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Chapter 7

The Role of Hydrophilic Sandblasted Titanium and Laser Microgrooved Zirconia Surfaces in Dental Implant Treatment

Aleksa Markovic

Additional information is available at the end of the chapter

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Abstract

Dental implant surface modifications affect surface roughness, chemistry, topography, and consequently influence biological bone response. Current surface treatments are directed toward increased hydrophilicity and wettability of dental surfaces that allow earlier implant loading due to accelerated osseointegration. This is clinically reflected in increased implant stability and mainteined crestal bone level. Further modification includes microgrooving of zirconia implants by femtosecond laser ablation. Favorable initial results encourage further clinical assessment of this microgrooved zirconia implants.

Keywords: Dental implant surface, Femtosecond laser, Hydrophilicity, Titanium, Zirconia

1. Introduction

Dental implant surfaces can be modified using several additive and subtractive techniques. Additive techniques involve impregnation and coating. In contrast to impregnation, when chemical agent is integrated into the core material (e.g., fluoride ions incorporated to titanium surface or calcium phosphate crystals within TiO_2 layer), the coating is addition of an agent of various thicknesses superficially on the surface of core material [1].

The subtractive techniques imply removal of the layer of core material or plastic deformation of the superficial surface. This is achieved through mechanical or chemical treatments. Mechanical methods used for surface alteration are grinding, blasting, and machining. Sand,

hydroxyapatite, TiO_2, and Al_2O_3 particles are usually used for grit blasting. Grit blasting is always followed by an acid etching to remove the residual blasting particles as well as to smooth out sharp peaks and to provide roughness that would promote protein adhesion during early healing [1, 2]. Acid etching is a chemical method of surface modification that usually implies hydrofluoric, nitric, or sulfuric acid or their combinations [3].

Current modification of dental implant surfaces is based on the use of lasers. Their main applications are laser-assisted coatings and laser texturing. Laser pulses are used to evaporate the target materials which later condense on the substrate forming a thin coating. Their further role is in dental implant surface texturing in order to form three-dimensional structures on micrometer or submicrometer scales. Several laser sources such as Nd:YAG, CO_2, Excimer, and diode lasers have been examined for surface modifications [4].

Lasers are particularly useful for dental implants with complex surface geometry or for those made of material difficult to be removed. Dental implant surface modification by laser is a noncontact, clean and fast process with high precision [4]. Lasers overcome drawbacks of conventional mechanical and chemical surface modification techniques such as unreliable control of achieved roughness or inability of surface texturing. However, laser processing might be associated with microcracks and heat-affected zones [4, 5].

1.1. Surface topography

Scientific evidence from in vitro studies indicates that micro-topography of dental implant surfaces affects cellular behavior, mainly, proliferation, cell differentiation, and cell adhesion as well as the production of growth factors [6]. Microgrooves at implant surfaces direct cell spreading and cell alignment and define the orientation of ECM proteins. This directional movement of bone cells known as "contact guidance" contributes to bone–implant interlocking and thus provides favorable conditions for further healing events [7]. Microtextured surfaces suppress fibroblast spreading and growth preventing fibrous encapsulation of dental implants. Important parameters of surface roughness are average height deviation (Sa) and developed surface area ratio (Sdr) that indicates surface enlargement if a surface is flattened out. According to Sa values determined by optical interferometry, implant surfaces are considered as smooth with an Sa value of 0.5 μm; minimally rough surfaces with an Sa of 0.5–1 μm, moderately rough surfaces with Sa 1–2 μm, and rough surfaces with an Sa of 42 μm. Moderately rough surfaces with Sa 1.5 μm and Sdr of 50% promote the strongest bone response [2]. To mimick the architecture of natural bone that consists of nanosized hydroxiapatit and organic protein collagen, dental implant surfaces with nano-features have been introduced. Although initial data are promising, the effect of surface nanoroughness on biological response is still uncertain [3].

Surface topography is usually examined by scanning electron microscopy (SEM), light interferometry (LIF), and atomic force microscopy (AFM), whereas X-ray photoelectron spectroscopy (XPS), Auger electron spectroscopy (AES) and energy dispersive X-ray spectroscopy (EDX) provide information related to surface chemical composition [3, 8].

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SEM is the gold standard for characterization of dental implant topography at the micrometer level. For the characterization of nanotopography of dental implant surfaces, field emission-SEM with higher resolution is required [3].

LIF is an efficient optical tool for quantitative analysis of implant surfaces. This technique uses reflecting light as an optical stylus allowing easy access to even unapproachable parts of the implant flunks. Despite its high resolution in height direction, LIF is suitable for characterization of dental implant surfaces at a micrometer scale, because of limited spatial resolution [8].

AFM can be used to assess dental implant surface topography at nano-level. An atomic force microscope consists of a tip mounted on a cantilever. When tip scans a dental surface, interatomic forces between the tip and the sample surface displace the tip which results in the cantilever bending. Consequently, specialized software produces topographical image of the surface with atomic resolution based on data from detector regarding the laser beam reflected from the cantiliver. This tool has resolution at molecular level, but its usage is unreliable for certain level of surface roughness because their microtopography significantly interferes with the vertical piezoelectric AFM scanning probe [3, 8].

XPS also known as electron spectroscopy for chemical analysis (ESCA) determines what elements (except hydrogen and helium) and in which chemical state and quantity are present within the top 1–12 nm of the implant surface. This tool also provides information related to possible contamination on the surface or in the bulk of the sample as well as those related to the presence and thickness of layers of different materials within the top 12 nm of the implant surface. XPS spectra are obtained by irradiating a dental implant surface with a beam of X-rays and measuring the kinetic energy and number of electrons emitted from the top of the surface [3, 8].

AES provides quantitative elemental and chemical state analysis of dental implant surfaces with lateral spatial resolution of only 8 nm. Approximate depth resolution of AES is 5 nm. However, ion-sputtering used with Auger spectroscopy allows depth chemical profiling up to 100 nm, which is suitable for the characterization of coatings on implant surfaces or impregnation within a TiO_2 layer [3]. For the AES analysis, implant surface is excited with a finely focused electron beam while an electron energy analyzer measures the energy of Auger electrons emitted from the surface. Based on the kinetic energy and intensity of an Auger peak, elements from the implant surfaces are identified.

EDX is used for the elemental analysis or chemical characterization of a dental implant surface. It is based on the unique set of peaks on X-ray emission spectrum of each element. Coupled to SEM, EDX determine the elemental composition of structures observed with SEM down to the nanoscale [3].

