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Smarter Surfaces, Stronger Bonds

Dental Implant Surface Modifications for Enhanced Osseointegration

DeepScience

Smarter Surfaces, Stronger Bonds : Dental Implant Surface Modifications for Enhanced Osseointegration

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Preface

Dental implantology has undergone a remarkable transformation from the rudimentary attempts of ancient civilizations to the precision-engineered systems of contemporary practice. Central to this evolution is our growing understanding of how implant surface design influences biological response and long-term success. Amongst these design features, implant surface topography has emerged as a pivotal determinant of stability, clearly surpassing the earlier concept of simple friction-fit implants.

This book offers an in-depth overview of the key scientific developments and technological advances that have shaped current implant surface modification strategies. It reviews the evolution of both additive and subtractive surface treatments—such as titanium plasma spraying, grit blasting, acid etching, and anodization—and explores newer innovations including biomimetic and antimicrobial coatings. Special attention is given to how these physical and chemical surface changes influence osteocyte activity, guide bone remodelling, and ultimately strengthen osseointegration.

By combining biological concepts with clinical research, this volume provides a clear and comprehensive understanding of how surface engineering improves the reliability and longevity of dental implants. It is intended to serve as a valuable reference for clinicians, researchers, and students who wish to gain deeper insight into the scientific foundations of modern implant surface technology.

Dr Kavita Verma

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1 Introduction

Dental implants have been used to replace the form, function and esthetics in persons with missing teeth. Uneventful and successful treatment outcome in the post – prosthetic placement phase is based on even occlusal force distribution, the osseointegration of the dental implant.

Dental implants may be forged from different materials, ranging from metals and alloys, to ceramics and composites. However, titanium and titanium alloys are most widely used due to its outstanding property of osseointegration with the alveolar bone apart from the property of forming a protective surface oxide layer, namely titanium dioxide.[1] These interactive properties of titanium with human bone can further be influenced for better success rate and enhanced survival and performance by the addition of other substances on the titanium surface, such as hydroxyapatite, that is known to hasten the process of osseointegration. Alternatively, drug molecules may also be added to the implant surface, in order to exert an anti-bacterial effect in the osteotomy. This decreases the occurrence of infection from any existing pathogens during surgical placement and thereby improves healing. In the modern times, biomimetic coatings that mimic the natural human molecular environment have been also used in conjunction with dental implants. These are said to possess osteogenic capability which allows for faster bone regeneration at the osteotomy site, along with better osseointegration.

The implant design significantly influences overall performance of the dental implant and prosthesis once it is loaded.[2] This includes structural design elements such as the length and diameter of the implant; the design of the crest module and implant apex; and also, the thread design in the threaded implant systems. The threaded implants have an upper-hand than smooth implants due to increased surface area which results in enhanced initial stability, better osseointegration, and desirable distribution of occlusal stresses.[3] These advantages can be attributed to the greater bone to implant contact (BIC) area gained by means of the threads that enhance the implant surface area. In addition, longer implants are known to provide better implant stability quotient. Implants with larger diameters are also preferred as they provide a larger surface area for osseointegration. This is especially useful in surgical osteotomies placed in D3/ D4 quality of alveolar bone where the bone is less dense. Also, a larger implant provides better stress distribution. The overall shape of the implants also influences the implant success. Screw shaped implants perform better compared to parallel shaped implants, again due to better primary stability and implant stability quotient associated with the former.

Implants with several types of surface modifications in the form of coatings, or texturization are presently available in the market. These different surface treatments are performed on a dental implant with the major aim of improving implant performance, that is based upon primarily – the primary stability, osseointegration, implant to bone contact area. These are in turn, a product of the implant dimension; the structure, placement and the number of implant threads; implant surface interaction with the surrounding bone at the nano level, based on chemical reactions which are influenced mainly by the presence of materials with different properties, that are added on top of the titanium surface in order to enhance its interactions so as to obtain a favorable implant performance. Additionally, the implant may also be subject to other treatments that work to increase the available surface area and roughness. These significantly influence the biologic interaction between titanium and osteoblasts, that favor osseointegration, as osteoblasts prefer a rough surface to attach and proliferate.

Thus, modifications on the titanium implant surface are done with the sole purpose of enhancing osseointegration. These may be classified as macro-geometric modifications namely the different implant macrostructure including the crest module and thread patterns; and the micro-geometric modifications that includes additive techniques like anodization, titanium plasma spraying; and subtractive modifications such as sand blasting acid etching, and laser preening.

