plasma-sprayed surface. Typical roughness and porosity values with this surface are 30–50 μm. Original magnification X1,700; bar = 10 μm.

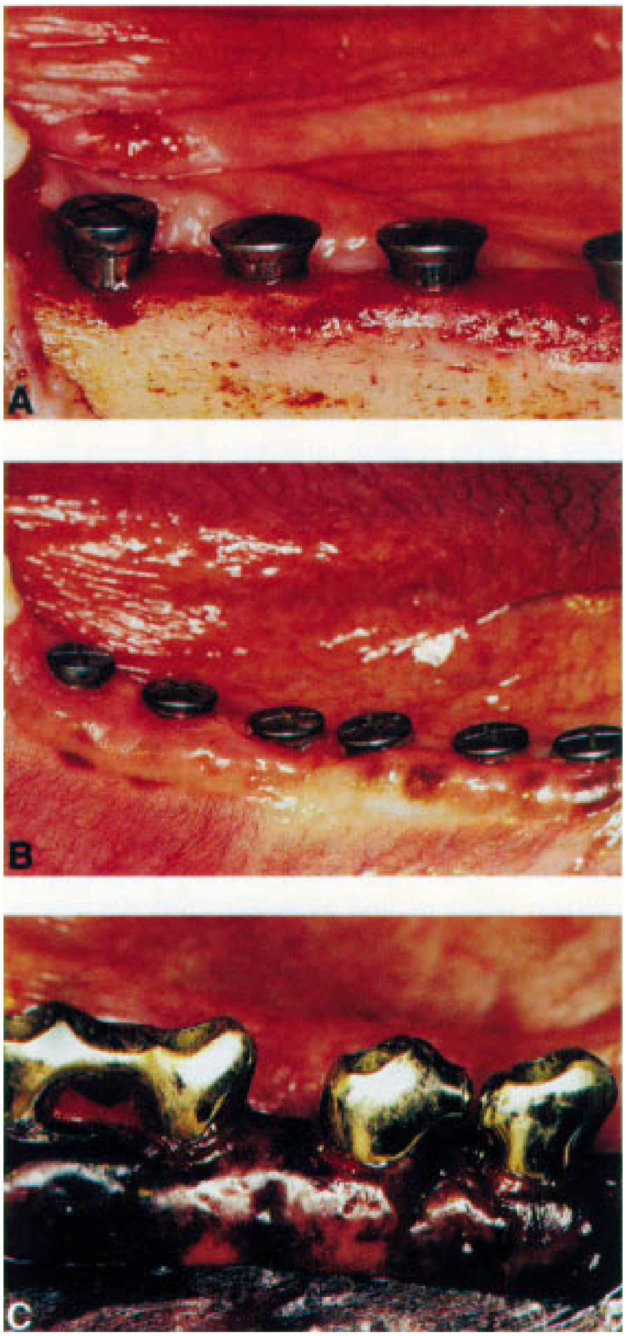

Figure 3. (A) Clinical view of implants in mandibular alveolar bone. The rough portion of the implant is embedded in bone while the smooth portion and cover screw protrude coronal to the alveolar crest. The alveolar crest has been flattened prior to implant placement. (B) Clinical postoperative view taken during first 3-month healing period after implant placement prior to implant restoration. Three times weekly oral hygiene was performed. TPS and SLA implants were placed in an alternating fashion, but the type of implant cannot be clinically distinguished. (C) Clinical view of four implant restorations. Most premolars (P2–P4) in the dog are bifurcated; therefore an effort was made to connect the restorations in this area (left side of picture). In some cases, however, a passive fit was assured only after a sectioning of the crowns (right side of picture).

be inserted, resulting in a total of 69 inserted implants. Subsequently, wound closure was achieved with interrupted sutures following close adaptation of the wound margins to the implant posts [Fig. 3(B)]. Postsurgically, a soft diet was utilized for the duration of the study.

Prosthetic reconstruction

Four of the six dogs constituted the loaded implant groups B and C (Fig. 4) since the implants were restored with fixed partial dentures. The restorations were placed 3 months after implant insertion [Fig. 3(C)]. They were evaluated and adjusted in the mouth to assure that the crowns were either out of occlusion or had only light contact since most premolars are not in occlusion in the foxhound. The occlusion was maintained as naturally as possible by taking models of both dental arches of each dog prior to extraction and duplicating each dog's occlusion.

