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REVIEW ARTICLE

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Biomaterial and biomechanical considerations to prevent risks in implant therapy

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1 | INTRODUCTION

Dental implant therapy is today widely accepted as one of the reliable options for restoring missing dentition.[1] The basic concept of implanting metallic devices in bone was first described by Bothe et al[2] in the 1940s, followed by Leventhal et al[3] in 1951, who described titanium as a potentially biocompatible surgical implantable material. Brånemark[4] later reported that the metallic microscope (titanium optic chamber) made of titanium to observe microvascular circulation in living bone could not easily be retrieved after the experiments. The external geometry of the titanium optic chamber possessed threads so that they mechanically engaged into the bone. It was not in their plan that the titanium microscopes integrated firmly into the bone; however, this coincidence inspired the research group to apply this concept to dental implants with hopes that they would functionally restore the dentition.

Titanium and its alloys are regarded as bioinert or biocompatible materials and are stable in the body owing to the spontaneously formed oxide layer.[5] Biomaterial research regarding dental implants has dominantly utilized titanium as a material of choice, and both basic and clinical research show that commercially pure titanium and several other titanium alloys are osteoconductive and promote osseointegration.[6]

The threaded-type implants were further tested in edentulous patients and the clinical outcomes were presented at the Toronto Congress in 1982.[7] The 15-year survival presented surprised everyone attending, and since then the application of osseointegrated dental implants has become a major alternative to restore full/partial edentulism.

Certain prerequisites have been proposed as essential factors for successful osseointegration,[8] and the development and evolution of the dental implants progressed based on these factors. In the past, research on each individual factor has been performed and a plethora of evidence has been established. In particular, research on implant surface topography/chemistry has been of major focus, and new surfaces are constantly applied on implants in an attempt to accelerate osseointegration rates.[9] As a result, new implant surfaces are often launched in the market with claims that the implant would osseointegrate faster.

Recently, the interplay between factors such as implant macrogeometry, topography, and surgical protocols has been of interest.[10,11] Evidence suggests that enhanced establishment of osseointegration may not be achieved by a single factor.[11] For instance, if the implant surface possessed state-of-the-art surface features for increased osseoconduction, implants may not present better osseointegration unless other factors generate a host bed for the surfaces to interact with osteogenic cells.[11,12]

This review presents an overview of the current existing evidence on osseointegrated implants. Factors such as implant microgeometry and surface micro- and nanotopography will be introduced, and a critical discussion regarding how they interact with each other is provided. Of special interest, and less explored in reviews, is how surgical instrumentation, drilling protocols, and drilling methods can be taken into consideration as important factors affecting osseointegration. Further, the biomechanical aspect of the titanium bulk itself, which had not been discussed much in the past, will be discussed as a possible factor for periimplant bone loss and, in the worst scenario, implant failure.

2 | IMPLANT HARDWARE AND OSSEOINTEGRATION HEALING PATHWAYS

Though a robust body of literature has described implant host response as a function of time, a large variation in surgical device design, the proportional variation in surgical instrumentation techniques, and the increasing focus of research on implant surface engineering, the attention has eventually shifted to the impact of the implant hardware features in bone healing pathways.[10] Although it is acknowledged that implant macrogeometry and the associated surgical technique (and thereby implant hardware) are different parameters, it has been demonstrated that their interplay not only leads to different osseointegration healing pathways but also affects how effective micrometer- and nanometer-scale designing should ultimately be.[10,11,13,14] The three osseointegration pathways have been described in detail in previous work,[10,11,13,14] and so the main effects on early- and long-term healing and biomechanical evolution are summarized.

2.1 | Interfacial remodeling osseointegration pathway

When implant placement results in an intimate contact between bone and the device's threaded bulk, consequently rendering high insertion torque values (tight fit) due to mechanical interlocking, an interfacial remodeling bone healing pathway takes place.[10,11] This scenario is observed in the majority of marketed implant systems, where insertion torque levels are chiefly dependent on implant geometry and micrometer-level surface modifications as well as osteotomy dimensions, which will eventually regulate the strain distribution in bone. Although debatable, higher primary stability has been associated with the clinical perception of higher rotational resistance during implant insertion, expressed by insertion torque levels.[15] However, high insertion torque levels result in strain and microcrack formation in bone, leading to compression necrosis that triggers bone remodeling. This scenario has been theoretically suggested,[16] and experimentally confirmed,[17] by the loss of mechanical interlocking between bone and implant due to extensive bone interfacial remodeling after implant placement under stable condition. Through the course of remodeling at the implant-bone interface, newly formed woven bone and the subsequent evolution toward a lamellar configuration will eventually render the system with secondary (biologic) stability that will support the implant device throughout its lifetime. When evaluated over the long term (5-10 years) the interfacial osseointegration pathway results in a compact, mature lamellar bone with small marrow spaces.[18,19] Figure 1 presents the main histologic features of the interfacial remodeling healing pathway.

