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Quasi-static strength and fractography analysis of two dental implants manufactured by direct metal laser sintering

Sergio Alexandre Gehrke DDS, PhD[1,2,3] | **Leticia Pérez-Díaz PhD**[4] | **Berenice Anina Dedavid PhD**[3]

[1]Department of Research, Biotecnos Research Center, Montevideo, Uruguay
[2]Department of Biotechnology of the University Catholica San Antonio de Murcia (UCAM), Murcia, Spain
[3]Department of Materials Engineering, Pontificial Catholic University of Rio Grande do Sul, Porto Alegre, Brazil
[4]Researcher of the Laboratorio de Interacciones Moleculares, Facultad de Ciencias, Universidad de la Republica, Montevideo, Uruguay

Correspondence
Prof. Sergio Alexandre Gehrke, DDS, PhD. Department of Research, Biotecnos, Cuareim, 1483 – Montevideo, Uruguay, CP 11.100.
Email: sergio.gehrke@hotmail.com

Abstract

Background: New manufacturing methods was developed to improve the tissues integration with the titanium alloy pieces.

Objective: The present in vitro study was to assess the resistance and fracture mode after applied a quasi-static compressive force on the two dental implants manufactured by direct metal laser sintering.

Materials and Methods: Twenty dental implants manufactured by direct metal laser sintering, using titanium alloy (Ti-6Al-4V) granules in two designs ($n = 10$ per group): Conventional dental implant (group Imp1) two-piece implant design, where the surgical implant and prosthetic abutment are two separate components and, the one-piece implant (group Imp2), where the surgical implant and prosthetic abutment are one integral piece. All samples were subjected to quasi-static loading at a 30° angle to the implant axis in a universal testing machine.

Results: The mean fracture strengths were 1269.2 ± 128.8 N for the group Imp1 and, 1259.5 ± 115.1 N for the group Imp2, without statistical differences ($P = .8722$). In both groups, the fracture surface does not present crack between the compact core and the superficial (less dense and porous) part of the implants.

Conclusions: Based on the measured resistance data for the two implant models manufactured by direct metal laser sintering tested in the present study, we can suggest that they have adequate capacity to withstand the masticatory loads.

KEYWORDS
dental implant, fracture mode, fracture strength, sintering implants

1 | INTRODUCTION

Macro- and micro-geometric alterations of dental implants, as well as materials and/or methods for their manufacture have been proposed to improve the success in terms of osseointegration and interaction with the pre-implant tissues. In this sense, several studies have identified that implant surfaces porosities and composition can positively influence cell behavior and therefore, bone apposition.[1–3] The treatment of the implant surface to promote irregularities, show superior molecules adsorption from biological fluids, improving early cellular responses, including extracellular matrix deposition, cytoskeletal organization, and tissues maturation and so, can lead to a better and faster bone response.[1,4] In this way, new manufacturing methods with the objective of obtaining porous titanium structure with controlled porosity, pore size, and location are being researched and developed.[4,5]

Initially, porous titanium implants were generally obtained using sprays techniques and coating on implant surfaces,[6,7] but this manufactured method may be reduced the fatigue resistance up to 1/3 when compared with standard uncoated implants.[8] In the last few decades, 3D printing/additive manufacturing (3DP/AM) technologies have become more and more important in the world of industry: these allow realizing physical objects starting from virtual 3D data project, without intermediate production steps, saving time, and money.[9–11] With 3DP/AM, porous titanium implants for medical applications can be fabricated. In fact, some high power focused laser beam fuses metal particles arranged in a powder bed and generates the implant layer-by-layer, with no postprocessing steps required.[10,11]

Innovations achieved through 3D laser printing for metals, also known as direct laser metal sintering (DLMS), while utilizing titanium powder in the field of healthcare are amazing. For decades, the healthcare industry not could imagine that additive manufacturing would become an appropriate tool for individualized titanium prostheses in the dentistry field.[9] These advances are evolving faster, enabling new and interesting ideas and fantastic innovations of the compact and porous sintering techniques.[12] The direct titanium alloys laser sintering is an additive metal fabrication process builds on the basic principles of 3D printing, essentially using the laser to melt or sinter layer per layer of metallic powder added concomitantly.[13]