Sacrifice and histologic preparation

Two of the six dogs (group A) constituted the unloaded (meaning no crowns were attached) implant group (Fig. 4) and were sacrificed after a healing period of 3 months. The other four dogs were sacrificed after loading. Two dogs were sacrificed after 3 months of loading (group B) and two dogs after 12 months of loading (group C). Harvested mandibles were immersed in a solution of 4% formaldehyde combined with 1% $CaCl_2$ for histologic preparation and analysis.[11] A radiograph was made of each specimen. The specimens were dehydrated and embedded in methylmethacrylate. Undecalcified sections of ~500 μm thickness were obtained using a low-speed diamond saw with coolant. First, two carefully oriented axial sections in the buccal–lingual plane were obtained for each implant. The remaining two parts of the block then were glued together and cut in a horizontal plane. This resulted in an optimal harvest of orthograde sections through the interface. In general, six to eight sections with a final thickness of ≈80 μm and stained

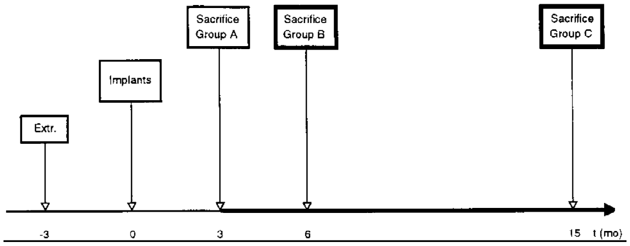

Figure 4. Study design time line. Two dogs (group A) had implants placed for 3 months and they constituted the unloaded implant group. Two dogs (group B) had implants placed and restorations for 3 months, and they constituted the 6 month healed, 3 month loaded group. The last two dogs (group C) had implants placed and restorations for 12 months, and they constituted the 15 month healed, 12 month loaded group.

with toluidine blue and basic fuchsin could be evaluated morphometrically.

Histometric analysis

The histometric measurements were carried out along the implant surface exposed to bone by means of a routinely used histometric technique with intersection counts at a microscopic magnification of 250X.[12] A square test grid, comprised of two sets of six straight, parallel test lines, was used for intersection counts. The lines were projected into the microscopic field with a special attachment. The number of hits were converted into percentage of contact surface with bone using the following categories: (a) primary bone contact, which was established during the insertion of the implants between the implant surface and the existing parent bone, (b) secondary bone contact, which was established by bony on-growth during the healing or remodeling phase, and (c) total bone contact, which was the sum of the primary and secondary bone contacts.

Statistical analysis

First, all data were analyzed by descriptive methods using box and QQ plots (Systat 5.2, Systat, Inc. Evanston, IL, USA). As they were not normally distributed, Kruskal–Wallis oneway analysis of variance and the Mann–Whitney U test were used. In case of multiple comparisons, the level of significance was corrected using the Bonferroni theorem.[13]

RESULTS

Clinical findings

Sixty-nine implants were placed in the mandibles without complications. No significant inflammation was found in the tissues around the implants with either the SLA or TPS surface during the study period. In fact, clinically it was impossible to distinguish the two types of implants. At the time of restoration, 3 months after implant placement, all the implants demonstrated ankylotic stability and the surrounding soft tissues were clinically healthy. During the loading period, none of the restored implants developed clinical signs of periimplant infection. At sacrifice, all the implants were considered successfully integrated. These clinical findings were confirmed with longitudinal standardized periapical radiographs that demonstrated no evidence for peri-implant radiolucencies.[9]

General histologic aspects of the implant sites

In axial buccolingual sections, the coronal part of the implant was surrounded by oral mucosa. The smooth surface of this portion was in contact with different soft tissue compartments, as described earlier.[10] The present study concentrated on the rough implant surface that was exposed to mandibular bone. Depending on the anatomical configuration of the mandible, the implant made contact with cortical or cancellous bone, or with bone marrow [Fig. 5(A–C)]. In addition, the placement of the implants during insertion modified the amount of contact with these various bone compartments. Besides implants that were almost ideally centered [Fig. 5(C)], implants may have been located eccentrically towards the lingual [Fig. 5(A)] or buccal side [Fig. 5(B)]. In view of the final bony incorporation of the different parts of the implant, location is more important the longer the implant is *in situ.*

Histology of the bone–implant interface

Primary contact along the bone–implant interface depends on congruency and press-fitting during implantation. Primary contact in cortical areas often was accompanied by compression of the bony lamellae and by the appearance of microfractures [Fig. 6(A)]. Interruption and compression of vessels in the Haversian canals causes cell death in the intracortical (Haversian) envelope and avascular cortical areas. Remodeling was in full progress in the nonloaded animals as characterized by disintegrating osteocytes and empty lacunae. Similar traces of mechanical alterations also were seen in cancellous areas in sites where direct contact between implant and trabeculae was forced by compression [Fig. 6(B)]. The consistency of the almost identical values in both SLA and TPS groups at 3 months of healing underlines the precision of fit achieved by the inherent accuracy of the instrumentation and surgical placement technique.

