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ORIGINAL ARTICLE

Prospective clinical evaluation of 201 direct laser metal forming implants: results from a 1-year multicenter study

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Abstract This prospective clinical study evaluated the survival rate and the implant-crown success of 201 direct laser metal forming (DLMF) implants in different clinical applications, after short-term follow-up of functional loading. At the 1-year scheduled follow-up examination, several clinical, radiographic, and prosthetic parameters were assessed. Success criteria included absence of pain, sensitivity, suppuration, exudation; absence of implant mobility; absence of continuous peri-implant radiolucency, DIB <1.5 mm; absence of prosthetic complications at the implant-abutment interface. A total of 201 implants (106 maxilla, 95 mandible) were inserted in 62 patients (39 males, 23 females; aged between 26 and 65 years) in eight different clinical centers. The sites included anterior (n=79) and posterior (n=122) implants. The overall implant survival rate was 99.5%, with one implant loss (maxilla: 99.0%, 1 implant failure; mandible: 100.0%, no implant failures). The mean DIB was 0.4±0.2 mm. Among the survived implants (200), five did not fulfill the success criteria, giving an implant-crown success of 97.5%. This 1-year follow-up prospective clinical study gives evidence of very high survival (99.5%) and success (97.5%) rates using DLMF implants.

Keywords Dental implants · Porous titanium implants · Bone ingrowth · Direct laser metal forming (DLMF)

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Introduction

Dental implants currently available on the market are manufactured from commercially pure titanium or its alloy Ti-6Al-4V (90% titanium, 6% aluminum, 4% vanadium). Until now, dental implants have been produced by machining titanium rods, with subsequent post-fabrication processing and application of surface treatments or coatings, with the aim to promote osseointegration, accelerating the bone-healing processes [1]. Over the last years, a considerable number of surface modifications, such as sandblasting, acid-etching, grit-blasting, anodization, discrete calcium-phosphate crystal deposition, coatings with biological molecules and chemical modification have been introduced, in the attempt to produce better implant surfaces [2]. Rough surfaces have demonstrated better biomolecule adsorption from biological fluids and a better bone response, because of the ability to influence cellular behavior. Several in vitro reports indicate that in comparison to smooth ones, rough surfaces improve initial cellular responses, including cytoskeletal organization and cellular differentiation with matrix deposition [3]. Histologically, it has been clearly demonstrated that rough surfaces can effectively promote better and faster osseointegration, when compared to smooth surfaces [4, 5]. From a clinical point of view, several studies have reported excellent long-term survival and success rates for rough surface implants [6–8].

All the traditional methods utilized for manufacturing and processing dental implants, however, result in a high-density titanium structure with a micro- or nanorough surface; using these methods, it is difficult to directly fabricate *porous implants* with a functionally graded structure, possessing a gradient of porosity perpendicular to the long axis, a relatively high porosity at the surface and a high density in the core. Porous titanium and its alloys have been introduced and used in dental and orthopedic applications since the end of the 1960s, with considerable success [9, 10].

Controlled porosity is desirable to decrease the mismatch between the elastic modulus of the titanium implant and that of the bone tissue, thus reducing stress shielding and achieving stable long-term fixation [11]. The elastic properties of conventional dental implants, in fact, are different from those of surrounding bone. The stiffness of a dental implant depends intrinsically on the elastic modulus (Young's modulus) of the employed material, as well as the geometric properties of the implant itself [12, 13]. The elastic modulus of commercially pure titanium (112 Gpa) and Ti-6Al-4V of titanium alloy (115 Gpa) are considerably higher than that of cortical bone (10–26 Gpa) [13, 14]. This difference could lead to stress shielding of the residual bone. Porous structures can certainly help to reduce the stiffness mismatch between implant and bone tissue [13]. Moreover, the presence of a porous surface structure increases the osseointegration of the implant in the body [14]. Osseointegration is favored by porous implants that improve fixation by creating a mechanical interlock via the growth of bone into the porous structure. Improved fixation can be achieved by *bone ingrowth* into and through a porous matrix of metal, bonding the implant to the bone. Finally, body fluid transport through the porous scaffold matrix is possible, which can trigger bone ingrowth, if substantial open pore interconnectivity is established [15].