These structures with controlled variable porosity can balance the mismatch between different elastic modulus of bone tissues and titanium implants, thus reducing stresses under functional loading and promoting long-term fixation stability and clinical success.[9,13] Conventionally, commercially pure (cp) titanium implants present a higher rigidity than surrounding bone because of Young's modulus (elastic modulus) of the material and the geometry of the structure.[9] Then, using the sintered 3D process to elaborate implantable structures, it is possible to fabricate these pieces with more similar physical values than the currently developed implants, which could result in a better interconnection between bone and implant.

It is well-known, that during mastication most of the forces are compressive in nature; therefore, it is of fundamental importance to investigate materials under this condition.[14] The compressive test is common to measure the resistance of a material, which is an important feature of metallic materials because it is the ability of the material to deform under tensile forces until the fracture moment and indicates the workability of an alloy.[15] After the mechanical test, an analysis of the fractured surfaces was carried out. The fractography provides a unique tool to determine potential cause of a fracture.[16]

Thus, the subjects of the present in vitro study, at quasi-static fatigue test, was to evaluate the resistance to fracture of two implants typical designs manufactured by DSML process and to compare with the human normal masticatory forces and the resistance to fracture of conventional dental implants related in the literature. In addition, use the fractography to investigate possible cracks between the core and the superficial (less dense and porous) part of the implants after the mechanical test.

2 | MATERIAL AND METHODS

2.1 | Materials

The implants were fabricated starting from powders of titanium alloy (Ti-6Al-4V) with a particle size of 25–45 ?m. The process of fabrication consists in layer-by-layer by an Yb (ytterbium) fiber laser system (EosyntM270, EOS GmbH, Munich, Germany), operating in an argon controlled atmosphere, using a wavelength of 1054 nm with a continuous power of 200 W at a scanning rate of 7 m/s and with the capacity to build a volume of 250 × 250 × 215 mm. Laser spot size was 0.1 mm. Post manufacturing of the implants, the both groups received the conventional procedures and treatment for commercialization. All pieces (implants, abutments, and screws) used in this study were manufactured by Leader Implants (Milan, Italy).

Twenty dental implants in two designs ($n = 10$ per group) were used and, separated as following: Conventional cylindrical dental implant with external hexagon connection (group Impl1) in 2-piece, where the surgical implant and abutment are two separates pieces, with dimensions at 3.75 mm in diameter and 10 mm in length for the implant and, a cementable titanium abutment, which was cut at 6mm in length and, then, fixed by a conventional titanium screw and torqued; for the other group, a cylindrical one-piece dental implant (group Impl2), where the surgical implant and prosthetic abutment are one integral piece, with dimensions at 3.75 mm in diameter and 16 mm in total length (10 mm of the implant + 6 mm of the abutment). All implants samples were manufactured by direct metal laser sintering (DMLS), which is a technology that allows fabrication of complex-shaped objects from powder-based materials, in accordance to a three-dimensional (3D) computer model. The implants design used are presented in the Figure 1.

2.2 | Mechanical quasi-static compressive test

The mechanical test was performed using quasi-static compressive forces to evaluate the fatigue strength of the dental implants in

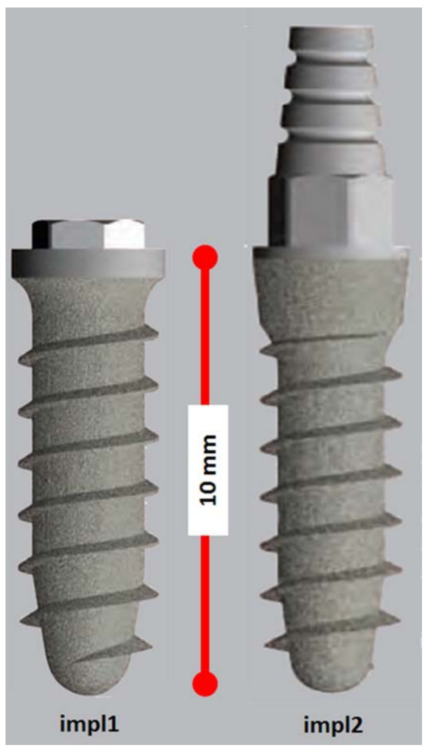

FIGURE 1 Images of the two dental implants manufactured by direct metal laser sintering. Impl1 = implant in two pieces and impl2 = implant one integral piece (implant plus abutment)

according to the International Organization for Standardization guidelines (ISO 14801:2016),[17] following these recommendations:

- Immersing the implants in an epoxy resin with a Young's modulus of elasticity similar to that of cortical bone;
- 3 mm of the exposed implant simulating a bone loss;
- positioning the implant in an angle of $30 \pm 2°$ with respect to the applied load;
- a cementable metallic crown was made with a hemisphere form;
- the final length of the abutment and crown was 11mm.

The Figure 2 showed the guidelines used in this study.

Cementable titanium abutments were selected and used for all implants of the group Imp1 and, all abutment screws received a torque of 25 N. The crowns were cemented on the cementable portion of both groups using conventional zinc phosphate cement. Ten implant specimens were used for each group.

According to the study design, all groups were subjected to quasi-static loading until fracture using a properly calibrated universal testing machine (model AME-5kN, Técnica Industrial Oswaldo Filizola Ltda, São Paulo, Brazil) with a test capacity of 5.0 kN. Tests were conducted at the Testing Laboratory of Biomechanics (Biotecnos, Santa Maria, Brazil) at a test speed of 1 mm/min.

After the test, all fractured samples were ultrasonically cleaned in 96% isopropanol and observed under low-power magnification. Digital photographs were taken using a Sony H9 digital camera (Tokyo, Japan), and the data were reported descriptively.

Statistical analyses of this data were performed using a Friedmann *t*-test analysis to determine the differences between the two groups. The comparison was conducted at the 95% level of significance ($\alpha = 0.05$).

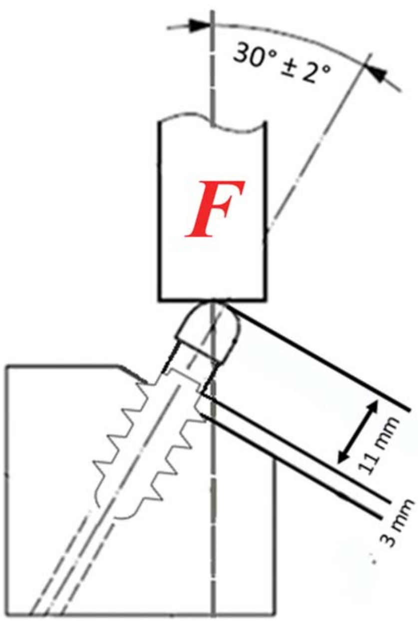

FIGURE 2 Scheme used in the compression test based on ISO 14801/2007 standards [18]. The distance between the red points (alpha) shows the extent of compression during the test

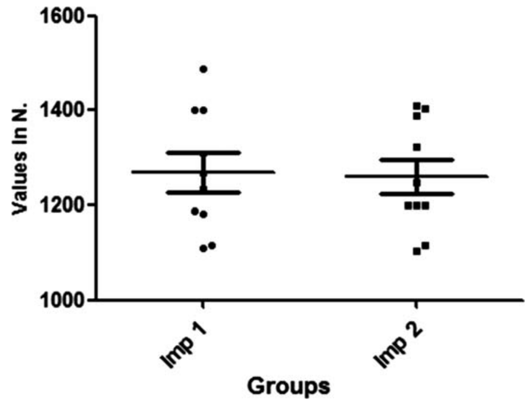

FIGURE 3 Average force required to overcome the resistance to quasi-static fatigue in the various two implants groups

2.3 | SEM fractography analysis

After the mechanical test, the microstructure analysis and morphological aspects of samples of each implant model were observed and interpreted on the images obtained using an Inspect F50 field emission-scanning electron microscope (FE-SEM) and XL-30 both of (FEI Company, Hillsboro, OR) operated at 5 kV.