Secondary bone–implant contact is achieved either by bony ongrowth or bone remodeling. In cortical areas, the walls of the bore holes produced at surgery often are detectable. The gap between a wall and a screw thread is about 200 μm wide and partially or completely filled with new bone at 3 months [Fig. 7(A)]. Bone ongrowth also spreads out in sites with trabecular bone contact and even along surface areas that are lined by bone marrow only [Fig. 7(B)]. There it often appears as a bony coating less than 50 μm wide and apparently without any mechanical function. Load transfer, on the contrary, is in the cancellous areas as reflected by the formation of bony anchors [Fig. 7(C)]. Thereby the preexisting and partially devitalized trabeculae are reinforced by deposition of thick layers of lamellar bone, and firm contact with the

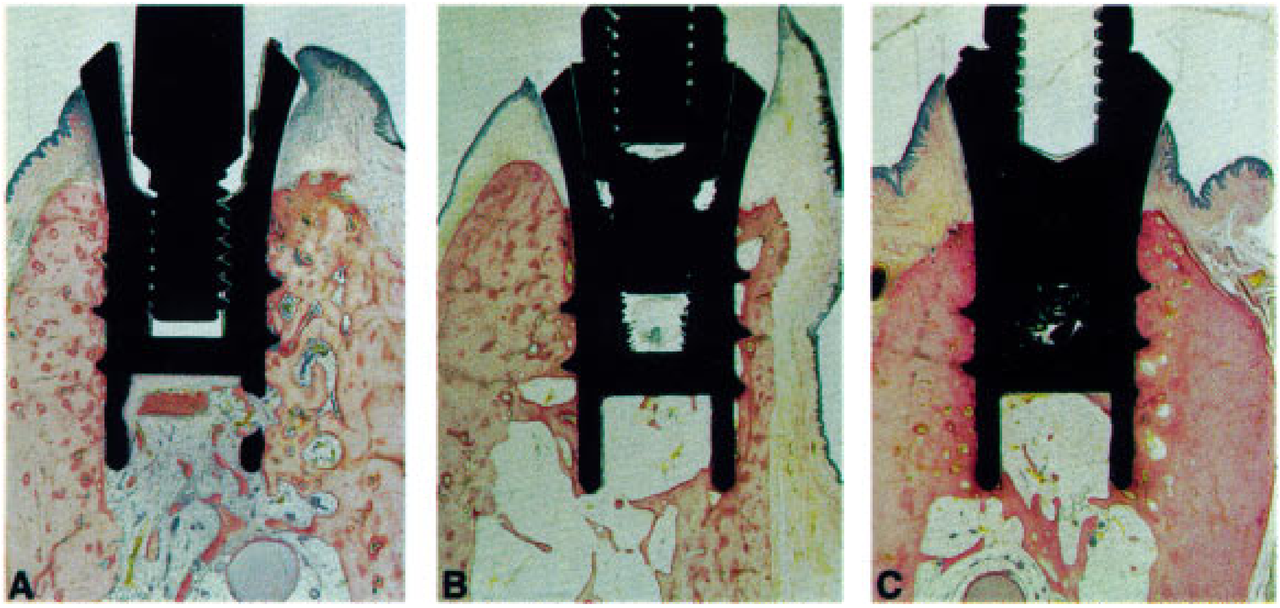

Figure 5. Low power micrographs of TPS-coated implants at 3 (A), 6 (B), and 15 (C) months (original magnification X25). The buccal wall of the implant site always faces to the right. A is somewhat displaced lingually, B buccally, whereas C is well centered. In general, the middle third of the implant is encompassed by cortical bone, the apical third by cancellous bone and marrow space. Perfect osseointegration is found in the nonloaded specimens at 3 months (A) and fully maintained throughout the 3-(B) and 12-(C) month loading period.

implant surface is established by the formation of bony bridges. The rather massive construct, with wall thicknesses of more than 200 μm, are metabolized (nourished) by vessels entombed in primary vascular canals or by vessels in secondary Haversian-like canals that result from remodeling.

Remodeling of the implant site

Cortical areas, presumably the sites damaged or devitalized by press-fitting, are subject to Haversian remodeling, that is to formation of resorption cavities or