Because of these reasons, there is a demand for fabrication methods for bulk porous titanium that can control porosity, pore size and distribution, and mechanical properties. Several techniques have been introduced in recent years to produce a porous coating on the implants [16, 17]. Spraying techniques are the most commonly used, however the fatigue strength of an implant coated by such techniques may be reduced by up to 1/3 in comparison with the uncoated implant [16]. Many available methods for producing porous titanium and titanium alloy scaffolds include co-sintering the precursor particles, plasma spraying of the powder onto a dense substrate followed by cutting off the porous layer, compressing and sintering of titanium fibers, solid-state foaming by expansion of argon-filled pores [18]. However, none of these conventional techniques has enabled the building of scaffolds with a completely controlled design of the external shape as well as the interconnected pore network [19].

In the last few years, considerable progress has been made in the development of rapid prototyping (RP) methods, including direct laser metal forming (DLMF) [12, 13, 20–24]. Rapid prototyping (RP) is a strategy to directly fabricate physical objects with defined structure and shape on the basis of virtual 3D data [13, 20]. DLMF is a timesaving and costless metal forming procedure in which a high-power laser beam is focused on a metal powder bed and programmed to fuse particles according to a CAD file, thus generating a thin metal layer. Apposition of subsequent layers gives shape to a desired three-dimensional form with the need of minimal post-processing requirements [12, 13, 21–23]. The performance of DLMF processing depends on several parameters, which include the diameter of the focused laser beam, power rating of the laser, scanning speed, average particle size of the starting material powder, layer thickness, track overlap, and process atmospheric conditions [22]. Using DLMF, it is now possible to create dental implants with different shapes and textures directly from CAD models by the laser fusion of titanium microparticles [12, 13]. Laser forming methods allow the fabrication of functionally graded titanium implants, with a gradient of porosity perpendicular to the long axis. With DLMF, a porous surface structure for improved bone ingrowth capability is provided, eliminating the need for

applying a thermally sprayed titanium coating on the implants. DLMF makes it possible to generate implants with a graded elasticity, incorporating a gradient of porosity, from the inner core to the outer surface [11–14]. This new functionally graded material has the potential to have a more similar elastic modulus (77 Gpa) to that of the surrounding bone, for a more natural transfer of loading stress [20–23]. The technique also permits the fabrication of a porous structure with controlled porosity, pore size, pore distribution, and interconnection [12, 23, 24]. DLMF implants, moreover, require no post-processing procedures. Considering that surface contamination is a potential problem with traditional processing for fabrication of dental implants, since it is carried out under mineral oil refrigeration and with different materials for machining burs, the low risk of surface contamination is a potential advantage of the DLMF procedure.

The chemical and physical properties of dental implants fabricated with the DLMF technique have been extensively studied [12]. The biological response to the DLMF implant surface has subsequently been investigated in different in vitro studies, in which human fibrin clot formation [23] and the behavior of human osteoblast and mesenchymal stem cells [24] have been analyzed. The biological behavior of DLMF implants has now also been investigated in vivo, in different histologic and histomorphometric studies in humans [25, 26].

The aim of this prospective study was to evaluate the survival rate and the implant-crown success of 201 DMLF implants (TixOs^{R}, Leader-Novaxa, Milan, Italy) in different clinical applications, such as single crowns (SCs), fixed partial prostheses (FPPs), and fixed full-arch prostheses (FFAs).