3 | RESULTS

3.1 | Mechanical quasi-static compressive results

The fracture strength values of all samples were recorded during quasi-static loading and, the mean value and standard deviation was 1269.2 ± 128.8 N for the group Imp1 and, 1259.5 ± 115.1 N for the group Imp2, within the conditions proposed in the present study. The data distribution is presented in the points graph of Figure 3. For a *t*-test, the values measured indicates no significant difference ($P = .8722$), at a significance of $P < .05$. In the Figure 4, are presented the images of the two groups sets after the quasi-static load test, showing the fracture of the implants. In the impl1 group, all samples showed the fracture at ~3 mm of the implant platform. In the impl2

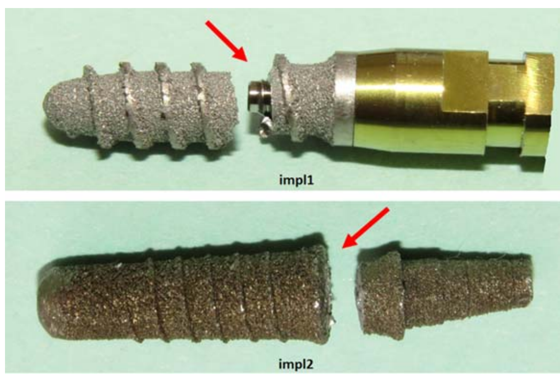

FIGURE 4 Images of the two groups sets that was submitted at the quasi-static load test, showing the fracture of the implants (red arrows). In the impl1 group, all samples showed the fracture at ~3 mm of the implant platform. In the impl2 group, all samples showed the fracture at corresponding to the base of the abutment

group, all samples showed the fracture at corresponding to the base of the abutment.

3.2 | SEM fractography results

All images obtained in SEM for the fractured samples were analyzed and the fracture mode was descripted. To facilitate the interpretation of the fractured the implant was considered divided into two parts, one compact core and one superficial less dense and porous part. It is observed that the porosity increases continuously in the core-surface direction. The representative fracture modes of the studied samples were presented in Figure 5 (group Impl1) and Figure 6 (group Impl2).

The Figure 5 shows the existence of crack between the particle and the solidified alloy. The particles appearance indicates the facture occurrence at quasi-cleavage planes, as expected of the rapid and unstable fracture, typical of as- sintering Ti-6Al-4V alloy (hexagonal close-packed).[16] No fractured particles were found in the outermost region of the implants examined, as expected. In the dense regions (near to the core) the bond between the molten mass and the particles is more consolidated than in the less dense regions. Then, it is correct to assume that a sintered layered material presents anisotropy in relation to the mechanical properties and the type of fracture.

The facture surface of the one-piece implant (group Imp2) is showed in Figure 6. In the core is observed a ductile fracture surface, and brittle facture occurrence in superficial less dense porous part of implants. The appearance of the fracture in the highlighted area of shows the quasi-cleavage planes occurrence, indicated the start of the fatigue crack. Fractures with dimples and micro void typical of ductile fracture are present.

4 | DISCUSSION

Currently, dental implants are considered a consistent and predictable form of treatment, with the failures concentrated in few patients,[18] and are widely used for prosthetic treatment in fully or partially edentulous patients. In situations where implant fracture occurs, it is impossible to repair the implant due to technical and physiological complications. The causes of fracture can be classified into three broad groups: (1) failure of the implant design or the used material; (2) an absence of passive adaptation of the prosthetic crown to the implant substructure; and (3) overload due to parafunctional habits. In the present study, the samples were manufactured used a new method (DMLS), which promotes an important structural alteration in comparison with the conventional dental implants manufactured using titanium bars. However, the results showed that this DMSL method, even with important porosities throughout its structure, showed high resistance values under the conditions that were tested.