This book gives a detailed account of the different surface modifications of dental implants, including the latest advances.

2.1 Dental Implants – A Brief History

Since ancient times, spanning civilizations and cultures, mankind has tried to replace missing teeth using different materials and techniques. Many of these procedures are noteworthy, having formed the basis for further advancements and modern implantology.

Dental implants were first used by the Mayans in around 600 AD, when pieces from sea shells were used in the form of mandibular implants, with proof of circumscribed compact bone formation upon radiographic evaluation done in 1970. At around 800 AD, the first stone implant was used for the mandible in the Honduras.

From around 1500’s to 1800’s allotransplantation of teeth from cadavers and underprivileged humans was widely experimented with. About this time, Dr. John Hunter successfully implanted a human tooth into a rooster’s comb. In 1809, J. Maggiolo used a tube fabricated from gold, as an implant in a freshly extracted socket. Following healing, a crown was placed, however, extensive gingival inflammation followed the procedure. Several other materials were used as implants during this period, including silver capsules, corrugated porcelain, and iridium tubes.[2,3]

Table 1: Historical Periods in Development of Dental Implants[1]

PERIODS	TIME
AD 1000	ANCIENT ERA
1000 - 1800	MEDIEVAL ERA
1801 - 1910	FOUNDATIONAL ERA
1911 - 1935	PRE MODERN ERA
1936 - 1978	PRE – BANEMARK PERIOD
1978 - 1998	BRANEMARK PERIOD
1998 – present	POST - BRANEMARK ERA

In the 1930's, Strock brothers, used Vittalium screws to provide anchorage and support, to replace the missing tooth. This was notably the first incidence of use of a biocompatible material in the oral cavity. In 1938, Adams patented a cylindrical endosseous implant having a smooth gingival collar. Following this, a post-type endosseous implant was developed by Formiggini and Zepponi in the 1940's. At about the same time, subperiosteal implant design was introduced by Dahl in Sweden. In this design, screws were placed on the alveolar ridge crest, followed by flattened abutments. This design was modified by Gershkoff and Goldberg to extend to the external oblique ridge. They used cobalt-chromium-molybdenum to fabricate the implant.[4] This was followed by the innovations of Dr. Lee who introduced an endosseous implant with a central post.[5]

Many advances in implant design were made in the 1960's. Cherchieve created a screw shaped, single piece, cobalt – chromium, double-helical spiral implant.[6] This design was modified by Giordano Muratori by adding an internal threading to the shaft of the implant. In 1963, Leonard Linkow turned the spiral design into a blade implant. He further developed the Ventplant implant.[7,8] In the mid-60's, Sandhaus designed a crystalline aluminum screw.[9] The Ramus Blade endosseous implant was created using stainless steel (surgical grade) by Roberts & Roberts to serve as a "synthetic third molar". The ramus frame implant was also created by them. This design provided stability by deriving anchorage from the ramus and the symphysis.

In the 1970's, Greenoble placed vitreous carbon implants.[10] Weiss and Judy introduced intramucosal inserts for the retention of removable maxillary prostheses.[11] In 1975, Small created the mandibular staple implant (first trans-osteal implant), thast was attached to the mandible through an incision in the submental region. It was used mainly for individuals with an atrophic mandible.[12] In 1978, Brånemark presented a two-stage threaded titanium root-form implant which he termed fixtures.[13] These implants were documented and remained in place for the next 40 years.[14] With his implant came the concept of 'osseointegration', defined as 'a direct structural and functional connection between ordered, living bone, and the surface of a load carrying implant'. This was discovered while studying rabbit femurs for blood flow. This involved placing chambers fabricated from titanium, into the bone. These got affixed, and could not be removed.[14] Subsequently, tapered forms of this implant appeared.