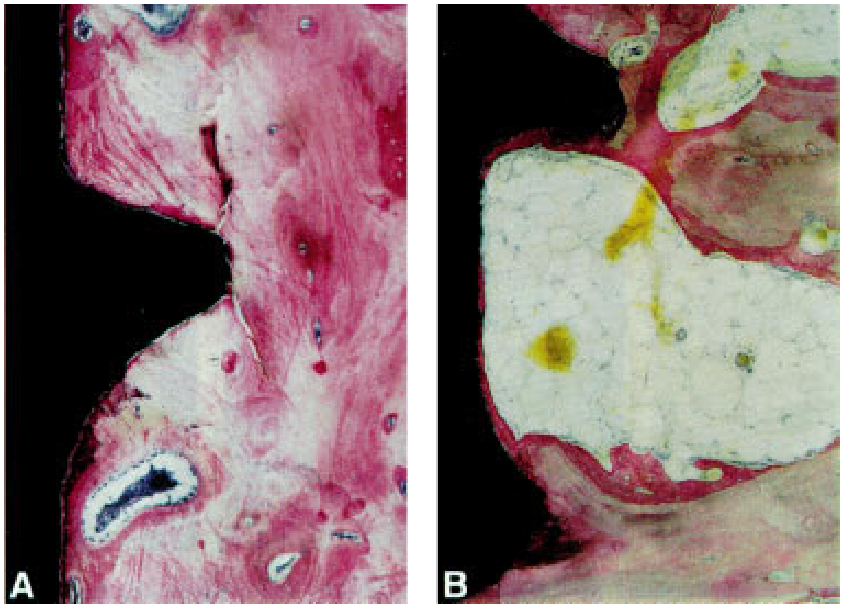

Figure 6. Primary contact with cortical (A, original magnification X25) and cancellous (B, original magnification X20) bone. These and all the following figures are oriented with the metal on the left side. Note the structural alterations caused by compression in the cortical bone as well as in the contact area with the lower trabeculum in B. Remodeling in the cortical area has started and bony ongrowth has enlarged the contact interface with the spongiosa.

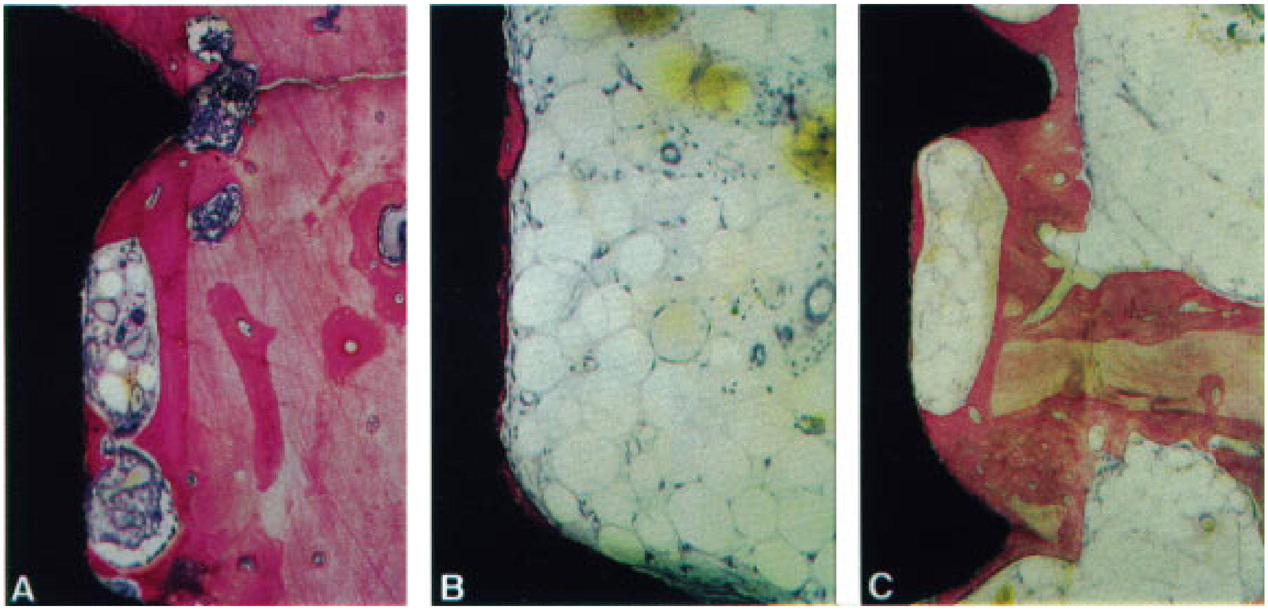

Figure 7. Secondary bone–implant contact by bone apposition. (A) A straight, vertical cement line marks the wall of the bore hole. The thread already is filled with new, more intensely stained bone. Original magnification X25; (B) A 20–40 μm thin bony "coat" is deposited upon a SLA surface exposed to red bone marrow. Original magnification X50; (C) Secondary contact with spongious bone is achieved by formation of bony anchors. Pristine (original) trabeculae appear brighter (higher mineral content) than the appositionally formed bone that is deposited on their surface and bridges the gap to the titanium surface. Bone layers thicker than 200 μm enclose vascular canals. Original magnification X20.

resorption canals followed by deposition of lamellar bone and completion of secondary osteons [Fig. 8(A)]. Remodeling is fully occurring already in the non-loaded group at 3 months and continues throughout the 3- and 12-month loading periods [Fig. 8(B,C)] although at this time the substitution of the necrotic areas and initial bone lesions is almost completed. In all groups there is a gradient in remodeling intensity