Materials and methods

Patient population

Between April and July 2009, a total of 68 patients (44 males and 24 females) were recruited from eight different clinical centers to take part in this prospective clinical study. Inclusion criteria were adequate bone height and width to place an implant of at least 3.3 mm in diameter and 8.0 mm in length. Exclusion criteria consisted of poor oral hygiene, active periodontal infections, uncontrolled diabetes, bruxism, heavy smoking habit (more than ten cigarettes/day). Six patients could not take part in the study (two for inadequate bone height and width, one for poor oral hygiene, one for active periodontal infections, one for bruxism, and one for heavy smoking habit). Sixty-two patients (39 males and 23 females, aged between 26 and 65 years) fulfilled the inclusion criteria, presenting none of the conditions listed and were all enrolled in the study. The Ethics Committee for Human Clinical Trials at the University of Varese, Italy, approved the study protocol, which was explained to each subject, and all patients signed informed consent forms.

Implant design and surface characterization

Two-hundred and one screw-type cylindrical implants (Fig. 1) were manufactured from titanium alloy (Ti-6Al-4V) with a DLMF technique (TixOs^{R}, Leader-Novaxa, Milan, Italy). The DLMF implants were made of master alloy powder with a particle size of 25–45 μm as the basic material. Processing was carried out in an argon atmosphere using a powerful Yb (ytterbium) fiber laser system (EOS GmbH, Munich, Germany) with the capacity to build a volume up to 250×250×215 mm using a wavelength of 1,054 nm with a continuous power of 200 W at a scanning rate of 7 m/s. The size of the laser spot was 0.1 mm. To remove residual particles from the manufacturing process, the sample was sonicated for 5 min in distilled water at 25° C, immersed in NaOH (20 g/l) and hydrogen peroxide (20 g/l) at 80°C for 30 min, and then was further sonicated for 5 min in distilled water. Acid etching was carried out by immersion of the samples in a mixture of 50% oxalic acid and 50% maleic acid at 80°C for 45 min, followed by

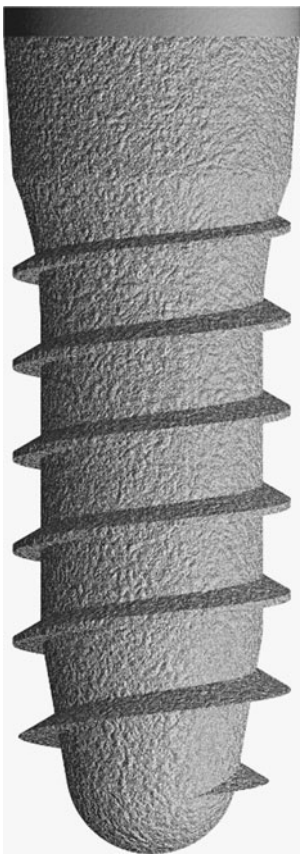

Fig. 1 Schematic drawing of the DLMF implant evaluated in this study

washing for 5 min in distilled water in a sonic bath [23–26]. The direct laser preparation provided an implant surface with a roughness surface that had an Ra value of 66.8, Rq value of 77.55, and Rz value of 358.3 μm, respectively [23, 24] (Fig. 2).

Pre-operative work-up

A complete examination of the oral hard and soft tissues was carried out for each patient. Panoramic radiographs formed the basis for the primary investigation. Pre-operative work-ups included an assessment of the edentulous ridges using casts and diagnostic wax-up. Where necessary, computed tomography (CT) scans were used as the final investigation. CT datasets were acquired using a modern cone beam scanner and then transferred in the DICOM format to specific implant navigation software to perform a three-dimensional reconstruction of the maxillary bones. With this navigation software, it was possible to correctly assess the width of each implant site, the thickness and the density of the cortical plates and the cancellous bone, as well as the ridge angulations. On the basis of this information, surgical templates were manufactured.