1.2. Surface wettability

Important characteristic of dental implant surfaces is surface energy that dertermines wettability of surfaces. It is measured by liquid–solid contact angle (CA) which is an angle between the tangent line to a liquid drop's surface at the three-phase boundary, and the horizontal solid's surface [9]. There are two methods commonly used to assess CA of dental implant

surfaces: the sessile drop technique where CA of the droplet deposited by a syringe onto the sample surface is measured directly by goniometer or image analysis software and the second, tensiometry (Wilhelmy method) that indirectly measure CA according to the force exerted on the sample surface by the liquid, while sample surface attached to a force meter is vertically dipped into a pool of the probe liquid [10].

The CA ranges from 0° to 180° where CA lower than 90° designate surfaces as hydrophilic and CA very close to 0° as superhydrophilic. Dental implant surfaces with CA above 90° are considered hydrophobic, and those with CA above 150° are superhydrophobic [9]. Currently available dental implants are mainly hydrophobic [11]. Although optimal degree of wettability is not known, there is abundant scientific evidence that hydrophilic surfaces enhances early stages of osseointegration compared to hydrophobic ones [12–14].

Hydrophilicity of dental implant surfaces determines adhesion of proteins on the surface of placed implant, interaction of hard and soft tissue cells with implant surface, and consequently the rate of osseointegration [9]. Hydrophilic surfaces promotes superior adsorption and functional orientation of proteins from blood and interstitial fluids. Composition of the proteins adhered to the implant surface affects cell adhesion, morphology, and migration [15]. Hydrophilic dental implants favor osteoblastic differentiation of mesenchymal stem cells [16], enhance osteoblast maturation [17], produce an anti-inflammatory microenvironment [18], and increase the quantity and quality of mineralization [19]. These molecular and cellular events provide accelerated osseointegration of hydrophilic dental implants in contrast to hydrophobic which has been verified histomorphometricaly as increased bone-to-implant-contact (BIC) at very early point in healing [12–14].

Advantages of hydrophilic surfaces recognized in in vitro and in vivo studies on dental osseointegration have directed contemporary modifications of dental implant surfaces toward to greater hydrophilicity. Today, several methods of hydrophilizing dental implant surfaces are available including radio frequency glow discharge treatment, atmospheric pressure plasma, surface coating with crystalline TiO_2, and irradiation by UV-A as well as Ti surface with native oxide hydrophilized using higher energy UV-C rays [9]. Also, changes in dental implant surface roughness and chemistry affect hydrophilicity, which complicates the analysis of the independent effect of each of these surface characteristics on clinical behavior of available dental implants.

1.3. Clinical outcome of dental implant surfaces

Osseointegration of dental implants is clinically reflected in implant stability. Primary implant stability is a mechanical issue determined by bone quantity and quality, surgical technique, and implant macro-design, whereas secondary implant stability as a biological phenomenon indicates bone apposition and remodeling processes and it is influenced by conditions of implant surface [20, 21]. Contemporary implant surfaces accelerate osseointegration and provide conditions for early or even immediate implant loading if sufficient implant stability is achieved. Therefore, non-invasive, objective and quantitative tool for the assessment and monitoring of implant stability in clinical conditions is of great importance.

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Resonance frequency analysis (RFA) is a wireless system for the measurement of implant stability that includes a metal rod (a peg) screwed into the implant body and stimulated by magnetic pulses from a handheld computer. The result of a measurement is expressed as the implant stability quotient (ISQ) ranging from 1 (lowest stability) to 100 (highest stability) (**Figure 1**). ISQ values higher than 47 indicate stable implant [22]. The recommendations for immediate and early loading of single-implant crowns are ISQ 60–65 [23]. Implants with high ISQ values during the follow-up are successfully osseointegrated, whilst low and decreasing ISQ values may be a warning sign of ongoing implant failure [20, 21].

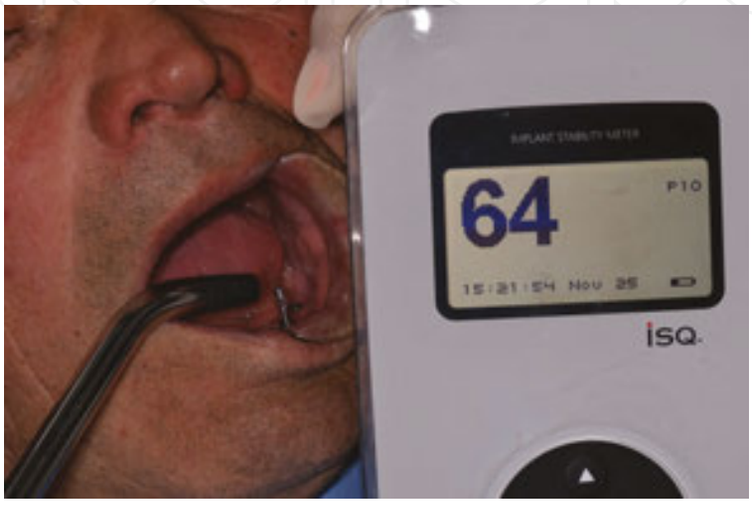

Figure 1. Resonance frequency analysis measurement using Ostell Mentor® device.

RFA is not suitable for the measurement of stability of one-piece dental implants and in such indication the use of the Periotest is recommended. The Periotest produces percussion of the implant and provides a stability number ranging from −8 to + 50, with the lower the Periotest value (PTV), the higher the stability (**Figure 2**) [24]. In the literature, different ranges of PTVs

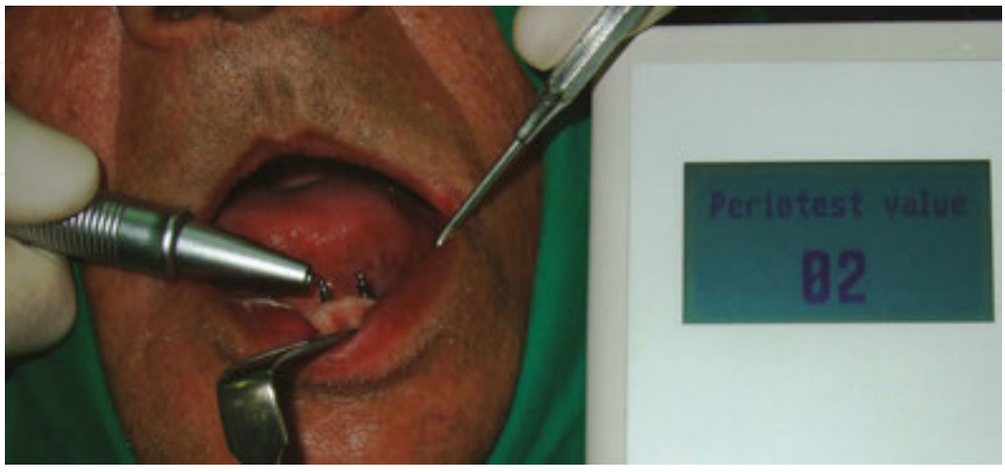

Figure 2. Measurement of implant stability of one-piece dental implants using Periotest.

for successfully osseointegrated dental implants have been reported (−9 to +9; −5 to +5; −7 to 0; −4 to −2; −4 to +2) [24–27].