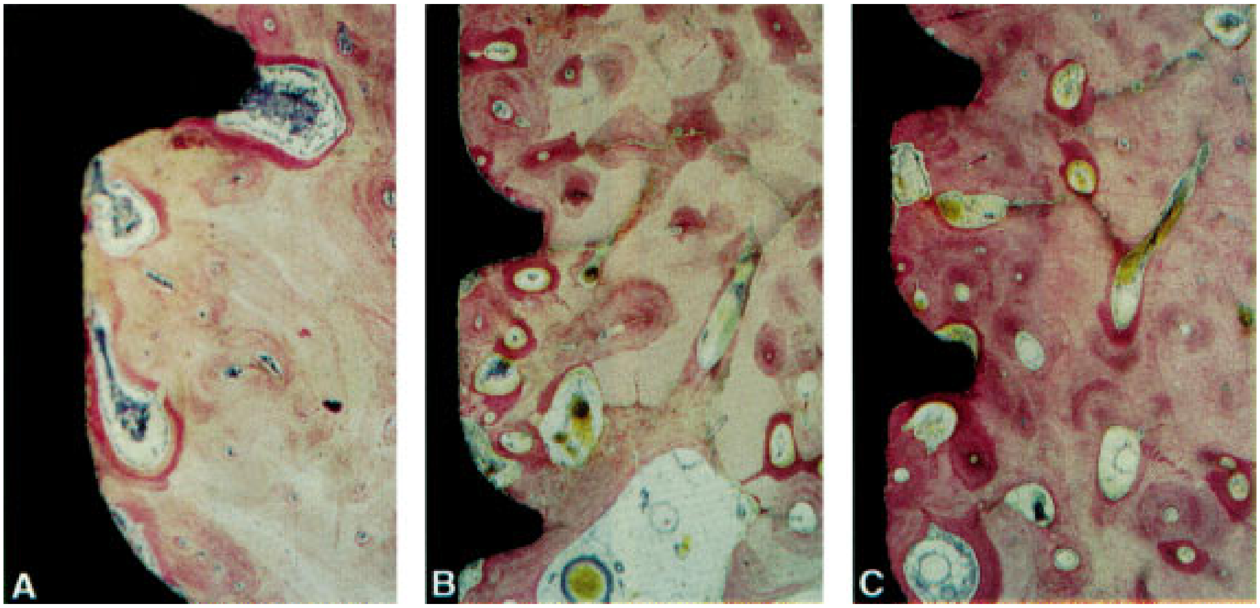

Figure 8. Secondary bone–implant contact resulting from cortical bone remodeling. (A) At 3 months, three cortical bone remodeling units are found in an early stage of bone formation. Original magnification X25; (B) At 6 months, remodeling has replaced large cortical areas by completed or still forming secondary osteons. Note the high remodeling activity in the vicinity of the thread. Original magnification X12.5; (C) Cortical remodeling in the implant site is still active at 15 months, i.e., after the 12-month loading period. During the whole period of bone substitution, an almost continuous bone–implant contact is perfectly maintained, indicating a perfect osseointegration. Original magnification X12.5.

that declines from the contact interface towards the periphery (or periosteal) surface [Fig. 8(B,C)].

Fine structure of the bone–implant interface

The two types of rough titanium surfaces tested in this experiment have provoked an identical reaction: bone is deposited directly upon the metal without any interposition of other tissue components detectable at the level of the light microscope [Fig. 9(A,B)]. This leads to a perfect congruency between the metal and the newly formed, fully mineralized bone.

TPS coating results in a rougher surface than the SLA treatment (sandblasting and acid etching). Microprotrusions and microundercuts characterize the microporous TPS surface, and mineralized bone penetrates between the protrusions. On the other hand, superficially attached particles can get loose and accumulate in the adjacent marrow tissue. The microprotrusions and microundercuts are somewhat less prominent in the SLA specimens.

The bone–implant interface is not always ideally preserved. Gaps of different width separate the bone from the titanium surface, but full congruency always is shown between the two parts [Fig. 9(C)]. Furthermore, the same sites often are found in perfect contact in adjacent sections. This proves that the bone mechanically was detached from the metal during processing and must be judged as contact sites in the morphometric evaluation.

Histometric results

The histometric analysis demonstrated quantitative differences between SLA and TPS surfaces (Table I, II, Fig. 10). After 3 months of healing, the unloaded SLA implants demonstrated a significantly greater amount ($p < 0.001$) of bone-to-implant contact with 72.33 ± 7.16% (mean ± SD) whereas the TPS surface had 52.15 ± 9.19% bone contact (Table II). At this evaluation period, the proportion of primary bone contact still was rather prominent (20.97 ± 7.62% for SLA and 18.45 ± 9.21% for TPS surfaces) compared to secondary bone contact with newly formed bone in intimate contact with the implant surface.

By 6 months (3 months of loading), the percentage of bone-to-implant contact significantly increased around the TPS implants, resulting in 78.18 ± 6.81% compared with 68.21 ± 10.44% for SLA implants (Table I). However, there was no statistically significant difference between the TPS and SLA implants (Table II). At this time point, primary contact had decreased significantly ($p < 0.001$) to 7.93 ± 3.38% for SLA-surfaced implants and 8.99 ± 4.53% ($p < 0.05$) for the TPS implants (Table I). Consequently, secondary bone contacts had increased accordingly.