2.2 | Healing chamber (intramembranous-like) osseointegration pathway

Differently from implant bulk and drilling dimension combinations that lead to high degrees of contact between implant and bone immediately after implantation, healing chamber implants are devices that present plateaus instead of threads, and are tapped into bone osteotomies with the diameter of the implant.[20] Such interplay between device and osteotomy dimension allows for minimal primary stability and results in large void spaces between the device and bone, allowing the formation of healing chambers.

Immediately after placement, the healing chambers will be filled with a blood clot that evolves toward an osteogenic connective tissue configuration that will ossify through an intramembranous-like pathway.[21] This osteogenic connective tissue is highly vascularized at early implantation times, allowing osteogenic cell migration throughout the healing chamber space.[21] Such a vascularization and cell migration pattern allows for woven bone formation simultaneously

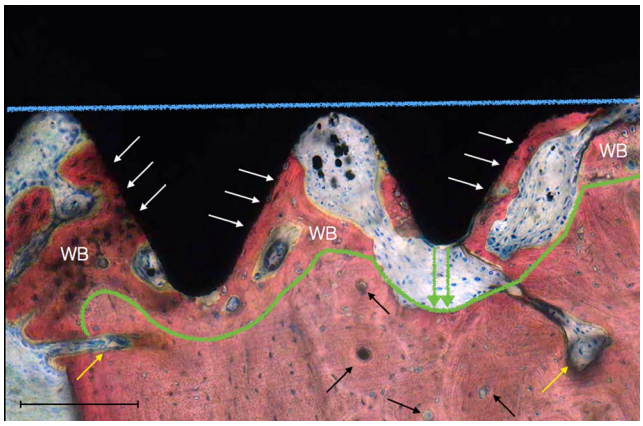

FIGURE 1 Optical micrograph depicting the main features of interfacial remodeling healing pathway at 6 weeks in a canine model. The blue line represents the theoretic position of the osteotomy outer diameter closely matching the implant threads inner diameter resulting in implant high insertion torque at the expense of interfacial bone compression. The green line demonstrates the perimeter and extension (green arrows) the cortical shell remodels (native bone osteonic structures, black arrows), and that woven bone growth occurs primarily through osteogenic cell migration from the native bone toward the implant surface that acts as a site for bone growth. The yellow arrows depict bone remodeling sites occurring farther away from the bone-implant interface. Bar, 150 μm

taking place at the implant surface, osteotomy bone walls, and within the center of the chamber, resulting in rapid device osseointegration. Shortly after early integration through woven bone formation, lamellar bone will gradually replace the woven bone, resulting in formation of primary osteonic structures. A series of human retrieval studies of plateau root form implants has consistently demonstrated that further bone morphologic evolution occurs toward a haversian-like configuration[22-25] that over time increases significantly in mechanical properties.[25] Figure 2 presents the main histologic features of the healing chamber healing pathway.

2.3 | Hybrid healing osseointegration pathway

The hybrid healing osseointegration pathway has been utilized in an attempt to obtain devices that are atemporally stable and is a combination of features presented in the interfacial remodeling and healing chambers osseointegration pathways. Screw-type implants with large thread pitch and outer to inner thread diameter differences have been designed to allow surgical instrumentation sufficiently large for the formation of healing chambers between threads, implant inner diameter, and bone instrumented walls.[26-31] For such an implant design configuration, primary stability is obtained by the interaction between the outer regions of the threads that engage bone and such initial stability is proportional to the thread design and the amount of mismatch between implant outer thread and osteotomy diameters. Hence, this osseointegration pathway simultaneously presents bone remodeling where engagement between implant and bone occurred, resulting in stability loss that is supposedly compensated by the rapid woven bone formation in the healing chambers formed between threads. Unlike screw-type implant systems that result in interfacial remodeling osseointegration and plateau root form implants that osseointegrate through healing chambers, implant systems deliberately designed for this purpose are relatively new in the market.[27-31] Thus, long-term human retrieved samples are not yet available for adequate assessment of the effect of this healing pathway on long-term bone morphologic evolution. Figure 3 presents the main histologic features of the interfacial hybrid healing pathway.

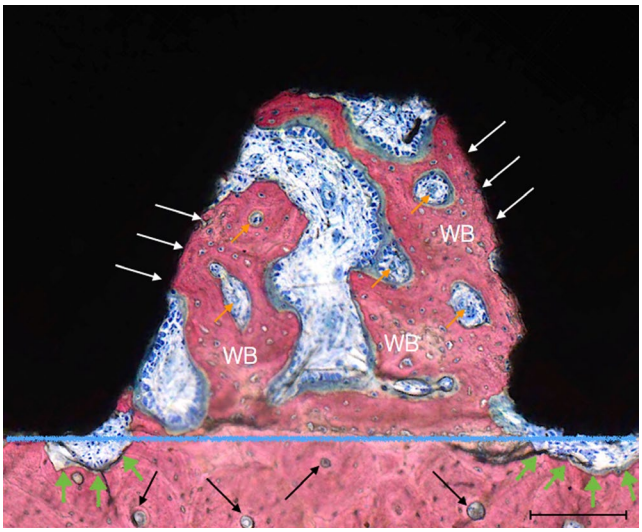

FIGURE 2 Optical micrograph depicting the main features of the intramembranous-like (healing chamber) healing pathway at 6 weeks in a canine model. The blue line represents the theoretic position of the osteotomy outer diameter closely matching the implant plateau outer diameter resulting in implant minimal compression in the bone walls. The green arrows demonstrate cortical shell remodeling due to initial compression due to implant-bone interaction (native bone osteonic structures, black arrows). Woven bone growth occurs from the instrumented native bone surface toward the healing chamber center, from the implant surface (white arrows), as well within the healing chamber central region. The orange arrows depict bone remodeling sites that at this time point are starting to replace woven bone by lamellar bone and will be the site of primary osteonic structures within the chamber. Bar, 150 μm