Another important clinical parameter that reflects condition at implant–bone interface is change in crestal bone level. It is recommended to follow this parameter on retroalveolar radiographs obtained via long cone technique. This technique uses film holder that allows repeatability of tube orientation (**Figure 3**). Image analysis software is used for precise measurement of digitized radiographs following their calibration (**Figure 4**). Implant is considered successful with an crestal bone loss of 1.5 mm following 1 year of loading and subsequent loss of 0.2 per year [28].

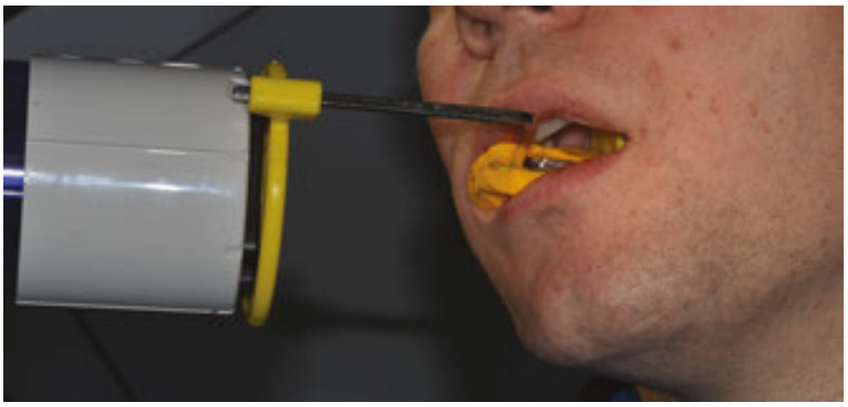

Figure 3. Obtaining radiographs using long cone technique. A plastic ring, connected to the film holder provided control of tube orientation.

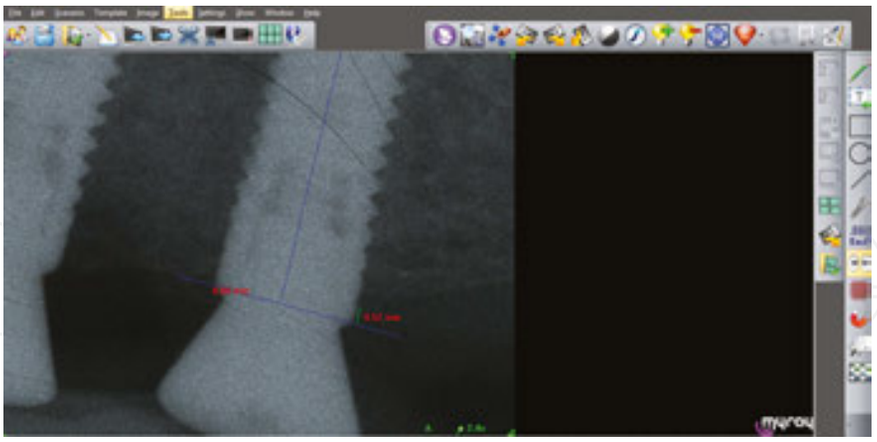

Figure 4. Image analysis software for crestal bone loss measurement.

2. Role of SLActive surface in dental implant treatment

SLActive dental implant surface (Institute Sraumann® AG, Basel, CH) is a hydrophilic with a sandblasted and acid-etched topography and was created as an aspiration to combine the

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advantages of surface roughness and hydrophilicity on implant osseointegration [14, 29]. This new surface is produced from the same cpTi alloy and subjected to the same roughening treatment with large grit size (250–500 μm) corundum sandblasting plus acid etching (H_2SO_4/HCl) as its predecessor SLA surface [30]. The only difference from its unmodified counterpart is that following acid-etching hydrophilic SLActive implant is rinsed under nitrogen protection and then stored in a sealed glass tube containing isotonic NaCl solution [29]. Such chemical modification provides hydrophilization of surface that is initially hydrophobic due to the microroughness, causing the air to be entrapped in the micropores, thus aggravating surface wetting [29]. Storage in NaCl solution allows prewetting of micropores and consequently faster wetting of implant surface [30]. CA of 0° designates SLActive dental implant surface as superhydrophilic in contrast to hydrophobic SLA surface with CA of 139.9° [9, 29, 30]. Another reason for reduced hydrophilicity of Ti implants is contamination due to air exposure [29]. However, cleaning under nitrogen protection and storage in NaCl solution prevents the adsorption of potential contaminants from the atmosphere onto the SLActive surface which is proven by the decreased carbon concentration [29, 30]. SLActive surface keeps hydrophilicity even after any drying which is important from a clinical point [29].

Another possible reason for improved biological response to SLActive surface compared with its unmodified counterpart SLA is the difference in microtopography and nanoroughness [31]. Although both surfaces have similar Sa value (1.78 and 1.75 μm for the SLA and SLActive, respectively), the SLActive surface has Sdr of 143% that is greater than Sdr of 97% for SLA. This difference indicates that SLActive has a much greater number of peaks/valleys across the surface compared with SLA [31, 32]. SLActive surface also exhibits nano-features with Sa value of 97 nm at the nanometer resolution level [31].

Biological effect of aforementioned improvements of SLActive implant surface comprises of enhanced osteoblastic differentiation [33, 34], improved angiogenesis [35] and reduced local inflammation and its associated osteoclastogenesis [36]. These cellular events provides stronger bone formation around SLActive implants compared with SLA during the early healing phase, particularly between the second and fourth weeks, while the difference disappears after the first 6 weeks [37]. Such features of SLActive implant surface indicate its possible clinical relevance in cases when faster implant loading is needed or when enhanced bone formation is desirable as in osteoporotic or irradiated bone and in diabetic patients.