This study proposed to examine the resistance to the static fatigue of implants with 2 designs of dental implants and found significant differences between the implant types. In accordance with ISO 14801:2007,[17] the set were positioned at an inclination of 30° with 3 mm of the cervical implant portion not inserted, reproducing bone loss in that area. Referring to the cervical portion inserted or not, Gehrke and collaborators (2014),[19] related that the loss of bone support around the cervical portion of the implants cause an important reduction of the set resistance. In the present study were used an implant of two pieces (group Imp1) with external connection, which showed a resistance similar of the implant produced by conventional method (machined bars) and related in the anterior referred study.[19]

Both implant models studied show distinct surfaces, although the manufacturing processes used are the same. The implant surface (group Imp1) is more porous than the one-piece implant (group Imp2). In addition, the surface of the one-piece implant (group Imp2) shows typical dendrites of liquid solids transformation. Therefore, the fracture of the one-piece implant (group Imp2) is fragile at the points that bind the layers; the molten part apparently resisted the force applied as expected. Conversely, the implant (group Imp1) which has a very discrete melting surface, showing unfused particles, shows the rupture of the layer as a whole with ductile appearance. Despite the differences in morphology and fracture type, the implants present minimal differences in the results of the mechanical tests, within the conditions proposed in the present study.

The implant diameter relative to the dimension of the supporting bone is critical for successful treatment.[19] The average maximum bite force for adults in the premolar and molar regions is 789 N for men and 596 N for women.[20] In our study, fracture strength after static loading of the specimens was significantly higher for both groups, demonstrated values close to those of previously reported masticatory forces.[20,21]

The fatigue test established by ISO 14801:2007 is an extremely important method of evaluating dental implants.[17] These guidelines enable mechanical analysis of the samples with the intention of mimicking clinical behavior. This study used static implant fatigue testing for two different implants design, and demonstrated that implant strength is equal fabricated in one or two pieces. While other meaningful results have been reported such as chewing simulation or fatigue loading studies of implant abutment systems,[22,23] clinical trials are necessary to validate the results of these investigations as well as those of the present in vitro study. The present in vitro study presents numerous limitations, mainly regarding the dynamic movements received on these structures implanted during the masticatory loads, which were not reproduced in this type of quasi-static test used. However, it is of paramount importance for the clinician to know the limitations of the products that will be used in their patients.

5 | CONCLUSIONS

Within the limitations of this in vitro study and, based on the measured resistance data for the two implant models manufactured by direct metal laser sintering, we can suggest that they have adequate capacity to withstand the masticatory loads. Furthermore, the design of implants (one or two pieces) not change the performance and resistance of the system on the conditions tested. No cracks were found at the interface between the core and the superficial surface of the implants analyzed under 1269.2 and 1259.5 N load pattern, respectively for group impl1 and impl2.