Calcitite, a synthetic polycrystalline ceramic hydroxylapatite was created in the early 1980s by the The Calcitek Corporation. In 1985, the Integral Implant System was created.[15]

Tatum introduced the Omni-R implant, in early 1980s. It had fins made of titanium alloy, placed horizontally. Around the mid 1980's, Niznick introduced the Core-Vent implant. It was shaped as a hollow basket. It also had a threaded piece for the purpose of engaging bone. Niznick also designed the Screw-Vent implant with a coating of hydroxyapatite, to allow ease of bone formation on the implant surface. Niznick formed the Core-Vent company which designed the Swede-Vent Implant System. This came with an external hexagonal interface used for engaging with the abutment.[16]

2.2 Dental Implant Materials

As the dental implants are required to function in a complex environment, with transgingival attachments and prostheses, it is imperative that the materials used for implant fabrication possess certain desirable properties. In order to standardize the implant materials, some guidelines have been provided by various organizations. These include, the 44 ASTM Committee F4 (ASTM F4) and ISO (ISOTC 106, ISOTR 10541).[115]

Biomaterials are synthetic substances used to replace and augment the biological tissues. The biomaterials used for dental implants are classified into the following types, based on their biodynamic activity- bioinert, biotolerant and bioactive, as shown in Table 2.

Table 2: Classification of dental implant materials based on biodynamic activity

	BIOTOLERANT	BIOINERT	BIOACTIVE
Metals	Gold Co-Cr Alloys Stainless Steel Niobium Tantalum	Cpti Titanium Alloy	
Ceramics		Aluminium Oxide Zirconium Oxide	Hydroxyapatite Bio Glass Carbon Silicon
Polymers	Polyethylene Polyamide Polymethyl Methacrylate Polytetrafluroethylene Polyurethane		

Several different materials have been tried over the years for manufacturing dental implants, including metal alloys, ceramics, carbon-silicon compounds, polymers and composites. For implant fabrication, metals and their alloys are the most commonly used materials. These include titanium, vanadium, tantalum, chromium, molybdenum and nickel.

Metals and Metal Alloys:

Ease of processing, ease of sterilization and optimal biomechanical properties of metals, make them suitable implant material. Importantly, with the advent of newer materials, gold, stainless steel, cobalt-chromium alloy, have now become obsolete. Titanium and its alloys are the material of choice. Platinum, gold, and palladium are metals with relatively low mechanical strength, which restricts their application in implant design. These materials also present challenges due to their high cost per unit weight and density, particularly when used in upper arch devices. Despite these limitations, gold remains a preferred choice for surgical implants because of its biocompatibility and availability. An example of this is the Bosker endosteal staple, which utilizes a gold alloy system in its design.[115]

a. Iron-Chromium-Nickel Based Alloys: These are often used for orthopedic and implant devices. On implantation, these form an external oxide layer of passivation.

However, these are easily prone to pitting type of corrosion, resulting in galvanic coupling and biocorrosion, especially when titanium, cobalt, zirconium or carbon implant biomaterials are used alongside. Also, as they contain nickel as a major element, they cannot be used in patients exhibiting nickel allergy.[18] Iron-based alloys, such as stainless steel, possess galvanic potentials and corrosion characteristics that can lead to galvanic coupling and biocorrosion when in contact with materials like titanium, cobalt, zirconium, or carbon used in implants. This occurs because these alloys have high galvanic potentials and corrosion resistance, making them prone to pitting corrosion when coupled with other metals. However, if these alloys are used independently—meaning they are not in contact with or electrically connected to other materials—then galvanic coupling does not occur. In such cases, each device can function independently without the risk of accelerated corrosion or ion release associated with galvanic interactions.

b. Cobalt Chromium Alloy: The major components of this alloy are cobalt, chromium and molybdenum. This alloy, exhibits oxides of chromium (primarily Cr_2O_3 with some suboxides) under normal implant surface–finishing conditions after acid or electrochemical passivation. These chromium oxides have very little thickness i.e., in nanometer dimensions, with an amorphous atomic structure.[115] Chromium offers corrosion resistance by forming the oxide layer, while molybdenum provides strength and bulk corrosion resistance.[19] These are utilized for the manufacture of customized implants including subperiosteal frames.

c. Titanium: This is the most commonly used alloy to fabricate dental implants due to its good mechanical properties, low density (4.5 g/cm^3) and good bone-contact biocompatibility. The main alloy used is commercially pure titanium, CP Ti.[20.] This metal is available in four grades numbered 1 to 4, according to the purity and the processing oxygen content.[21] These grades differ in corrosion resistance, ductility and strength, and it is grade-4 CP Ti, with the highest oxygen content (around 0.4%) and best overall mechanical strength that is most widely used for dental implants.[21]

The titanium alloy Ti-6Al-4V, commonly known as grade 5 titanium, incorporates aluminium and vanadium in its surface oxide layers. These elements can leach into the surrounding biological environment, potentially causing adverse effects. Aluminium interferes with the mineralization of bone, often causing structural deficiencies. Vanadium has toxic effect on the cells and incite allergic responses.[22,23] However, the concentrations released are typically below those encountered through normal dietary intake of these ions.