3 | SURGICAL INSTRUMENTATION

A critical appraisal of the literature will be presented in this section to suggest that modest interest and advances have been achieved in the subject of surgical instrumentation since the introduction of

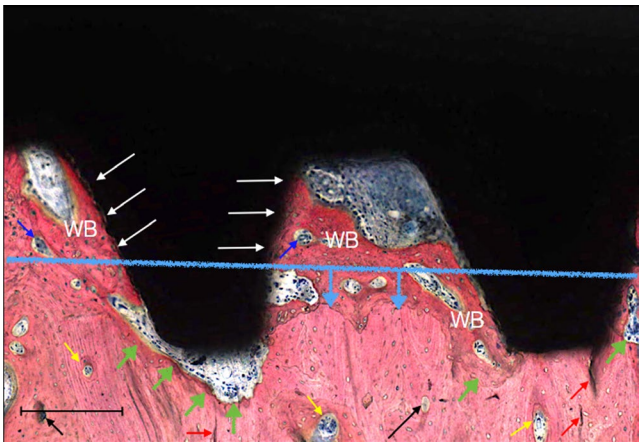

FIGURE 3 Optical micrograph depicting the main features of the interfacial hybrid healing pathway at 6 weeks in a canine model. The blue line represents the theoretic position of the osteotomy outer diameter between the implant threads' inner and outer diameter resulting in implant high insertion torque (at the expense of bone compression by the thread that results in interfacial remodeling depicted by green arrows and microcracking depicted by red arrows) along with the formation of a healing chamber between the instrumented bone wall and the surfaces between implant threads. Worth noting is the extensive woven bone formation from the native bone surgically instrumented surface (retracted from remodeling due to surgical instrumentation trauma—extension depicted by light blue color arrows) toward the center of formed healing chamber, from the implant surface (white arrows), as well within the healing chamber central region. The dark blue arrows depict bone remodeling sites that at this time point is starting to replace woven bone by lamellar bone and will be the site of primary osteonic structures within the chamber. Bar, 150 μm. The black arrows depict native bone cortical shell osteonic structures; yellow arrows depict bone remodeling sites occurring farther away from the bone-implant interface. Bar, 150 μm

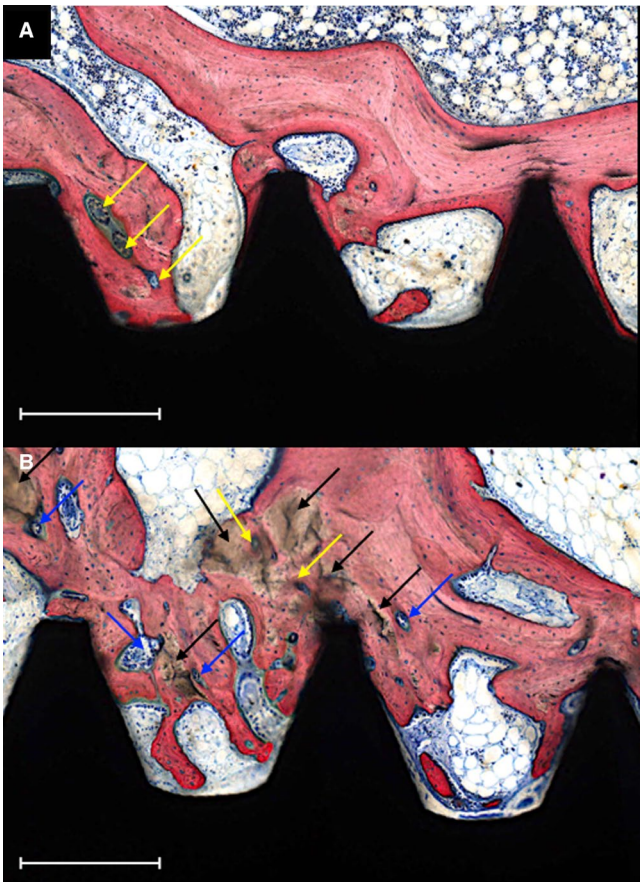

FIGURE 4 Optical micrograph depicting the main features of subtractive drilling and counterclockwise osseodensification in a low-density sheep hip model at 6 weeks in vivo. A, Uneventful osseointegration takes place through subtractive drilling, where new bone formation occurs through contact osteogenesis from cells migrating toward the implant surface due to continuity between native bone and implant surface, as well as from bone native tissue. Yellow arrows depict remodeling sites initially replacing woven bone by lamellar bone. B, Counterclockwise osseodensification drilling resulted in the presence of autograft particles (bone chips depicted by black arrows) in proximity to the implant surface. These bone chips did not preclude cell migration toward the implant surface and acted as nucleating surfaces that bridged the implant surface with native bone. At 6 weeks in vivo, extensive bone chip remodeling was observed either in proximity with the bone chip surface (blue arrows) or directly through bone chip bulk (yellow arrows). Bar, 300 μm