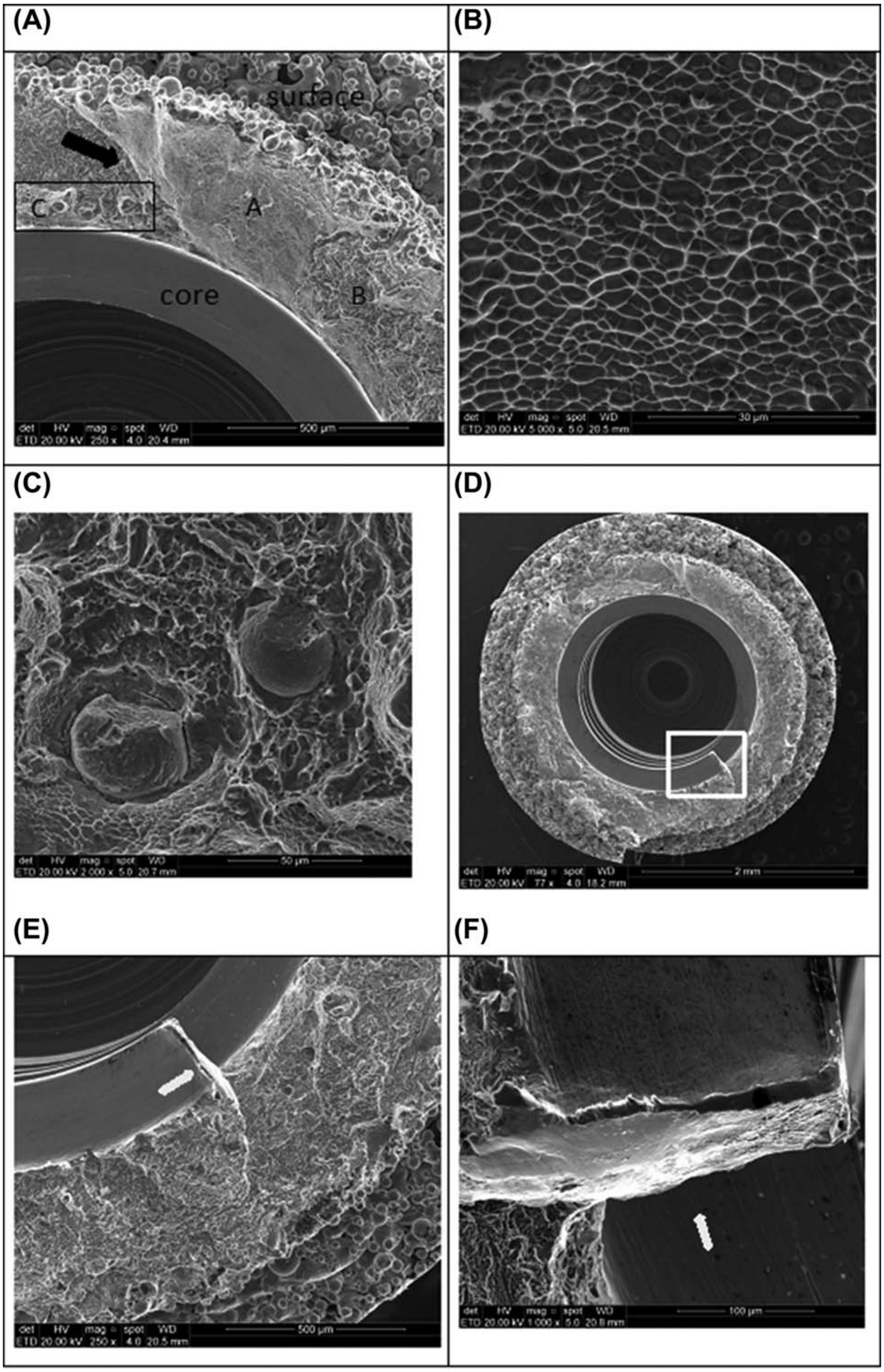

FIGURE 5 FEG-SE images of two-piece dental implant (group Imp1) after fracture. A, Fracture surfaces of the implant specimen at magnification (250×). The appearance of superficial part is like a river solid of the volcanic lava with spherical particles and large voids, resulting from the titanium alloy melting in the process. The voids and spherical particles of diameter range 5–50 μm, remaining from the powder used for fabrication enveloped in the remainder melting alloy at surface part. The area (A) show a flat and quasi-smooth surface formed by homogeneous dimples which indicated ductile overload fracture at this location. Note that the flat region extends along the core, indicating the start of superficial region rupture, and the crack outset in the core. B, Typical dimples of the fractured area (A) at superficial part at high magnification (5000×). The arrow indicates facets typical of fracture per cleavage. Note that appearance of fracture surface in area (B) is a mix of dimples, facets, microvoids and flat parts. C, Facture appearance of the spherical particles. D, End of core fracture of the implant, in highlighted. Note that fracture begins in the implant surface and propagates radially through the core, subsequently rupturing the adjacent surface region. E, Shows aspect of the striations in core surface. The arrow indicates that the fracture occurred by shearing opposite the fracture angle, typical of the cyclic loading. F, Indicates that the implant shows a ductile fracture in the transversal and, fragile in the longitudinal, characterizing the behavior expected for consolidated powders pieces, in layer by layer 3D processes