Contemporary improvement of dental implant surfaces has allowed shift from a conventional loading protocol providing 3–6 months of undisturbed healing toward immediate (within 1 week) or early loading (between 1 week and 2 months) in selected patients when sufficient primary implant stability could be achieved [38, 39]. Clinical and radiographic outcomes of SLA and SLActive dental implants submitted to immediate or early occlusal loading is comparable to those of implants submitted to conventional loading protocols (3–6 months) [40]. SLActive implants loaded at 3 weeks after placement have survival rate of 95–98% following 1–3 years after placement [41, 42]. Hydrophilic and nanostructured SLActive implants are safe and predictable for immediate and early loading even in poor-quality bone [43, 44]. Early loaded SLA and SLActive implants achieves similar short- and long-term

survival rates, although SLActive implants have better stability and a reduced marginal bone loss at the loading stage [40, 45].

As much as 97.3% of SLActive implants placed in low-density bone achieves implant stability of at least 60–65 ISQ required for immediate or early implant loading (**Figure 5**) [23, 46]. The stability dip in the second postoperative week indicates that afterward, formative processes predominated over the resorptive one within bone remodeling. This result suggests that nanostructured and hydrophilic SLActive implant surface promotes enhanced bone formation during the early stage of osseointegration. It is important that even in this critical time point stability values did not fall below the threshold for early loading. Afterward, implant stability steadily increases over time.

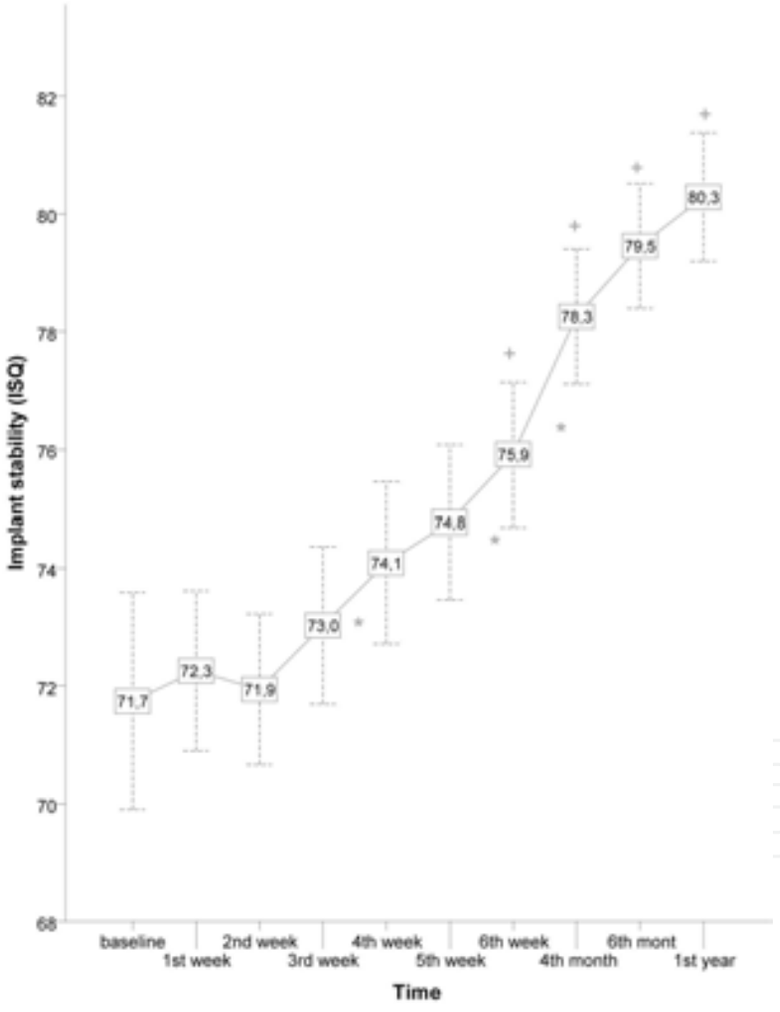

Figure 5. Stability of SLActive dental implants placed into low-density bone. Line represents mean, error bars represent 95% CI of mean. Asterisks indicate a statistically significant difference between two-consecutive weeks, whereas crosses indicate a statistically significant difference to baseline (implant placement) [46].

SLActive dental implants placed in low-density bone and early loaded (at week 6) are associated with mean bone loss of −0.4 1 ± 0.1 mm after 1 year that is in accordance with the

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acceptable 1 mm bone loss during the first year (**Figure 6**) [46]. These data suggest that SLActive dental implants predictably achieve and maintain successful tissue integration in low-density bone after undergoing an early loading protocol.

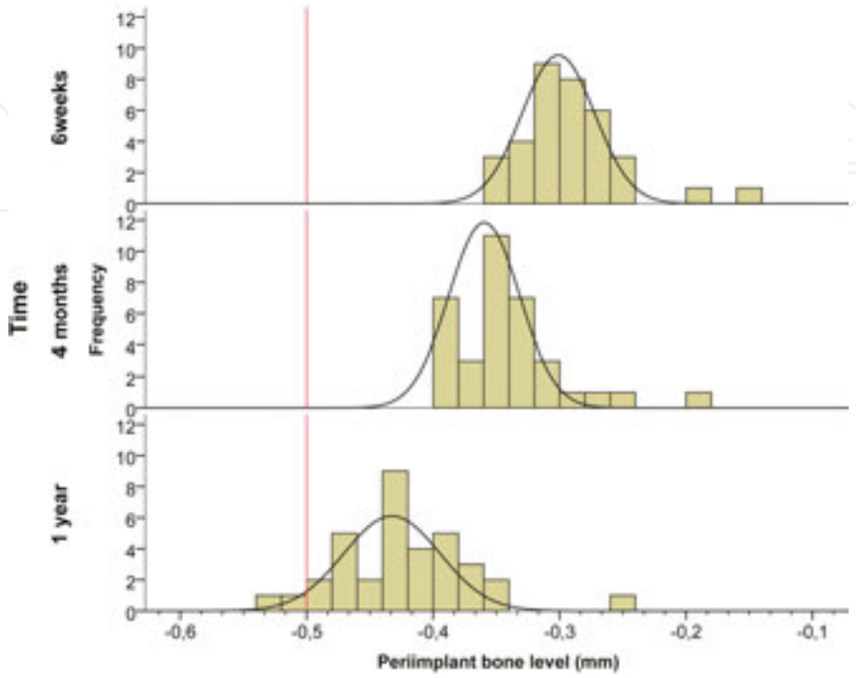

Figure 6. Frequency analysis of peri-implant bone level around early loaded SLActive dental implants in low-density bone.