implant therapy. Numerically, preclinical and clinical studies investigating the effects of surgical instrumentation on osseointegration are at least one order of magnitude lower than the most investigated surface treatment effects. In addition, the available literature on the topic is contradictory, and commonly debated from an anecdotal perspective, as dental implant surgical instrumentation has been largely employed based on textbook knowledge rather than on informed research platform. While unequivocally high clinical success rates have been achieved through the classic protocol of using several subtractive bone drills of increasing diameter at moderate drilling speeds under copious saline irrigation, recent preclinical studies have pointed out that substantial work is required to determine the optimal drilling method that maximizes osseointegration.

For example, a series of previous studies have shown that the simplified version (pilot drill immediately followed by the final diameter drill for implants of 3.75, 4.2, and 5 mm) of the traditional gradual drilling expansion results in comparable osseointegration relative to classic protocols.[32-37] Recent work has confirmed no statistical difference on osseointegration parameters between simplified and conventional protocols.[32] Alternative to reducing the number of drills, alteration in drill design to include progressive diameter achievement through the employment of a single progressive drill has been demonstrated to not decrease early osseointegration.[38] From a surgical perspective, the total surgical time of the simplified protocol from incision to closure would be significantly shortened and lead to less postsurgical complications.[39] However, the reduction in the number of drills used in the simplified protocol would have the drawback of reducing the number of attempts to correct location or angulation of final implant placement.

Drilling speed is an important factor of the surgical technique and has an influence on the temperature in the surrounding bone. Excessive drilling speed has been reported to prevent adequate irrigation, thus potentially allowing for heat generation that could lead to thermal osteonecrosis.[40-42] On the other hand, studies have suggested that low drilling speed can potentially generate more heat when compared with high drilling speeds since the surgeon would possibly apply not only more vertical compression during drilling, but also drill for longer periods of time.[43,44] Furthermore, it has been suggested that low drilling speed may potentially increase wobbling and result in overpreparation of the osteotomy site.[45,46] Despite the literature suggesting potential issues with decreasing drilling speeds, a recent study has pointed that decreasing drilling speeds of 1000, 500, and 100 rpm all resulted in successful osseointegration.[45] Several other studies have pointed out that lower speed drilling did not decrease osseointegration levels when compared with the most usually employed higher drilling speeds.

It should be noted that the work referenced in this section is system specific and extreme caution should be taken, since substantial deviation in drilling recommended dimensions and speeds exists between various implant systems. Nonetheless, these studies present variations in subtractive drilling techniques utilizing each individual system's drilling methods that may or may not improve osseointegration.

Whereas the traditional osteotomy using different step drills is generated by a subtractive method, where bone is drilled away (Figure 4A), a recently developed surgical technique for site drilling has suggested to enhance implant stability via osseodensification drilling (Figure 4B).[47,48] The concept of this technique is to execute a nonsubtractive multistepped drilling process, which results in the densification of the osteotomy site wall, creating an environment that enhances primary stability due to compaction-autografting that generates a springback effect that allows bone tissue recoil back into the implant surface.[49] The compaction autografting and the

springback phenomena have been documented in orthopedic literature,[49-52] and its effect on increasing implant primary stability is different than the scenario observed in traditional subtraction drilling, where insertion torque levels are mainly dependent on the osteotomy downsizing dimensions. The densification of the osteotomy wall and the presence of residual bone chips establishes immediate contact between the implant system and bone structure, which creates stability via physical interlocking, and induces osteoblast nucleation on the instrumented bone surrounding the implant, thereby accelerating bone healing in the proximity of the implant.[53-55]

The classical drilling technique involves a positive rake angle, which is used to extract a small amount of bone as each flute passes to create an osteotomy cleared of bone residue prior to implant placement. In contrast, the osseodensification drilling process utilizes a tapered, multi-fluted bur to create the osteotomy site. This bur design contains at least four tapered flutes with a negative rake angle, which allows for the creation of a layer of compact, dense bone that surrounds the wall of the osteotomy. The densifying bur is capable of controlling the expansion process due to the features of the cutting chisel and tapered shank, which allows the bur to progressively increase in diameter as it drills deeper into the osteotomy site. The expansion process can be operated in both clockwise and counterclockwise drilling directions and is performed at high drilling speeds under irrigation.[55] The counterclockwise drilling direction is more efficient at the densification process and is utilized in bone with low density, whereas the clockwise drilling direction is better suited for higher density bone.[55] The osseodensification surgical technique has been extensively verified in bench top in vitro[56] and in vivo animal[54,55,57] studies. More recent translational, large animal preclinical models have unequivocally shown significant biomechanical improvements measured through implant primary stability and biomechanical pullout tests relative to subtractive drilling controls. These studies have also shown, relative to subtractive drilling controls, significantly higher osseointegration for implants placed by clockwise and counterclockwise osseodensification drilling methods.[54,55]