At 15 months postimplant placement (12 months of loading), SLA implants again showed a significantly greater amount ($p < 0.001$) of bone-to-implant contact with an overall 71.68 ± 6.64% compared to 58.88 ± 4.62% for TPS implants (Table II). This value for TPS implants reflects a decrease in percent contact area from the 3-months loaded group, which had 78.18%. Primary bone contacts were reduced further to 5.28 ± 2.55% for SLA implants while the control TPS implants demonstrated 5.06 ± 4.74% primary bone contacts. Secondary bone contacts were significantly different ($p < 0.001$) between the two implant surfaces, with SLA implants having 66.40 ± 6.21% and TPS implants having 52.82 ± 7.95% (Table II).

DISCUSSION

In a previous study characterizing the contact of bone tissue to implant surfaces, Buser et al. (1991)[7] demonstrated that a sandblasted and acid-etched (SLA) implant surface achieved the greatest amount of bone contact of five different titanium surfaces in cancellous bone after 3 and 6 weeks of healing. This study was performed in long bones of miniature pigs. The present investigation in mandibular bone of fox hounds confirmed this observation since the SLA-surfaced implants achieved significantly more bone apposition at 3 months of healing than did control implants with a titanium plasma-sprayed (TPS) surface. Thus the present study documents for the first time that titanium implants with a SLA surface have superior bone apposition during healing compared with a TPS surface in mandibular bone. At a later time point of 3 months after loading, no statistical differences were observed between the two groups. The percentage of original bone contact with the implant (primary contact) decreased over time, which can be interpreted as a sign of on-going bone remodeling at the bone–implant interface. The fact that the amount of bone increased to a similar amount as that found on the SLA implants after 6 months of healing (3 months of loading) indicates that there is not an absolute difference in the bone growth on both surfaces, that is, bone growth on the TPS surface is, compared to SLA, only delayed. Most of the changes occurred in the secondary bone contact measurement, indicating an effect on bone remodeling (see references 14,15). It is possible that the differences between the TPS and SLA surfaces at 15 months is related to a more favorable osteophilic property of the SLA surface relative to the TPS surface. New bone deposition could occur on the SLA surface while in the case of TPS bone deposition