Placement of implants into posterior maxillary region is often compromised by the bone resorption pattern, and pneumatization of the maxillary sinus beside the low-density bone present at this jaw region. Therefore, in such cases, sinus elevation is necessary to accommodate implants of sufficient length. Residual bone height determines surgical technique for sinus lift as well as whether implants can be placed simultaneously with sinus lift procedure or in the second stage [47]. When limited elevation of the sinus mucosa is required, this can be achieved through an implant bed using osteotomes, a technique known as osteotome sinus floor elevation and implant can be placed simultaneously [48]. The healing time prior to loading of implants inserted following sinus floor elevation is usually longer than the loading time required for implants inserted in bone of sufficient quantity [49]. Reduction of the healing time in atrophic posterior maxilla with low-density bone is particularly challenging due to reduced bone to implant contact and doubtful implant stability.

Around 95% of SLActive implants placed in the posterior maxilla via the osteotome sinus floor elevation technique without grafting achieves stability sufficient for early loading, in the sixth week of healing (**Table 1**). Favorable mid-term success rate indicates that implants with a sandblasted large-grit acid-etched active surface, when placed with the osteotome sinus floor elevation technique, can be subjected to an early loading protocol, providing their stability is confirmed by RFA [50].

ISQ	Time						
	Placement	**1 week**	**2 weeks**	**3 weeks**	**4 weeks**	**5 weeks**	**6 weeks**
Minimum	47	48	52	57	60	63	64
Maximum	75	74	75	75	77	77	78
Mean ± SD	59.55 ± 7.06	61.12 ± 6.34	62.23 ± 5.53	63.75 ± 4.56	65.88 ± 3.64	66.80 ± 3.03	67.75 ± 3.06

Table 1. Stability of SLActive implants placed via OSFE.

Density of bone at implant site affects implant stability. SLActive implants placed in the region of the second and the first premolar have comparable stability but their stability is significantly higher than implants inserted in the region of the first molar (**Figure 7**). Implant stability positively correlates with residual bone height (**Figure 8**) [50].

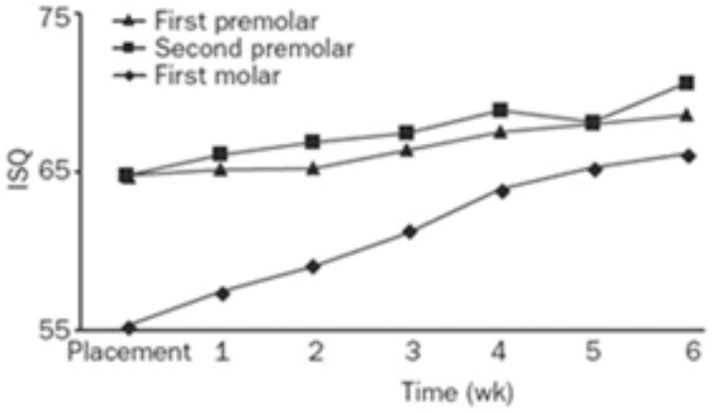

Figure 7. Stability of SLActive implants placed via OSFE regarding the jaw region.

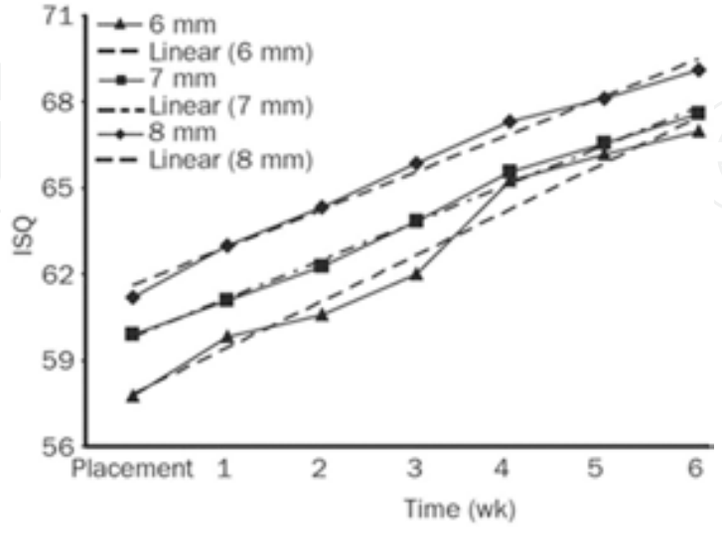

Figure 8. Comparison of implant stability with initial residual bone height.

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Grafting material is not a prerequisite for the osseointegration of dental implants with hydrophilic and nanostructured SLActive surface placed via OSFE procedure. The usage of grafting material offers no significant advantage to stability or clinical success of dental implants placed simultaneously with OSFE (**Figure 9**) [51].

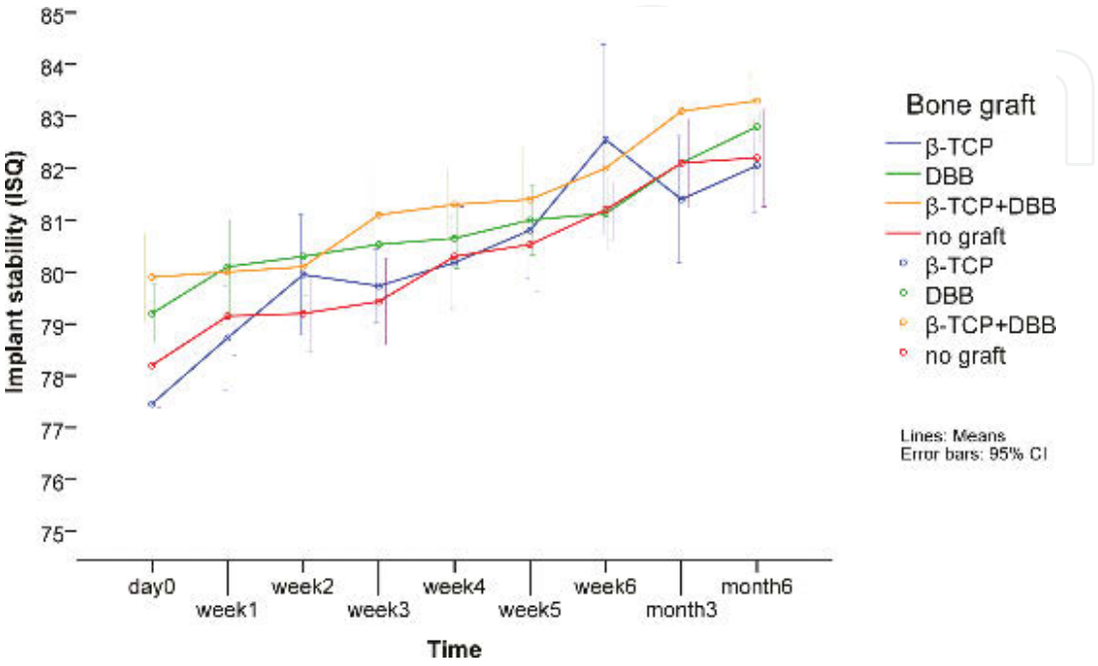

Figure 9. Stability of SLActive implants placed via OSFE regarding the usage of grafting material.