The improvement of quality/quantity of bone surrounding the implant to increase primary stability is a theory that has been previously explored, but it mainly focused on improving primary stability after maxillary sinus elevation.[58,59] The osteotomy technique involves the compression of the surrounding bone through an impaction, where improved primary stability is perceived by clinicians via increased insertion torque values. Currently, there is insufficient evidence supporting this surgical technique as a superior surgical approach compared with other techniques currently available.[60]

Surgical instrumentation, in summary, contributes to both the achievement of initial stability and secondary stability. The degree of mechanical interlocking of the implant to the bone can be controlled through different surgical techniques, as introduced in this section. Moreover, it has been shown that secondary stability (osseointegration) can be promoted through these techniques. Interestingly, techniques that were considered to be more invasive than the conventional ones, such as osseodensification, proved to be effective in enhancing bone healing to the implant.

It can be speculated that different surgical instrumentation techniques can create more favorable conditions to promote both primary stability and osseointegration of implants. The osseodensification technique proved to be effective in improving the biomechanical properties and secondary stability of implants in translational studies, opening an interesting insight on the role of surgical instrumentation for stability, integration, and an ultimately better outcome of osseointegrated implants. Despite the encouraging result, however, the osseodensification technique remains almost the only instrumentation method that has been investigated through a systematic scientific process, leaving a substantial void in the understanding of the instrumentation role in the implant integration and success. Further research studies are necessary in order to explore different instrumentation methods, by both bone-subtractive and bone-densifying techniques, as well as alternative osteotomy methods (eg, by piezosurgery), in order to identify the ideal integration between instrumentation protocol, implant geometry, and surface as well as host characteristics (density, morphology, structural, and biologic features of the alveolar bone) to optimize short- and long-term osseointegration.

4 | SURFACE TREATMENT AT THE MICRO- AND NANOMETER LENGTH SCALES: IMPORTANCE AS AD HOCS FOR OSSEOINTEGRATION

Whereas the field of implant surface engineering has been extensively explored over the last decades,[61,62] especially the modifications at the micrometer level, the rationale presented previously in this paper will unequivocally demonstrate that surface modifications will better perform when implant hardware is designed to allow immediate host interaction with the surface.[10,11] Also, a correlation between the potential benefits and the current contribution of nanotechnology to implant therapy will be presented.

Implant surface treatment at the micrometer scale has been the center of focus over the past three decades. This was based on substantial in vitro and in vivo evidence that moderately microtextured surfaces promote osseointegration compared with other surfaces.[9,36,63-65] As a result, the majority of the commercially existing implants on the market possess these moderately microtextured features. Moreover, it has been suggested that the application of nanofeatures on surfaces enhances the osteogenic gene expression,[66-68] which in some in vivo studies resulted in enhanced osseointegration.[69] It has been further proven that nanostructures are further effective when they exist within hierarchic structures.[11,70] However, in some other studies, the effects of nanofeatures were not significantly evident, although some tendencies indicate the benefits of the nanotexturing.[71,72] Numerous reasons for the less effective results could be speculated on; however, one of the major factors that seems to influence the degree of effectiveness of the applied hierarchic micro- and nanotextures is how surgical instrumentation is performed and which type of macrogeometry is applied

on the implant.[11] Depending on these factors, the surrounding bone follows different healing pathways that affect which cells will first interact with the implant surface.[10,11] Thus, given the free migration pathway between implant and native bone created by the osteogenic tissue evolved from the blood clot within healing chambers, higher efficiency in hastening bone formation is expected for this configuration, compared with scenarios of interfacial remodeling, where a cell-mediated bone resorption is expected prior to bone formation.[10,29] In hybrid healing, where initial interlocking between implant threads and native bone ensure primary stability, the presence of healing chambers, hierarchically incorporating micrometer- and nanometer-level texture for early interaction between implant surface and plasma, cells and proteins, will result in rapid bone formation; this will maintain implant stability while interfacial remodeling occurs due to compression between implant threads and native bone. Such a scenario, where concerted bone loss is compensated by rapid bone formation within chamber, may likely result in an implant that is atemporally stable from a biomechanical standpoint.