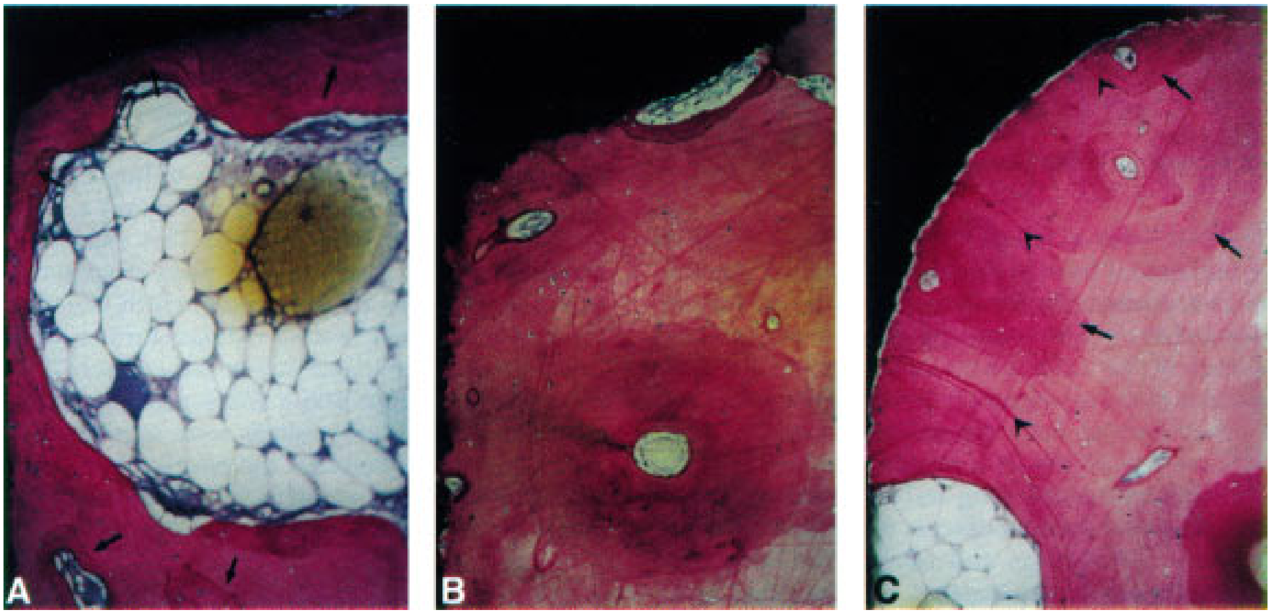

Figure 9. Microscopic structure of the bone–titanium interface. Original magnification X50. (A) SLA implant at 15 months showing a secondary contact site with perfect interdigitation between the rough surface and mature bone. Note the cement lines (arrow) separating several lamellar packets deposited by consecutive remodeling steps; (B) TPS-coated implant at 15 months. More pronounced irregularities of the surface but intimate contact with fully calcified bone matrix that is deposited even into the smallest clefts in between the plasma grains. Cortical bone remodeling resulted in secondary osteons, lined by cement lines and formed also in immediate contact with the metal. (C) SLA implant at 15 months. Small cracks separating the bone from the implant surface are typical artifacts. They are produced when the sections are glued under firm compression to the plastic slides. The perfect congruency between the two components and lack of any interpositioned tissue components proves the original direct contact of bone and implant. Note secondary osteons (arrows) in the contact area. Cement lines (arrowheads) running perpendicular to the implant surface indicate the pattern of lamellar bone deposition in an apical direction.

could be restricted to the bony walls of the resorption canal (Fig. 11, see p. 8).

The mechanism for the significantly greater percentage of bone contact at early healing periods around SLA-surfaced implants compared to TPS-surfaced implants is not known. Because the implant types had alternating positions in the mandibles, it is likely that

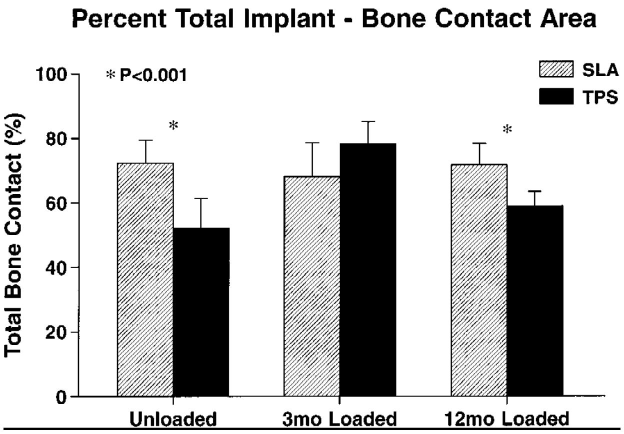

Figure 10. Percent total implant–bone contact area. Histogram comparing implants with a TPS surface to implants with a SLA surface over time. The bars reflect the total percent implant-to-bone contact area comprised of primary and secondary bone contact.

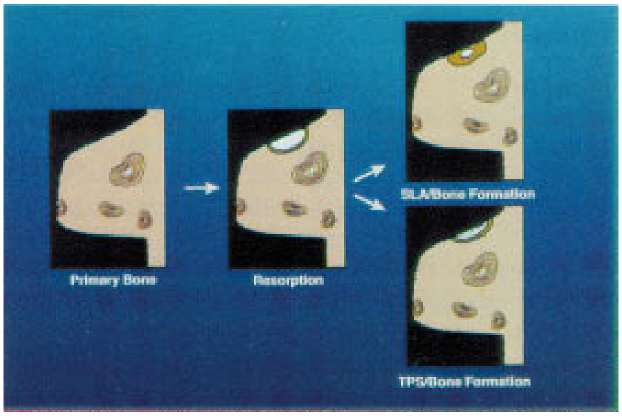

Figure 11. Schematic drawing of possible bone-to-implant contacts. (A) Area of primary bone contact established during insertion of the implant with the existing bone; (B) Resorption cavity formed along the implant–bone interface, which results in an implant surface devoid of tissue; (C) Secondary bone formation by bone remodeling units on a more ''osteophilic'' surface, such as the SLA surface, resulting in bone-to-implant contact; (D) On a relatively less ''osteophilic'' surface, such as TPS, the bone remodeling unit initially would result in bone formation only on the remodeling bone surface and only later on the implant surface.

TABLE I
Implant–Bone Contact*

Implant Type	Primary Bone Contact	Secondary Bone Contact	Total Bone Contact
SLA Implants			
Unloaded	§20.97 ± 7.62§	51.36 ± 10.18‡	72.33 ± 7.16
3-mo loaded	§7.93 ± 3.38§	60.28 ± 8.79‡	68.21 ± 10.44
12-mo loaded	5.28 ± 2.55§	66.40 ± 6.21‡	71.68 ± 6.64
TPS Implants			
Unloaded	†18.45 ± 9.21‡	§33.69 ± 7.42§	52.15 ± 9.19§
3-mo loaded	†8.99 ± 4.53‡	§69.19 ± 6.12§	78.18 ± 6.81§
12-mo loaded	5.06 ± 4.74‡	§52.82 ± 7.95§	58.88 ± 4.62§

*Expressed as a percentage, mean ± SD, n = 24; significance: $^{\dagger}P < 0.05$; $^{\ddagger}P < 0.01$; $^{\S}P < 0.001$.