In atrophic maxillary ridges which require substantial raise of the sinus membrane implant placement using lateral sinus lift is mandatory. Eighty-three percent of SLActive implants placed simultaneously with lateral sinus lift and a mixture of autogenous bone chips and deproteinized bovine bone mineral reach the threshold stability after 8 weeks of healing, allowing an early loading protocol. This treatment protocol is associated with low early failure rate of 0.9% [52].

Another challenging indication that requires stronger bone response is implant placement into irradiated jaw. Radiation therapy causes endarteritis leading to hypoxia, hypovascularity, and hypocellularity that might jeopardize dental implant osseointegration [53]. Long-term survival rate of implants placed in irradiated jaws is 69–78%, and it is influenced by the jaw region, irradited dose, and surface roughness [54–56]. Roughened dental implants have 2.9 times reduced risk for failure in irradiated jaws compared with turned implants [54]. Sandblasted acid-etched implants with or without a chemically modified surface can be used in irradiated patients with a high predictability of success. The overall cumulative 5-year survival rates of SLA and SLActive implants in irradiated jaws are similar and the crestal bone level around both implant surfaces remains stable at least 5 years after placement. Hydrophilic surface might affect only early survival of dental implant placed in irradiated bone [57].

Microangiopathies and hyperglycemia associated with diabetes mellitus impaires bone regeneration and might affect early implant failure rates in such patients. Diabetic patients with glycated hemoglobin above 8.0% have delayed implant osseointegration and require a longer healing time [58, 59]. Despite the promising result of animal research that SLActive surface provides accelerated osseointegration of dental implants and better prognosis for implant treatment in diabetic patients, clinical assessment revealed similar outcomes for SLActive and SLA surfaces [60, 61].

3. Role of laser microgrooved zirconia surface in dental implant treatment

Although titanium can still be considered the reference standard material for dental implants with a few limitations such as rare allergy to metals or gingival retraction or translucidity in thin gingival biotypes and subsequent unsatisfactory esthetic [62, 63]. The development of high mechanical strength ceramics has made them a viable alternative [64]. Yttrium-partially stabilized tetragonal zirconia (Y-TZP) offers several advantages due to its flexural strength and high resistance to fracture, favorable esthetics as well as excellent osseointegration observed in animal studies [65, 66].

However, roughening the surface of the zirconia implant is a challenge mainly due to its resistance to chemical or physical modifications. Several approaches have been proposed as follows: chemical and pharmacological surface modification, sand-blasting and acid etching, the use of nanotechnology, or biomimetic coatings, and addition of micro-and macro-retentions [67–69]. These modifications result in various degrees of surface roughness and content of contaminants.

The zirconia dental implants available on the market are sandblasted. Recently, technique for microstructuring cylindrical zirconia implants by femtosecond laser ablation has been introduced. In addition to sandblasting, surface is modified using femtosecond laser ablation, which creates an isotropic pattern of microgrooves on the implant surface [5]. This technique is fast, provides precise control of texture allowing production of textures with complex shape, and as a non-contact procedure, it does not cause contamination [5].

Cells modify their morphology, adhesion, and cytoskeletal organization according to the substrate topography [70]. On flat zirconia dental implant surface, osteoblasts are disorganized and loosely attached with few lamellipodia mainly directed toward the cracks or other topographical accidents (**Figure 10a–c**). Creation of microgrooves of 30 μm width and 70 μm separation on zirconia dental implant surface induces favorable cell morphology, increases cell density, and enhances cell activity [71]. Osteoblasts align along the axis of microgrooves with lamellipodia directed toward the inner surface and connected to the base and walls of the microgrooves (**Figure 10d–f**). Filopodia extensions retained in the nanometric structures of the microgroove walls cell further improve adhesion of osteoblasts to modified zirconia surface and increase cell density (**Figure 11a–d**). Osteoblasts adhesion occurs first in microgrooves and later on the flat area of zirconia surface (**Figure 12a–f**). Further, activity of the osteoblasts is tripled by adding the microgrooves to zirconia surface [71]. Microgrooves on

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zirconia implants host bioactive molecules and enhance the initial stages of bone formation [72].

Favorable cellular events directed by microgrooved zirconia implant surface are provided by increased roughness and enhanced chemical composition of the sandblasted zirconia surface following its laser modification. This surface treatment increases proportions of zirconium and oxygen, whereas decreases content of carbon and aluminum allowing high osteoblastic activity on sandblasted, laser micro-grooved zirconia [71]. This modified zirconia surface exhibit higher values of roughness parameters and reduced the presence of contaminants not only in comparison with its predecessor, nongrooved sandblasted zirconia, but also to sandblasted, hightemperature-etched titanium implants (**Tables 2** and **3**) [73].

Roughness parameters	**Surface**			
	Sandblasted zirconia	**Sandblasted zirconia with microgrooved neck**	**Sandblasted zirconia all microgrooved**	**Sandblasted, high temperature etched titanium**
R_a(μm)	1.28 ± 0.2	2.43 ± 0.6*	9.50 ± 0.25*	1.78 ± 0.6
R_q(μm)	1.82 ± 0.51	3.48 ± 0.30*	11.51 ± 0.31*	2.02 ± 0.43
R_z(μm)	11.4 ± 0.6	40.42 ± 0.25*	40.74 ± 0.28*	15.8 ± 0.5
R_t(μm)	18.46 ± 0.82	52.68 ± 0.9*	60.36 ± 0.22*	23.63 ± 0.32

Surface roughness parameters (R_a, R_q, R_z, R_t) expressed as ($\bar{x}$ ± SD) (*p < 0.05).

Table 2. Topographic characteristics of implant surfaces.