5 | IMPLANTS AND CUMULATIVE DAMAGE DEGRADING THEIR STRENGTH

The two primary types of complications that may occur with implants are biologic (eg, peri-implantitis, implant loss) and mechanical (eg, abutment screw fracture, prostheses veneering material fracture, implant fracture), with biologic issues generally taking place earlier than mechanical ones.[73] Peri-implantitis is a major and growing problem, as it affects the tissues surrounding dental implants likely leading to implant loss if untreated.[74] Almost all implant systems are susceptible to biologic complications, and a Swedish clinical trial involving 588 patients revealed the presence of 45% of peri-implantitis (bleeding on probing, suppuration and bone loss more than 0.5 mm) and another 14.5% presented severe peri-implantitis (bleeding on probing, suppuration and bone loss more than 2 mm) in a retrospective and cross-sectional clinical and radiologic examination of patients approximately 9 years after treatment completion.[75] However, as time in function elapses, the opportunity for crack initiation and growth increases, which may be accelerated by bone loss, with consequent exposure of implant neck and threads. It has been reported that 62% of implants retrieved due to peri-implantitis presented several flaws in the thread and neck region, evidenced by the presence of full cracks (more than 0.5 mm crack extension) and cracklike defects (25-100 μm length).[76] Such cracks, detected in human retrieved implants, predominated in commercially pure titanium grade II compared with Ti-6Al-4V implants and have been reported to be the embryos for fatigue crack nucleation that eventually leads to implant fracture.[77] Thus, it is indisputable that the strength degradation process leading to failures of the restored implant system is chiefly regulated by fatigue,[78] and that implant bulk biomaterials presenting improved fatigue-resistance properties are desirable for long-lasting rehabilitations. With patients using implants for increasing lifespans, the understanding of maintenance and failure issues has been explored more often and reported on.[79]

Whereas it is desirable to discuss the performance of restored implants based on clinical trials, several challenges are encountered that hinder sound comparison between studies. The limitations include the variation in success criteria and/or in selected primary outcome, reduced follow-up times and patient sample, variations in prosthetic rehabilitation scenarios including implant-abutment connection design, prostheses type and materials, antagonist, and so on. Therefore, controlled laboratory fatigue testing, while also having its limitations, can be used to simulate a variety of clinical conditions and serve as a baseline to design proper clinical trials. An emphasis will be presented on the cumulative damage that inexorably emerges once a prosthesis is in function, regardless of implant system and restoration type.

6 | IMPLANT BULK MATERIAL: TITANIUM AND ALLOYS, CERAMICS, AND POLYMERS

Metals, ceramics, or polymers, or a combination, can be used to fabricate the dental implant bulk material. Although titanium has been promulgated as a material capable of "growing bone in contact with it" since the 1940s,[2] and despite its broadest use amongst implantable metallic devices, the quest for end-stage materials that mimic tooth structures in color and some mechanical properties continues. Ceramic implants were introduced over four decades ago for orthopedic applications, made of aluminum oxide, with a current robust clinical follow-up of implanted patients. Owing to quite high failure rates only for its early versions (up to 13% when fabricated before 1990s), alumina is currently recommended for less-demanding loading scenarios.[80] Reports on the use of alumina dental implants were from short term clinical studies,[81] but the increased fracture toughness and transformation toughening mechanism of yttria-stabilized zirconia[82] led to its use in the 1990s for orthopedic femoral heads and later on in dentistry. Because of critical events in 2001 leading to 400 yttria-stabilized zirconia femoral head fractures associated with two batches, and due to evidence of progressive aging degradation even under normal conditions, the long-term stability of zirconia still appears to be an issue.[83] Whereas the development of more stable polycrystalline ceramic composites with even higher mechanical properties, such as zirconia-toughened alumina and alumina-toughened zirconia, has been witnessed in orthopedics since the last decade,[84] ceramics for dental implants are ironically focused on yttria-stabilized zirconia. For implant or tooth-supported reconstructions, yttria-stabilized zirconia phase transformation has indicated the presence of monoclinic phase content from 2.13% up to 81.4% after aging, suggesting an outrageous variation in yttria-stabilized zirconia metastability.[85] The recent introduction as a dental implant biomaterial candidate is the high-performance thermoplastic polymer polyetheretherketone, which has been enthusiastically brought to implant dentistry motivated by its tooth-colored appearance and gradient difference in elastic modulus (100-110 GPa for Ti, 200 GPa

for yttria-stabilized zirconia, and 3.6 GPa for polyetheretherketone up to 18 GPa for carbon-reinforced polyetheretherketone, which is dark in color). The performance and evidence for use of titanium, ceramics and polyetheretherketone as a dental implant biomaterial will be discussed.

6.1 | Rationale for implant biomaterial selection

Implant bulk biomaterial choice is paramount due to longevity concerns and especially considering that a clear trend in implant dimension reduction (width and height) has been noticed from early to current implant dentistry. Previously available implant dimensions, such as long implants (equal to or more than 13 mm),[86] have undergone a steady decrease in market share over the last few years since advances in implant hardware and software engineering have led to comparable survival of both short (more than 6 mm but <10 mm)/extra-short (<6 mm) and narrow (more than 3 mm but <3.75 mm)/extra-narrow (<3 mm)[86] diameter implants relative to standard ones when properly indicated.[87] In addition, certain restorative approaches technically obliging the use of long implants, such as bicortical anchorage, resulted in three times more implant fractures compared with techniques confined to monocortical anchorage when evaluated in the long term (up to 15 years).[88] Therefore, considering the increasing launch of short and narrow-diameter implants where implant walls and an overall decreased area for stress distribution is observed, the implant biomaterial becomes more challenged and should present higher mechanical properties for improved survival. From an overall longevity perspective, improved mechanical properties may be desired for the implant biomaterial, regardless of length and dimension.