Implant placement

Local anesthesia was obtained by infiltrating articaine 4% containing 1:100.000 adrenaline. A midcrestal incision was made at the sites of implant placement. The mesial and the distal aspects of the crestal incision were connected to two releasing incisions. Full-thickness flaps were reflected exposing the alveolar ridge, and preparation of implant sites was carried out with spiral drills of increasing diameter (2.0 and 2.3 mm, to place an implant with 3.3-mm diameter; 2.0, 2.6, and 2.8 mm, to place an implant with 3.75-mm diameter; 2.0, 2.6, and 3.2 mm, to place an implant with 4.5-mm diameter; two additional 3.8- and 4.2-mm drills were used to prepare the site for 5.5-mm-diameter implants), under constant irrigation. Implants were positioned at the bone crest level. The flaps were repositioned to cover the implants completely and were secured in position by interrupted sutures.

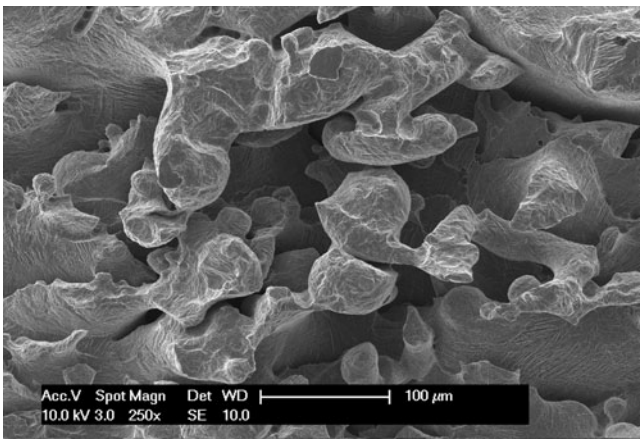

Fig. 2 Scanning electron microscopy of the implant surface (x250). The implant has an irregular surface with ridge-like and globular protrusions, interspersed by intercommunicating pores and irregular crevices

Postoperative treatment

All patients received oral antibiotics, 2 g each day for 6 days (Augmentin^{R}, Glaxo-Smithkline Beecham, Brentford, UK). Postoperative pain was controlled by administering 100 mg nimesulide (Aulin^{R}, Roche Pharmaceutical, Basel, Switzerland) every 12 h for 2 days, and detailed instructions about oral hygiene were given, with mouthrinses with 0,12% chlorhexidine (Chlorexidine^{R}, OralB, Boston, MA, USA) administered for 7 days. Suture removal was performed at 8–10 days.

Healing period

A two-stage technique was used to place the implants. The healing time was 2–3 months in the lower jaw and 3–4 months in the upper jaw. Second-stage surgery was conducted to gain access to the underlying implants and healing abutments were placed. A mesio-distal crestal incision, limited to the implant site, was placed and the ridge mucosa was elevated to uncover the implant, followed by the replacement of the cover screw with a healing abutment. The mucosal flap was adjusted to the healing abutment and then sutured in position. In all prosthetic rehabilitation protocols (SCs, FPPs, and FFAs), the abutments were placed 2 weeks after the second surgery so that acrylic interim restorations could be provided. Acrylic resin provisional restorations were used to monitor the implants' stability under a progressive load and to obtain good soft-tissue healing around the implant before fabrication of the definitive restorations. The temporary restorations remained in situ for 3 months, and after this period definitive restorations were placed. All definitive restorations were ceramo-metallic, cemented with zinc phosphate cement or zinc-eugenol oxide cement.

Clinical, radiographic, and prosthetic evaluation

The following clinical parameters were investigated, after 1 year of functional loading, for each implant:

- presence/absence of pain – sensitivity [27]
- presence/absence of suppuration – exudation [27]
- presence/absence of implant mobility, tested manually using the handles of two dental mirrors [27].

Moreover, intraoral periapical radiographs were taken for each implant at the baseline (immediately after implant

insertion) and at the 1-year scheduled control. Radiographs were taken using a Rinn alignment system with a rigid film-object-X-ray source coupled to a beam-aiming device in order to achieve reproducible exposure geometry. For each implant, customized positioners were used for precise repositioning and stabilization of the radiographic template.

Two different radiographic parameters were evaluated:

- presence/absence of continuous peri-implant radiolucencies
- the distance between the implant shoulder and the first visible bone contact (DIB) in mm, measured by means of an ocular grid.