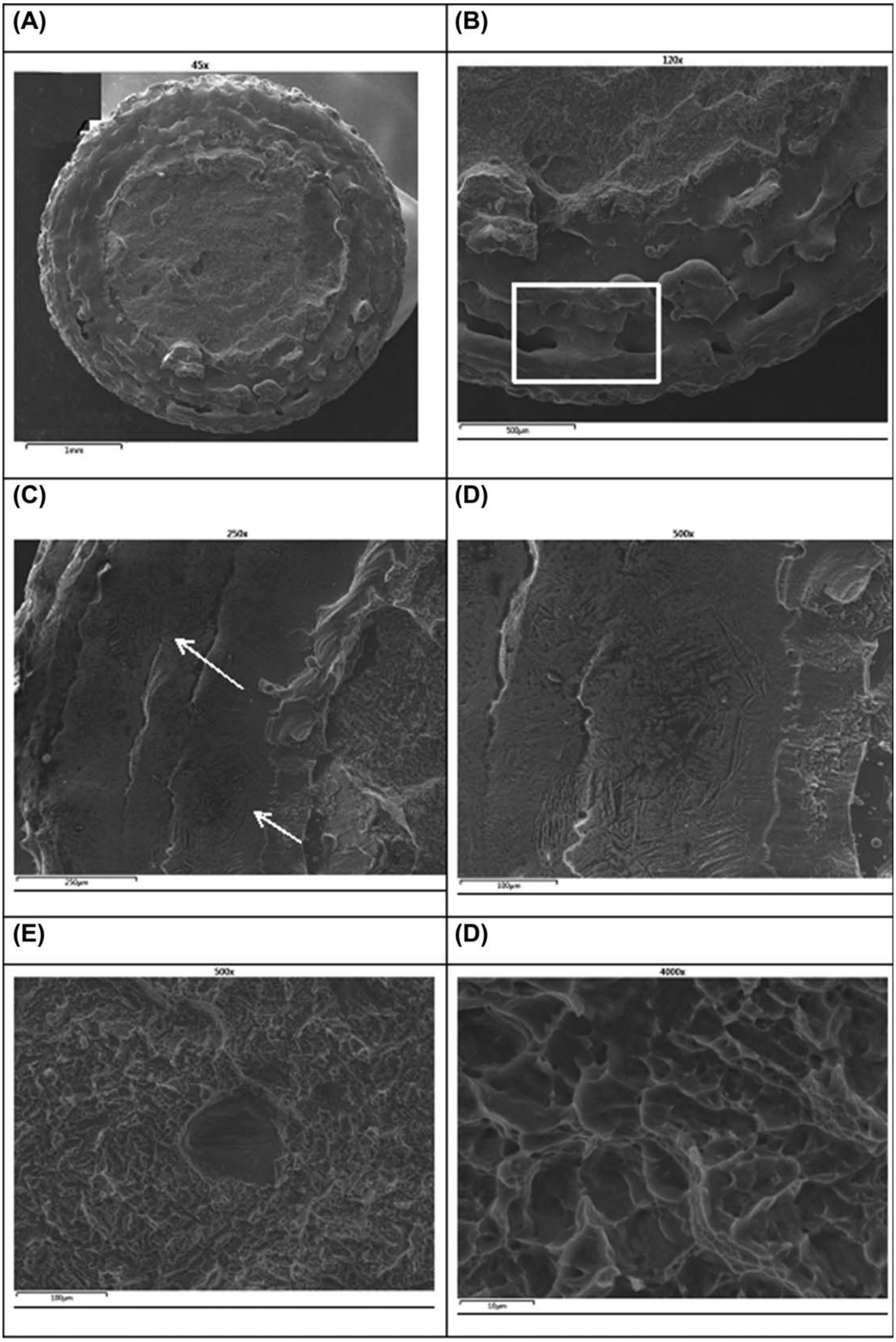

FIGURE 6 SEM-SE images of the one-piece implant (group Imp2) as fractured. A, Fracture surface of the implant at magnification (45×). B, Image shows details of fracture in higher magnification (250×). The highlighted area shows a brittle fracture. C and D, The image shows details of the superficial area of the implant. The surface shows typical dendrites of liquid solids transformation. E and F, Images of core fracture show dimples and micro void typical of ductile fracture

CONFLICT OF INTEREST

The authors report no conflict of interest

ORCID

Sergio Alexandre Gehrke DDS, PhD http://orcid.org/0000-0002-5863-9101

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