EDX surface analysis	**Surfaces**			
	Sandblasted zirconia	**Sandblasted zirconia with microgrooved neck**	**Sandblasted zirconia all microgrooved**	**Sandblasted, high temperature etched titanium**
C %	19.7 ± 0.8%	1.6 ± 0.35%*	0.3 ± 0.12%*	2.3 ± 1.7%
Al %	4.3 ± 0.9%	1.16 ± 0.2%*	0.18 ± 0.1%*	1.7 ± 0.3%
O %	12.6 ± 0.5%	22.7 ± 0.2%*	23.1 ± 0.12%*	15 ± 0.6%
Zr %	60.2 ± 0.7%	73.7% ± 0.15%*	76.3 ± 0.2%*	0%
Ti %	0%	0%	0%	81 ± 1.3%

Expressed in percentages as $\bar{x}$ ± SD (*p < 0.05).

Table 3. Elements present in surface chemical composition.

The addition of microgrooves in the 2-mm wide neck area of the implant increases surface roughness by 6.5 times and almost 12 times in the zirconia implants processed over the entire intraosseous surface. Microgrooves provide more retentive areas and greater bone-to-implant contact resulting in higher stability of this implants proven by the increase in insertion and removal torque and decrease of PTV values (**Tables 4–6**) [73].

Surface	IT(Ncm)			
	$\bar{x}$	SD	SE	Median
Sandblasted, high temperature etched titanium	57.10	1.80	0.51	55.76
Sandblasted zirconia	46.08	0.70	0.20	44.87
Sandblasted zirconia with microgrooved neck	53.20	1.30	0.37	50.98
Sandblasted zirconia all microgrooved	69.60	1.20	0.34	67.82

Table 4. Insertion Torque values (IT) recorded at implant placement.

Surface	RT (Ncm)		
	Month 1	Month 2	Month 3
Sandblasted zirconia	64.08 ± 0.42 (64.07)	78.24 ± 0.35 (78.38)	199.19 ± 0.99 (199.47)
Sandblasted zirconia with microgrooved neck	69.19 ± 0.37 (69.17)	88.82 ± 0.41 (88.86)	215.13 ± 0.99 (215.06)
Sandblasted zirconia all microgrooved	84.95 ± 0.25 (85.03)	126.96 ± 0.81 (126.65)	240.15 ± 1.04 (239.90)
Sandblasted, high temperature etched titanium	71.25 ± 0.43 (71.28)	99.85 ± 0.44 (99.98)	226.98 ± 1.06 (226.72)

Table 5. Removal torque test (RT) performed at three evaluation time points.

Surface	PTV		
	Month 1	Month 2	Month 3
Sandblasted zirconia	−1.52 ± 0.01 (−1.52)	−2.17 ± 0.01 (−2.17)	−2.41 ± 0.02 (−2.41)
Sandblasted zirconia with microgrooved neck	−1.85 ± 0.02 (−1.85)	−2.42 ± 0.01 (−2.42)	−3.11 ± 0.01 (−3.11)
Sandblasted zirconia all microgrooved	−2.49 ± 0.02 (−2.5)	−4.16 ± 0.01 (−4.16)	−5.69 ± 0.03 (−5.7)
Sandblasted, high temperature etched titanium	2.11 ± 0.35 (−2.00)	−2.70 ± 0.01 (−2.70)	−3.59 ± 0.05 (−3.60)

Values expressed as ±SD (median).

Table 6. Changes in Periotests values (PTV) over time.

Microgrooved implants reduces crestal bone level in comparison with microthreaded titanium implants and particularly with rough neck implants without microthreading (sandblasted zirconia) (**Table 7**). Although microthreads at implant neck transform the shear force between the implants and crestal bone into the compressive force to which bone is the most

resistant allowing preservation of bone tissue, the addition of microgrooves that interlock the adjacent bone seems to be more efficient [73, 74].

Surface	RCBL (mm)		
	Month 1	Month 2	Month 3
Sandblasted zirconia	0.27 ± 0.03 (0.26)	0.32 ± 0.01 (0.32)	0.56 ± 0.01 (0.56)
Sandblasted zirconia with microgrooved neck	025 ± 0.03 (0.23)	0.22 ± 0.02 (0.23)	0.36 ± 0.01 (0.36)
Sandblasted zirconia all microgrooved	0.24 ± 0.02 (0.22)	0.24 ± 0.01 (0.24)	0.26 ± 0.01 (0.26)
Sandblasted, high temperature etched titanium	0.27 ± 0.04 (0.28)	0.30 ± 0.02 (0.30)	0.36 ± 0.01 (0.36)

Values expressed as ±SD (median).

Table 7. Radiographic crestal bone loss (RCBL).

The addition of microgrooves to the entire intraosseous surface of zirconia dental implants enhances primary and secondary implant stability, which promotes bone tissue ingrowth and preserves crestal bone levels [73]. Data from animal models indicate that zirconia femtosecond laser all-treated surface achieves good osseointegration and could be predictable treatment option in the implantological daily practice [75]. Histological, radiological, and histomorpho-