With the latter value, crestal bone level changes at 1 year were registered as modifications in the distance from the implant shoulder to the bone level on the mesial and distal implant side. To correct dimensional distortion, the apparent dimension of each implant was measured on the radiograph and then compared to the real implant length.

Finally, at the 1-year follow-up session, prosthesis function was tested. Static and dynamic occlusion was evaluated using standard occluding papers. Careful attention was dedicated to the analysis of any prosthetic complications at the implant–abutment interface (such as abutment screw loosening, abutment fracture), which were considered as primary endpoints of this study; other complications (such as ceramic fractures or overdenture-related problems) were also reported.

Implant survival and implant-crown success criteria

The evaluation of implant survival and implant-crown success was performed according to the following clinical, radiographic, and prosthetic parameters. Implants were basically divided into two categories: "survival" and "failed" implants. An implant was classified as a "survival implant" when it was still in function at the end of the study, after 1 year of functional loading. Implant losses were categorized as failures; implants presenting pain upon function, suppuration, or clinical mobility were removed, and were all classed as failures. The conditions for which implant removal could be indicated included failure of osseointegration or infection, recurrent peri-implantitis, or implant loss due to mechanical overload. A distinction was made between "early" (before the abutment connection) or "late" (after the abutment connection) implant failures.

To achieve implant-crown success, the following clinical, radiographic, and prosthetic success criteria should be fulfilled:

- absence of pain or sensitivity upon function
- absence of suppuration or exudation
- absence of clinically detectable implant mobility
- absence of continuous peri-implant radiolucency
- DIB <1.5 mm after 12 months of functional loading
- absence of any prosthetic complications at the implant–abutment interface

Results

Patient population and implant-supported restorations

Two hundred and one implants were placed by eight experienced surgeons (one for each different clinical center involved in this study). One hundred and six implants (52.7%) were inserted in the maxilla, while 95 implants (47.3%) were inserted in the mandible. Thirty-nine implants (19.4%) were placed in the maxillary anterior region, 67 implants (33.3%) were placed in the maxillary posterior region; 40 implants (19.9%) were placed in the mandibular anterior region and 55 implants (27.4%) in the mandibular posterior region. The distribution of implants by length and diameter is shown in Table 1. The most frequent indication was the treatment of single tooth gaps (105 implants), while the least frequent was the restoration of fully edentolous patients (16 implants). Eighty implants were used to restore partially edentolous patients. The prosthetic restorations comprised 105 single crowns (SCs), 45 fixed partial prostheses (FPPs), and two fixed full-arch prostheses (FFAs). Each fixed full-arch prosthesis was supported by eight implants.

Implant survival

At the end of the study, the overall implant survival rate was 99.5%, with 200 implants still in function. One implant failed and had to be removed, in the posterior maxilla. This failure was classified as "early failure", showing clinical mobility due to lack of osseointegration. No other failures due to infections with pain or suppuration were observed before the connection of the abutment. No "late failures" (observed after the abutment connection) were reported. In the maxilla, the survival rate was 99.0%, with one implant

Table 1 Implant distribution by length and diameter (mm)

Diameter	Implant length				
	8.0	10.0	11.5	13.0	Total
3.3	10		2	17	29
3.75	20	30	27	50	127
4.5	4	4	11	16	35
5.5	-	-	5	5	10
Total	34	34	45	88	201

Table 2 Number of patients, implants, failures, survival, and success rate for each clinical center involved in the study

Center	No. of patients	No. of implants	No. of failures	Survival rate (%)	Success rate (%)
1	20	96	0	100.0	96.8
2	12	43	1	97.6	95.3
3	12	38	0	100.0	100.0
4	6	9	0	100.0	100.0
5	4	7	0	100.0	100.0
6	4	4	0	100.0	100.0
7	2	2	0	100.0	100.0
8	2	2	0	100.0	100.0
Overall	62	201	1	99.5	97.5

failed and removed. In the mandible, the survival rate was 100.0%, with no implant failures (Table 2).