6.1.1 | Titanium and alloys

Commercially pure Ti (grade II, ASTM F67) has debuted as the material of choice in modern implant dentistry and has traditionally been used for several years. Some alternatives with higher mechanical properties, including commercially pure Ti grade IV (ASTM F67) and Ti-6Al-4V (ASTM F136), were later introduced as implant biomaterials. The ultimate tensile strength is approximately 345 MPa for commercially pure grade II, 550 MPa in commercially pure grade IV, and 930 MPa in Ti-6Al-4V.[89] Whereas Ti grades and its alloys' mechanical properties alone should not be used as performance predictors, mechanical testing of restored implants has shown a significant improvement in survival of Ti-6Al-4V implants compared with commercially pure grade II.[90] When a selection of restored narrow implant systems was evaluated, a commercially pure grade IV 3.5 mm implant presented significantly lower characteristic strength than other 3.4 mm Ti-6Al-4V systems, whereas a 3.3 mm TiZr system (approximately 85% Ti and 15% Zr) showed intermediate values.[91,92] Grade IV commercially pure Ti can be further enhanced by cold working, which showed that 2.9 mm extra-narrow diameter implants resulted in similar survival to 3.3 mm narrow implants.[93]

From an osseointegration standpoint, recent biomechanical and histometric in vivo studies evaluating implants with similar surface topography have shown that commercially pure Ti grades II and IV and Ti-6Al-4V implants were equally biocompatible and osseoconductive.[31,94] However, biomechanical evaluation showed increased removal torque for Ti-6Al-4V compared with commercially pure grade II implants,[94] which may be expected due to higher coefficient of friction between bone and the textured Ti-6Al-4V implant surface resulting from more energy demanded to deform during reverse torque compared with commercially pure grade II. From a clinical standpoint, this finding may be of limited clinical relevance, and there is no current evidence that any implant system is superior from 1 to 10 years of service.[95] In the meantime, human retrieval studies are encouraged as they may provide valuable information regarding implant biomaterial performance. Recent human retrieval analysis showed that commercially pure Ti grade II is more prone to crack development than Ti-6Al-4V is.[76] In addition, the severity of the defects of implants under function eventually leading to failure are also more pronounced in commercially pure Ti implants.[77] Future clinical studies are warranted to detect whether the long-term survival (10 years or more) of implant-supported restorations is affected by Ti and its biomedical alloys that are thus far considered the gold standard material for endosseous dental implants.

6.1.2 | Ceramic implants

One motivation for the use of tooth-colored implant materials including ceramics is the reduction of peri-implant soft tissue discoloration observed with titanium implants that are clinically visible when soft tissue thickness is <2 mm. In such thin gingival biotypes, both titanium and zirconia implants lead to a clinically visible discoloration of the peri-implant mucosa, but it is significantly less visible with zirconia implants.[96] Allergic reaction to titanium, although rare (0.6%),[97] has also stimulated the development of dental implant biomaterial surrogates. The clinical relevance of other concerns, including titanium corrosion, are yet to be confirmed.

The most commonly used ceramic dental implant biomaterial is yttria-stabilized tetragonal zirconia polycrystals. This is because of its good biocompatibility and its optical and transformation toughening properties. However, metastability of yttria-stabilized tetragonal zirconia polycrystals triggered by stress/low-temperature degradation may be protective when of a toughening nature or detrimental to its strength and tissue integration if excessive.[84] During fabrication, the varying manufacturing processing among industries may result in final yttria-stabilized tetragonal zirconia polycrystals material microstructures that also vary in stability.[98] For this reason, the use of other dopants, such as ceria, to stabilize zirconia has been suggested;[99] in particular, the more promising use of alumina-toughened zirconia and zirconia-toughened alumina composites that present improved mechanical properties and limited low-temperature degradation compared with stabilized zirconia has been encouraged.

Studies have shown that zirconia implants presented survival rates inferior (85% at 1 year[100] to 88.9% at 3 years[101]) to titanium

implants, although still promising. Immediately loaded one-piece zirconia implants placed in fresh extraction sockets showed unacceptably high failure rates compared with delayed loading,[100] and they should be avoided until a learning curve is reached and more evidence is gathered. A systematic review limited by low-quality controlled trials and short-term follow-ups has indicated that, whereas promising, the use of zirconia and its composites as implant biomaterials warrants better designed long-term trials.[102] Thus, based on the low evidence for indication, considering its use in spans no longer than three-unit fixed dental prostheses, and the selective cases where zirconia implants are truly well indicated (esthetic zone, thin gingival biotype), it is clear that future developments are warranted along with long-term results. In addition, advances witnessed in titanium implants at the macro-, micro-, and nanogeometric scales are yet be explored in ceramic implants.[11]