the difference between the SLA and TPS implants, as regards bone apposition, was due to the surface characteristics alone.[16–20] *In vitro* evidence supports this hypothesis. Martin et al. (1995)[21] have demonstrated in tissue culture that alkaline phosphatase activity in osteoblast-like cells is greater on SLA surfaces than on TPS surfaces. Since alkaline phosphatase activity is an indication of bone cell maturation, these results suggest that the bone cells in contact with the SLA surface would be more differentiated, that is more like bone-forming cells, than would be bone cells in contact with TPS surfaces. Furthermore, the *in vitro* results demonstrated that more bone cells attached and proliferated on the SLA surface than on the TPS surface, again consistent with the *in vivo* histological findings on bone-to-implant contact described in this paper. These results demonstrated a significant advantage for the SLA surface at early healing periods and also could explain the findings of Kirsch and Donath (1984)[22] that titanium disks with a microporous TPS surface demonstrate significantly faster bone-to-implant contacts compared to those of smooth titanium surfaces.

This study evaluated nonsubmerged titanium implants with the same macroscopic shape but with two different surfaces and showed that both were clinically and histologically stable under unloaded and loaded conditions in the canine mandible. All 69 inserted implants achieved and maintained successful tissue integration up to 15 months following implant placement, demonstrating ankylotic stability without clinical signs of peri-implant infections. The radiographic evaluation utilizing longitudinal standardized periapical radiographs[9] confirmed the clinical findings of these implants with functional ankylosis since none of the 69 implants demonstrated a continuous peri-implant radiolucency. Radiographic assessment of crestal bone showed significantly less bone loss for SLA implants at the 3-month healing period compared to the TPS implants. These results are consistent with the histological findings of bone-to-implant contact described here since the SLA-surfaced implant demonstrated significantly greater bone apposition histologically at 3 months of healing. Furthermore, the radiographic results indicated that crestal bone levels stabilized after loading, again consistent with the described histological findings of bone-to-implant contact during the loading period. Additionally, Cochran et al.[10] have shown that the dimensions of the soft tissues immediately coronal to the rough implant surface have a relatively fixed biologic width similar to the same findings around teeth.[23] These findings compare well to results in human patients with nonsubmerged titanium implants with a TPS surface. Buser et al. (1990)[24] reported 1-year radiographic data on 100 ITI implants with a TPS surface in partially edentulous patients with a mean crestal bone loss of approximately 0.8 mm and a pocket depth of 2.74 mm.

In summary, the histological analysis of 69 titanium implants with TPS or SLA surfaces demonstrated that both implant types achieved and maintained successful tissue integration in the canine mandible, with functional ankylosis up to 15 months under unloaded and loaded conditions. Earlier healing periods (3 months healed, unloaded) and long-term loaded periods (12 months) indicate an advantage for the implants with an SLA surface as a significantly greater percentage of bone-to-implant contact was observed for this implant. Implants loaded for 3 months showed no differences in bone-to-implant contact between the two tested titanium surfaces. Loading the SLA implants for up to 12 months did not change the percentage of bone-to-implant contact relative to 3 months of

TABLE II
Comparison of Implant Types*

Implant Type	SLA		TPS
Unloaded Implants			
Primary bone contact	20.97 ± 7.62	ns	18.45 ± 9.21
Secondary bone contact	51.36 ± 10.18	†	33.69 ± 7.42
Total bone contact	72.33 ± 7.16	†	52.15 ± 9.19
3-Month Loaded Implants			
Primary bone contact	7.93 ± 3.39	ns	8.99 ± 4.53
Secondary bone contact	60.28 ± 8.79	ns	69.19 ± 6.12
Total bone contact	68.21 ± 10.44	ns	78.18 ± 6.81
12-Month Loaded Implants			
Primary bone contact	5.28 ± 2.55	ns	6.06 ± 4.74
Secondary bone contact	66.40 ± 6.21	†	52.82 ± 7.95
Total bone contact	71.68 ± 6.64	†	58.88 ± 4.62

*Expressed as a percentage, mean ± SD, n = 24; ns = no significance; $^{\dagger}p < 0.001$.

loading. This study has provided significant new data on implants with an SLA surface. However, several questions remain: What are the functional consequences of the increased bone-to-implant contact at earlier healing times, that is, is more force required to remove SLA-surfaced implants at early healing periods than is required to remove TPS-surfaced implants at that time? Additionally, can the advantages of this surface be demonstrated in humans? This latter question is particularly important due to the fact that if these results are observed in humans, then the possibility exists that the time may be significantly reduced between implant placement and implant restoration, for example from 3 months to 2 months. Therefore, this study has confirmed and extended encouraging results for the SLA surface of previous studies in long bones[7,19] and *in vitro.*[21] Furthermore, this study confirmed the potential of the titanium SLA surface to become a valuable, or even superior, alternative to the clinically well documented TPS surface[24–28] in implant dentistry.

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