Implant-crown success

Two hundred implants were still in function at the end of the study. Among these implants, 195 (97.5%) were classified in the implant-crown success group. None of these implants caused pain or clinical mobility, suppuration, or exudation, with a DIB <1.5 mm after the first year of functional loading, and none had any prosthetic complication at the implant–abutment interface. Five implants (2.5%), on the contrary, could not fulfill the implant-crown success criteria. One single implant (0.5%) showed sensitivity upon function. Two other implants (1.0%) revealed a DIB >1.5 mm after the first year of function. In two additional implants (1.0%), the prosthetic abutments became loose during the first year of loading in two single crowns (SCs) situated in the posterior area of the mandible. These abutments were reinserted and no further loosening was observed in the period of this study. No complications were observed at the implant-abutment connection fixed partial prostheses (FPPs) and fixed full-arch prosthesis (FFAs) and no abutment fractures were seen either. At the end of the study, the radiographic evaluation of the implants revealed a mean (± standard deviation) distance from the implant shoulder to the first crestal bone to implant contact (DIB) of 0.4±0.2 mm.

Discussion

Requirements for rapid bone ingrowth are precise control of scaffold porosity and internal pore architecture parameters (pore interconnectivity, size, geometry, orientation, distribution) [28]. Porosity is necessary to maximize nutrient diffusion, interstitial fluid and blood flow, to control cell growth and function, to manipulate tissue differentiation, and to optimize scaffold mechanical function [43]. Porosity and pore interconnectivity play a critical role in bone ingrowth [28, 29]. It is recognized that one of the critical factors for bone ingrowth is the size of interconnecting pores, and several investigators have studied bone ingrowth into porous systems [30]. The scaffold's porosity and degree of pore interconnectivity directly affect the diffusion of physiological nutrients and gases [31] too, and the removal of metabolic waste and by-products from cells that have penetrated the scaffold. Moreover, an open pore geometry that allows cell ingrowth and reorganization in vitro provides the necessary space for neovascularization from surrounding tissues in vivo [30, 31]. The highly porous microstructure with interconnected porous networks is critical in ensuring spatially uniform cell distribution, cell survival, proliferation, and migration in vivo [32]. Some researchers [33] indicated the need for pore size ranging from 200 to 400 μm, while Yoshikawa [34] successfully employed scaffolds with 500-μm nominal pore size. When the pores employed are too small, pore occlusion by cells will prevent cellular penetration and matrix elaboration within the scaffold [33, 34]. Although optimum pore size required for implant fixation remains undefined, the consensus is that in order to optimize mineralized bone ingrowth, pore sizes between 100 and 400 μm are necessary [30, 33].

These structural and geometric features are of paramount importance, but they are difficult to achieve using current manufacturing methods. A key requirement for rapid prototyping (RP) technologies is control over the scaffolds' structure, including porosity, pore size, shape, volume, and interconnectivity. The DLMF technique has been proposed to build implants with graded controlled internal porosity [12, 13, 23–25]. With DLMF it is possible to control the porosity of each layer and consequently of the 3D model by changing the processing parameters, such as laser power and peak power (for CW and pulsed lasers, respectively), laser spot diameter, layer thickness, hatching pitch (or scan spacing), scan speed and scanning strategy, or by modifying the size of the original titanium particles [11–13, 23–26]. With DLMF, moreover, pore interconnectivity, size, shape, and distribution are controlled [11, 23].