6.1.3 | Polymer-based implants

In an attempt to use implant materials with properties (eg, modulus of elasticity) more similar to bone and with instances of tooth-colored materials, such as fiber-reinforced composites,[103] polyetheretherketone and others already used in orthopedics have been suggested as alternatives to titanium. Applications of polyetheretherketone include dental implants and abutments and removable and fixed dental prostheses.[104] Intriguingly, the theoretic benefit of polyetheretherketone in presenting a Young's modulus closer to bone compared with titanium, and especially with ceramic implants, has not been confirmed in long-term clinical controlled trials thus far, but corroborated only in a virtual in silico simulation[105] and contrasted in another.[106]

Since the introduction of polyetheretherketone in 1998, its bioinert and hydrophobic nature have been known. This has demanded a substantial amount of effort on surface engineering to develop a host to implant interface capable of establishing, maintaining, and improving long-term contact. The bioactivity and osseointegration of polyetheretherketone has shown to be inferior to those reported for titanium.[107] Because polyetheretherketone seems to be under active research and development, its indication as a prophetic alternative to titanium cannot be expected at this point.

7 | A BRIEF ON IMPLANT-SUPPORTED PROSTHETIC TREATMENT

Since its introduction, implant therapy has broadened its treatment scope, and a plethora of implant designs and materials, implant-abutment connection types, prostheses materials and several other variables have been introduced. From presurgical choices, including implant-abutment connection design, to rehabilitation choices (eg, cemented or screwed, prostheses material) the resulting treatment may be functionally and esthetically acceptable, but different longevities may be expected as a result of the combination of such choices. Evidence from biomechanical testing and some from systematic reviews will be briefly presented to corroborate that some combinations of the aforementioned variables may lead to reduced lifetimes.

7.1 | Implant-abutment connection designs

In general, implant-abutment connections can be external (eg, external hexagon) or internal nonconical (eg, internal hexagon), or conical, which can be further subdivided into taper-integrated screwed-in abutment (demands a screw to secure the abutment) and purely tapered interference fit (presence of interference between implant well and abutment enough to secure the prostheses in place, erroneously called morse taper).[108-110] The overall performance between external and internal implant-abutment connections has been the focus of systematic reviews which indicated that more technical complications are expected in external relative to internal connections.[111] Specific comparisons between connection types show that, as surmised, similar connections may perform different among manufacturers.[112,113] This may be explained by differences in milling parameters, materials, and quality-assurance and quality-control discrepancies.

7.2 | Implant diameter and fixation mode (screwed vs cemented)

Implant diameter selection may also impact the survival of narrow-, standard-, and wide-diameter implants given that survival of the restored system may proportionally increase with diameter and significantly improve when cemented in lieu of screwing.[114] Although both cementing and screwing have clinical advantages, final performance still seems to be controversial. Regarding single crowns, systematic reviews showed that technical complications, such as screw loosening, are significantly more often encountered in screw-retained than cemented prostheses, whereas biologic complications are more common in the latter.[115] In fixed dental prostheses, significantly more technical complications have been reported for screw-retained than cement-retained prostheses,[116] and for full-arch the complications were comparable between both retention techniques.[117] A comprehensive review on the topic with clinical recommendations can be found elsewhere.[118]

7.3 | Prostheses material selection

Despite the vast array of materials available for restorative purposes,[119] the current indication seems to be more evidence based for tooth-supported crowns[120,121] and fixed dental prostheses[122,123] than for implants. It is still debated whether the performance is different between tooth- and implant-supported reconstructions, but the differences in support (ankylosis in bone versus mechanoperception in periodontal ligament) have demonstrated that implants do result in a reduced tactile sensibility compared with teeth,[124] which may eventually lead to a reduction in masticatory muscle coordination and increase in occlusal overload

susceptibility.[125] It is interesting that whereas porcelain-fused-to-zirconia crowns currently present similar survival to metal ceramics when tooth supported,[121] porcelain cohesive fractures have been shown to vary from 7.5% in 6 months,[126] to 24.5% in 2 years,[127] to 4% in 3 years,[128] to an outrageous 42.8% in 5 years[129] when implant supported, suggesting that differences in treatment results may arise not only due to laboratory processing parameters but also due to variations in the implant-abutment-prostheses treatment conception. Since substantial efforts will likely improve the laboratory processing and clinical parameters of porcelain fused to zirconia, upcoming results will eventually show increased and comparable survival rates to metal ceramics. Regardless of restorative material, it must be kept in mind that implant-supported reconstructions demand more maintenance due to technical and biologic complications.[130] Therefore, strategies to save teeth remain highly recommended because, when compared with dental implants, natural teeth even when previously compromised by periodontal disease and/or endodontic problems, but successfully treated, still surpass their longevity when evaluated after 10 years of service.[131-134]

8 | FINAL REMARKS

The majority of studies reporting on high survival rates for implant-supported reconstructions commonly include follow-ups of <10 years. However, the increasing life-span of patients along with an increasing number of implants being placed will certainly result in a significant challenge for the next generation of dental professionals to address peri-implantitis and along with a variety of mechanical complications. Through extensive multidisciplinary research it is expected that the endeavors to improve implant hardware, surface treatment, implant bulk biomaterials, and prostheses rehabilitative strategies may result in increased dental implant treatment longevity.

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