The corrosion behaviour of metal implants plays a crucial role in determining their biocompatibility. When metals corrode, they release metal ions that can adversely affect the surrounding tissues and potentially lead to systemic issues. Titanium alloys, such as

Ti-6Al-4V, are known for their excellent corrosion resistance due to the formation of a protective oxide layer. However, the presence of proteins like albumin can influence this behaviour. Albumin can adsorb onto the titanium surface, and under certain conditions, this adsorption can alter the corrosion dynamics, potentially leading to increased metal ion release into the surrounding tissues. This interaction underscores the importance of considering protein-implant interactions when evaluating the long- term performance and safety of titanium-based implants.[24, 25] Both commercially pure titanium and Ti-6Al-4V are bioactive and capable of promoting bone formation in direct contact with the metal surface. The interfacial zone between the titanium alloy implant and living bone is critical in the development of osseointegration. Generally, CP Ti is slightly favoured; however, in vitro studies have often found Ti-6Al-4V to be superior.[26]

1. Binary Alloys of Titanium: Titanium alloys are commonly used in dental implants due to their excellent mechanical properties and biocompatibility. Various metals, including niobium, silver, gold, manganese, and zirconium, can be alloyed with titanium in different proportions to enhance these properties. Zirconium is a corrosion- resistant metal that readily forms alloys with titanium.[27] Despite their advantages, compared to commercially pure titanium, the Ti-Zr alloys show inferior osseointegration.[28] Niobium is a ductile, non-toxic metal that stabilizes the β-phase titanium alloys, resulting in improved mechanical properties and a lower elastic modulus that more closely matches that of human bone. Alloys with less than 10% niobium exhibit hardness, yield strength, and tensile strength superior to those of commercially pure titanium. However, studies indicate that human fibroblasts proliferate more slowly on Ti-Nb alloys compared to CP Ti, suggesting potential differences in cellular response.[29, 30]

Different metals such as niobium, silver, gold, manganese and zirconium may be added to titanium in different proportions in order to prepare alloys for use as dental implants. Niobium and zirconium are inert with respect to their biological effects. Zirconium readily forms alloys with titanium, and it strongly resists corrosion which means that it releases only trace amounts of metal ions into the body.[27] Moreover, studies aimed specifically at dental applications have shown the alloy to have mechanical properties comparable with CP Ti and with suitable surface preparation, good biocompatibility and improved osteoblast adhesion compared with cpTi.[28] Binary alloys containing minor amounts of niobium (less than 10% by mass) have been found to have good mechanical properties. Their hardness, yield strengths and tensile strengths typically exceed those of CP Ti. However, experimental studies have shown that human fibroblasts grow slower and less extensively in Ti-Nb alloys than on CP Ti.[29, 30]

2. Multi-Component Alloys of Titanium: Ti-6Al-7Nb is a titanium alloy containing niobium that has been extensively researched for its applications in bone-contacting implants. This alloy demonstrates enhanced mechanical properties, such as improved corrosion resistance, compared to commercially pure titanium.[31] Biologically, Ti- 6Al-

7Nb exhibits similar responses to CP Ti. Studies have shown that human gingival fibroblasts adhere, spread, and proliferate on both alloys to comparable extents. Additionally, there is evidence suggesting that Ti-6Al-7Nb allows better adhesion of osteoblast-like cells as compared to CP Ti.[32]

3. Titanium-zirconium alloy (Straumann Roxolid): As compared to pure titanium, titanium zirconium alloys with 13%-17% zirconium have better mechanical attributes, such as increased elongation and fatigue strength, combined with the advantage of excellent osseointegration qualities. It is chemically inert, with minimal local or systemic adverse reactions, possessing enhanced cell adhesion, favorable tissue responses, and excellent biocompatibility with the surrounding hard and soft tissues.[115] Roxolid proves to be 50% stronger than pure titanium. Narrow implants and implant components may be subjected to high strains due to its superior mechanical properties.[33]

Ceramics:

Ceramics are inorganic materials consisting of metallic and non-metallic components chemically bonded together by means of ionic bonds. Properties of ceramics, such as compressive strength, tensile strength exceed that of compact bone by three to five times. They also have a high modulus of elasticity. However, steam sterilization leads to a decrease in strength for some ceramics resulting in the formation of scratches or notches. These may act as fracture initiation sites.[115] Depending on the chemical properties, they can be classified as (a) inert, (b) low or medium surface activity, or (c) bioresorbable (adsorbable) ceramics. Various ceramics such as aluminum oxide, zirconia, hydroxyapatite, calcium phosphate and bioglass are being used as dental implant materials.

a) Aluminium Oxide: This is bioinert biomaterial, in nature, and shows excellent corrosion resistance, high wear resistance and high strength. However, it has been discontinued due to poor survival rate.

b) Calcium Phosphate Ceramic: Calcium Phosphate Cement biomaterials offer several advantages. They are composed of elements like calcium, phosphorus, oxygen, and hydrogen, which are naturally found in human bone. This composition ensures excellent biocompatibility across various tissues and minimizes the risk of immune rejection. Also, their mechanical properties particularly elasticity, closely resemble that of natural bone, making them suitable for load-bearing implants. However, they also have some limitations. The chemical composition and structural characteristics of CPCs can vary, leading to inconsistencies in performance. Under conditions of fatigue loading, they may exhibit relatively low tensile and shear strengths, which can affect their durability in certain applications.[115] Two commonly used calcium phosphate materials in implant fabrication are Hydroxyapatite (HA) and Tricalcium Phosphate. Tricalcium Phosphate, particularly β-TCP, is known for its higher resorption rate compared to HA, making it beneficial for bone regeneration. Both have the ability to bond directly with bone tissue, enhancing the integration of implants. They are often used as coatings on implants to promote bone healing and improve the overall success of the implant.

c) Bioglass (SiO_2-CaO-Na_2O-P_2O_5-MgO): Bioactive glass is recognized as a bioactive material due to its ability to promote bone formation, exhibiting osteoinductive properties at the interface between the implant and bone.[28] The porosity and surface roughness of bioactive glass, which can be adjusted through deposition techniques, are crucial for its bioactivity.[126] However, its inherent brittleness limits its application in stress-bearing areas, leading to its reduced use in implant fabrication.[34]

Zirconia:

Because of possible esthetic concerns with titanium implants, newer materials like Zirconia were introduced as an alternative. They are considered to be inert in the body and exhibit minimal ion release compared with metallic implants. Yttrium-stabilized tetragonal zirconia polycrystals appear to offer advantages over aluminium oxide for dental implants because of their higher fracture resilience and higher flexural strength. The material also provides high strength, fracture toughness, and biocompatibility. The inflammatory response and bone resorption induced by ceramic particles are less than those induced by titanium particles, suggesting the biocompatibility of ceramics. Although zirconia may be used as an implant material by itself, zirconia particles are also used as a coating material of titanium dental implants. The compressive strength and stress distribution patterns were also observed to be similar to titanium implants. 35

(a) Aluminium, Titanium and Zirconium Oxides: Implants composed of aluminium, titanium, and zirconium oxides are favoured for pin- type, root form, and endosteal plate-type designs due to their superior mechanical properties. These materials exhibit strength levels 3 to 5 times greater than that of compact bone. They also possess high elasticity, fatigue, and fracture resistance, along with increased compressive and tensile strength.36 Polycrystalline alumina is a ceramic biomaterial known for its low thermal and electrical conductivity, minimal biodegradation rates, and reduced reactivity with bone, soft tissue, and the oral environment. These properties make it more desirable compared to other synthetic biomaterials. However, in the 1970s, a series of root form and plate form devices made from these materials experienced intraoral fractures after several years of use. These fractures were attributed to fatigue cycling, where biomechanical stresses led to localized bending and tensile loading. Although initial testing indicated adequate mechanical strengths for these polycrystalline alumina materials, long-term clinical results revealed limitations related to both functional design and material properties.[115]

Polymers and Composites:

Polymers were first utilized as implant biomaterials in the 1930s. These materials typically exhibit lower mechanical strength and elastic modulus, and high fracture elongation compared to other forms of biomaterials. They also serve as thermal and electrical insulators and, when formulated as high-molecular-weight systems without plasticizers, demonstrate relative resistance to biodegradation. In comparison to bone, most polymers have a lower elastic modulus. Early polymers used for medical implants included polymethyl methacrylate (PMMA) and polytetrafluoroethylene (PTFE), followed by polyamide, polyethylene, polyurethane, polypropylene, polydimethylsiloxane, polysulfone, and silicone. A notable recent addition is polyether ether ketone (PEEK). PEEK's primary advantage over materials like titanium and zirconium lies in its elastic modulus of approximately 3.6 GPa, which is closer to that of bone. Furthermore, PEEK can be reinforced with carbon fiber to achieve an elastic modulus of 17.4 GPa, approximating that of cortical bone. This enhancement not only improves mechanical compatibility but also offers better aesthetic properties and is suitable for patients with titanium allergies.[37, 38] However, polymers and their composites are particularly sensitive to sterilization and handling procedures. However, steam or ethylene oxide is not suitable for sterilizing the implants.[115]

Carbon and Carbon-Silicon Compounds:

Carbon and carbon-silicon compounds were first used for fabricating dental implants in the 1970s. These materials are known for their excellent biocompatibility and mechanical properties that closely resemble those of bone tissue. Despite these advantages, the use of carbon and carbon-silicon compounds in major load-bearing areas is limited due to their brittleness and low tensile strength, particularly along the interface between the substrate and coating. For instance, the Vitredent two-stage root replacement system, introduced in the early 1970s, gained popularity but was eventually withdrawn from clinical use. This decision was attributed to a combination of design flaws, material limitations, and application challenges. Thus, while carbon and carbon-silicon compounds offer promising properties for dental implants, their application in areas subjected to significant stress is constrained by their mechanical limitations. The experience with the Vitredent system underscores the importance of addressing these challenges to ensure the long-term success of dental implants. [115]

2.3 Implant Surface Macrotopography

Osseointegration is significantly influenced by various factors, with initial stability being a key determinant. One of the primary contributors to initial stability is the implant design, which encompasses both macro and micro features. Macro-design elements, such as thread geometry, implant shape, and surface roughness, are important for the long-term success of implant treatments. Continuous advancements in implant design aim to improve these features, thereby increasing the overall success rate of implants. Implant designs have evolved over time from press fit blade implants to cylindrical or straight smooth surfaced implants to tapered implants with threaded surface, claiming to provide several advantages over designs adopting a smooth surface. By property, bone is best capable of handling compressive stress as compared to shear forces. Therefore, the implant – bone contact area available for occlusal force distribution in compression is the effective area for transfer of compressive force. This force transfer is not uniform across the cortical and cancellous bone components of the alveolar bone. The implant design in the cortical bone region dictates suitable load distribution without overloading the bone. Again, within the trabecular bone, the geometry of the implant in contact, determines the pattern of stress transfer.

Elements of Implant Design:

This includes the overall shape, diameter, length, and threads on the implant surface.

Shape:

The two most common forms of dental implants are the cylindrical form, tapered form (root form) and mixed forms. Shape of the implant, whether tapered, or cylindrical is one of the major determinants of the surface area, along with any geometric features that extend outward from the implant axis. Conical implants possess better primary stability compared to the standard Branemark design.[79] Tapered dental implants generate lateral compressive forces on the cortical bone, which significantly contribute to their enhanced primary stability. This design allows for better load distribution compared to cylindrical implants.[80] Among hybrid implant designs, apically conical implants offer increased primary stability due to more substantial crestal compression. Conversely, crestal parallel or back taper designs are more effective in relieving stress on the bone. These findings underscore the importance of implant geometry in achieving optimal primary stability, particularly in challenging bone conditions. The tapered design's ability to exert lateral compressive forces enhances initial stability, making it a preferred choice in various clinical scenarios.[81]

Diameter:

Diameter of the implant is another important factor in determining the surface area and hence success of the implants. Larger the diameter, more the surface area for load transfer. Wider diameter implants reduce the stress at the bone – implant interface, while smaller diameter implants increase the stress at this interface. Moreover, wider implants are more resistant to fracture from occlusal overload and fatigue. They also possess more thickness of the implant wall after the abutment screw bore. Wider diameter collars fixtures have better prognosis compared to narrower implants. Thick dental implants are engineered to offer enhanced mechanical stability, enabling them to endure higher occlusal loads. These implants feature a larger diameter, making them suitable for supporting extensive prosthetic restorations or areas subjected to significant biting forces. They are particularly advantageous in the posterior regions of the maxilla and mandible, where bone density is typically greater, and the bone height is sufficient to accommodate their size.[82]

Length:

Length is another important parameter of concern while designing the dental implant for effective stress transfer. It also affects the feature of implant stability apart from the transfer of forces mainly within the cancellous part of the bone, along its length. in other words, a shorter length offers a much smaller surface area compared to one with an increased length, even with the same diameter involved. Longer fixtures have larger surface area to facilitate better bone contact. Micromovements, stress, and strain were found to be lower for long implants as compared to short ones.[83] There is also statistically significant decrease in von Mises stress as the implant diameter increased.[84] The ITI Consensus Group advises the use of short implants in situations where bone grafting is not recommended, where such procedures could lead to increased morbidity, or to shorten the overall treatment duration. Additionally, short implants are beneficial when there is a need to minimize the risk of damaging adjacent anatomical structures, such as the maxillary sinus, nerves, or neighbouring implants. However, when there is sufficient bone height and no increased surgical risk, implants longer than 6 mm are preferred.[85]

Implant Collar:

The implant collar represents a transitional zone between body and the prosthesis bearing part of the implant. Its design determines the pattern of stress transfer to the cortical bone and also the placement of the prosthetic interface, relative to the bone and surrounding gingival tissue. It is also designed to reduce bacterial invasion. The height, diameter, taper and surface texture of the implant collar are closely related to the crestal bone loss, incidence of peri-implantitis, and fracture resistance of the implants after prosthesis placement. Collar has increased height in implants that receive supragingival prosthetic connections, while in the bone level implants, the height is lesser. The diameter is preferably larger than the implant body for better distribution of stresses as compared to tapering of the collar. A polished collar is preferred to one with texture, as any texturization might lead to plaque accumulation and lead to peri-implantitis. The collar may be parallel, flared or tapered in shape. Collars were also machined to achieve better soft tissue adaptation.[86]

However, some designs incorporate fine micro threads in the collar to aid in the maintenance of the crestal bone. Roughened collar fixtures with micro threads added to it, show a success rate of 96.4%.[87] A nonthreaded microtextured design showed lesser marginal bone loss compared to microgrooves.[88] Also, a machined collar was better at maintaining marginal bone height, compared to micro threaded or microtextured implant collar. [89] Also, the probing depth and bleeding index in the machined collar group was also less. [90]

Threads:

Most of the present implants have threads on the external surface. They allow for increased surface area, increased bone to implant contact, positively contributing to effective osseointegration and also to efficient transmission of stress from the implant structure to the bone, when compared with smooth surface implants. They are also an important feature for gaining a good primary stability, which becomes even more important in the regions with poor bone density. The threads are governed by important characteristics including the pitch, depth and shape. Study of threaded implant bodies shows a greater percentage of bone-implant contact compared with non-threaded cylinder implants.[91]

a) Thread Pitch: Pitch of the thread refers to the axial distance between adjacent threads on a screw,

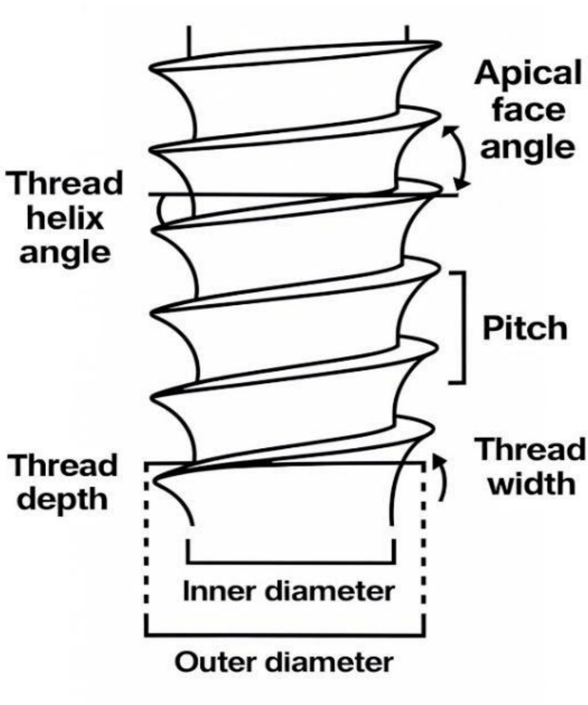

Fig 1 : Implant Thread Design [92]

measured parallel to its axis. A smaller thread pitch increases the number of threads per unit length, enhancing the implant's surface area and potentially improving stress distribution in surrounding bone. However, the optimal thread pitch for stress distribution may vary depending on factors such as thread form and depth.