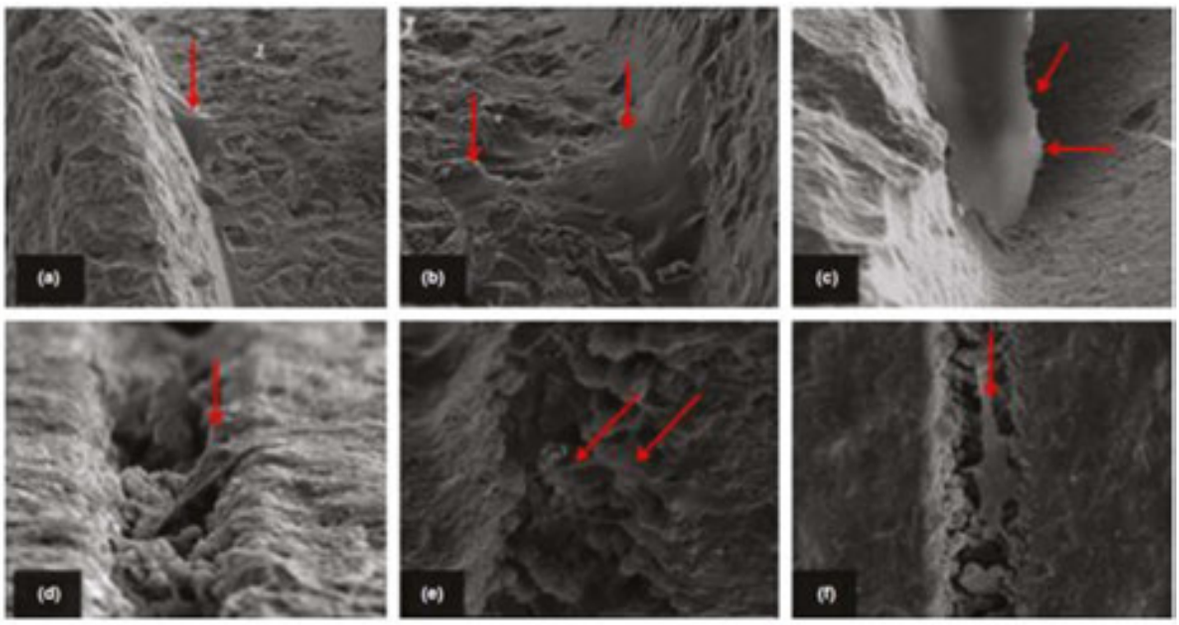

Figure 10. SEM evaluation of cell morphology on sandblasted (a-c) and sandblasted, laser micro grooved zirconia (d-F) at 7 days. (a) a cell in the base of an implant thread; (b) close-up view of a couple of cell bodies close to the base of the thread; (c) a cell body with very short cell lamellipodia located in a crack surface (shown at high magnification); (d) cell at the border of a microgroove; (e) lamellipodia extend inside microgrooves, bridging microgroove borders; (f) cell body aligned in the direction of the microgroove.

metric evaluation of zirconia implants treated with femtosecond laser revealed that they can be successfully subjected to immediate loading protocol [72].

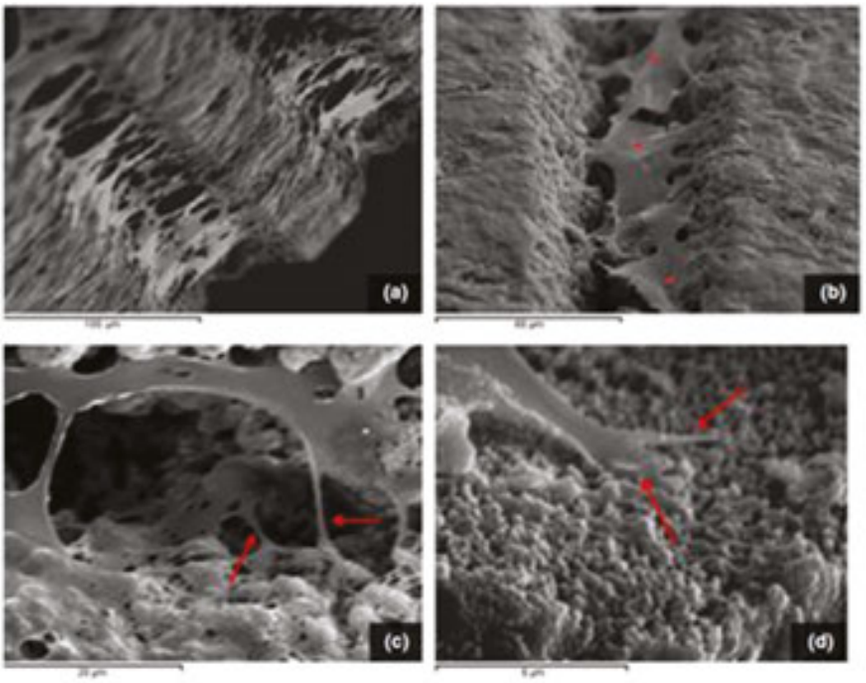

Figure 11. Cells on sandblasted, microgrooved zirconia surface at 7 days (high magnification). (a) lateral view of multiple cells firmly adhered to the inner surface of the microgrooves; (b) cell body alignment at the base of the microgrooves; (c) lamellipodia network at walls and base of the microgrooves; (d) inner wall of a microgroove showing filopodia connected to the nanorough texture of the microgroove walls.

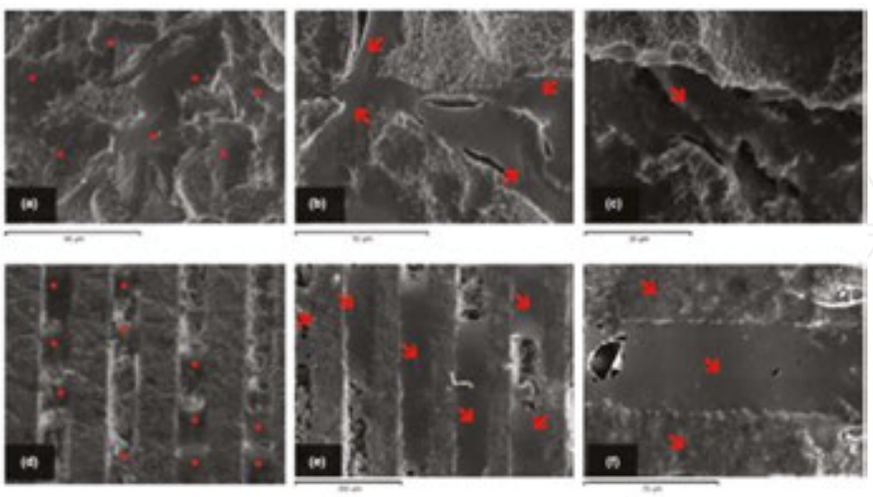

Figure 12. SEM evaluation of cell morphology on sandblasted (a–c) and sandblasted, laser micro grooved zirconia (d–f) at 15 days. The cells established intercellular contacts and formed layers; contact between the cell and the surface occur mainly in topographic accidents (high magnification) (a–c). Cells fill the microgroove completely; cells also form in the pitch areas (d–f).

4. Conclusion and implications for future research

Surface modifications are of great clinical importance, regardless of the core implant material, titanium, or zirconium. Hidrophilic and nanostructured SLActive surface accelerates osseointegration and provides conditions for early or immediate loading. This surface ensures predictable implant outcome even in low-density bone due to enhanced biological response. Initial in vitro and in vivo animal studies indicate improved osteoblastic activity, enhanced osseointegration and stabile crestal bone level around microogroved zirconia implants treated with femtosecond laser. This recent findings make them a potential treatment option for everyday implantology. However, future research should examine clinical effects of laser microogroved zirconia implants in randomized clinical trials using sufficient sample size and proven methodology.

Author details

Aleksa Markovic

Address all correspondence to: maleksa64@gmail.com

Faculty of Dental Medicine, Belgrade, Serbia

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