In a recent study of Xue et al. [35], the effects of DLMF porous titanium structure on bone cell responses were evaluated in vitro with human osteoblast cells. The results showed that cells spread well on the surface of porous titanium and formed strong local adhesion. The results obtained also indicated that a critical pore size of 200 μm or higher is needed for cell ingrowth into the pores, below which cells bridged the pore surface without ingress. The physiologic response to an inserted porous titanium implant is comparable to the healing cascade of cancellous defects, with newly formed tissues infiltrating the void spaces of the porous material. Capillaries, perivascular tissues, and osteoprogenitor cells migrate into porous spaces and incorporate the porous structure by forming new bone. With initial sufficient stability, the early tissue that infiltrates the pores differentiates into bone by either direct bone formation within the pores, or appositional bone growth from the adjacent bone in the porous region. The recruitment of osteoprogenitor cells is of paramount importance for the deposition of new bone on the implant surface [36]. Osteoprogenitor cells are driven onto the implant surface through the established fibrin network around the implant. This is crucial to the early stages of osseointegration. Previous studies have indicated that there is a correlation between surface roughness and fibrin clot retention [37]. In a recent in vitro study with human fibrin, a stable three-dimensional fibrin network 35 has completely and immediately covered the new porous DLMF surface. Moreover, the DLMF surface is able to attract osteoprogenitor cells; these cells can adhere to this porous substrate and subsequently differentiate into functional osteoblasts, producing wowen bone together with appreciable amounts of bone morphogenetic proteins, vascular endothelial growth factor and other specific bone proteins [24]. After many years of research, mechanisms that regulate cell function and differentiation have been partially elucidated. Cells interact with their substratum via integrins, specific linkage proteins [38, 39]. Integrins are proteins associated with the cellular membrane, and they are responsible for focal adhesion plaque formation [38]. The formation of focal adhesion plaques is a prerequisite for the development of signaling transduction in cell adhesion, and is one of the important indicators for cell activity on the substrates. In fact, integrins are linked through their cytoplasmic domain to specific cytoskeleton linkage proteins such as alfa-actinin, talin, vinculin, paxillin, and tensin. Since the cytoskeleton is functionally connected to the nucleus [38, 39], it is not surprising that, through focal adhesion, mechanical forces applied from the substratum can be transformed into biochemical signals within cells [39, 40]. In fact, specific adhesion receptors linked to the deep cytoskeleton, such as integrins, cadherins, and mechanoreceptors, if spatially and temporally "activated" from the geometry of the substratum, can provide preferred paths for mechanical signals to enter the cell, "activating" the mechanisms of transduction [38–40]. Structural and geometric properties of an implant surface can influence cell shape and size, with consequences on gene expression.

The DLMF surface geometry, rich in interconnecting pores and cavities of 100–200 μm size, could represent an ideal environment for osteogenic phenotype expression. The shape cells are forced to adopt within the three-dimensional microstructure of pores and cavities may be responsible of creating mechanical stresses that modulate osteogenic phenotype expression [38–40].

Further, long-term clinical studies will be necessary to investigate the potential of DLMF implants in restoring partially or completely edentolous arches. However, this 1-year follow-up prospective clinical study gives evidence of very high survival (99.5%) and success (97.5%) rates using DLMF implants. In this study on 201 DLMF implants, in fact, only one implant failed and had to be removed. Among the 200 implants still in function at the end of the study, 195 were classified in the implant-crown success group, an implant-crown success of 97.5%. Five implants could not fulfill the implant-crown success criteria. One single implant showed some kind of sensitivity upon function. Two additional implants revealed a DIB >1.5 mm after the first year of function. Two implants, finally, failed to fulfill implant-crown success criteria, as their prosthetic abutments became loose during the first year of loading. These abutments were reinserted and no further loosening was observed. At the end of the study, the radiographic evaluation of the implants revealed excellent crestal bone stability, with a mean distance from the implant shoulder to the first crestal bone to implant contact (DIB) of 0.4 mm.

Conclusions

In our 1-year follow-up prospective clinical study, DLMF porous implants have shown a high survival (99.5%) and success (97.5%) rates. The use of DLMF implants seems to represent a successful procedure for rehabilitation of partial or completely edentulous patients. However, further long-term clinical studies will be necessary to investigate the potential of DLMF implants in restoring partially and completely edentulous arch in longer periods.

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