b) Thread Shape: The geometry of dental implant threads significantly affects how stresses are transferred from the implant to the surrounding bone. Thread designs such as V-shaped, square, buttress, and reverse buttress each influence stress distribution differently. V-shaped and Square Threads are associated with lower stress on cancellous bone compared to thinner square threads. Square threads, in particular, are beneficial for immediate loading due to their ability to distribute compressive forces effectively. Buttress threads are optimized for axial loads, efficiently distributing compressive forces along the implant. Reverse Buttress Threads convert axial forces into a combination of shear and compressive forces, which can be advantageous in certain clinical situations. The face angle of a thread—defined as the angle between the thread face and the implant's longitudinal axis is significant in determining the direction of forces transmitted to bone. V-shaped Threads typically have a face angle of around 30°, leading to higher shear forces at the implant-bone interface. Reverse Buttress Threads have a smaller face angle of approximately 15°, these threads result in lower shear forces compared to V- shaped threads.

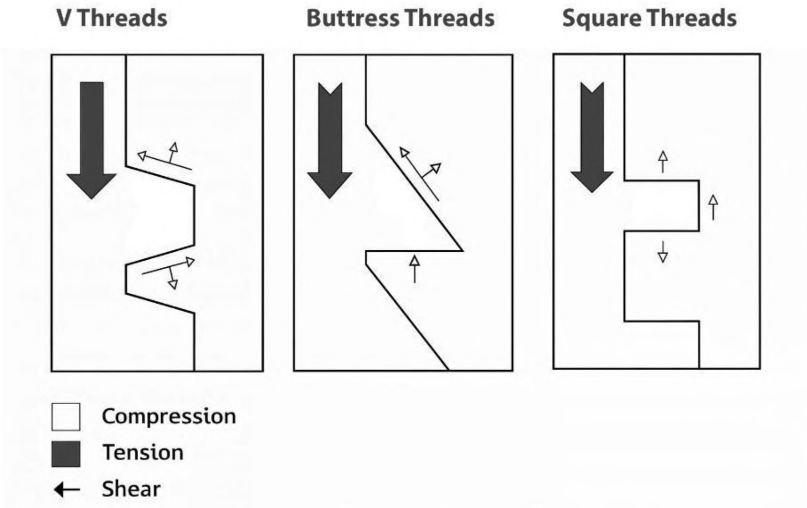

Fig 2 : Implant Thread Shapes[92]

c) Thread Depth: This may be described as the vertical distance from the crest to the root of an implant's thread. This significantly influences the load-bearing capacity of the implant's inferior flank. A greater thread depth increases the surface area available for transferring compressive forces to the surrounding bone, thereby enhancing the implant's stability.

Fig.3: Implant thread depth [92]

However, for implants with smaller diameters, increasing thread depth can reduce the cross-sectional thickness of the implant wall. This reduction may negatively affect the implant's structural strength. Therefore, balancing thread depth with wall thickness is essential, especially in smaller implants. In regions with low-density bone, deeper threads can improve primary stability by increasing the initial bone-implant contact. This enhancement leads to higher insertion torque and better mechanical stability. Conversely, in denser bone areas, the increased insertion torque required for deeper threads might necessitate the use of a bone tap to ensure proper seating of the implant. Implants featuring progressive threads—where thread depth gradually decreases from the apical to the coronal end—have been shown to offer superior bone-implant contact both histologically and radiologically compared to cylindrical designs. This design contributes to higher primary stability, making it particularly beneficial in areas with varying bone densities.[96]

d) Apical Region Geometry: The apical part of the implant facilitates its insertion into the bone. The tip is tapered to allow axial length of the implant to enter the osteotomy before the threads come into contact with the prepared walls. This allows the alignment of the implant axis with the axis of the osteotomy. The taper of the implant matches with the apical portion of the implant drill used to prepare the osteotomy. Some small-diameter implants are designed to be placed deeper than the prepared osteotomy in order to improve the primary stability. The apical end of conventional implants is kept flat or round in shape to minimize the probability of perforating the maxillary sinus membranes during placement. The apical end on small-diameter implants typically tapers to a sharp point. This area of the implants also expresses flat slots or grooves on the external surface, circumferentially arranged, to stabilize the implant against rotational forces, such as those generated during abutment adjustment and prosthetic procedures. Bone grows into these slots during healing and resists rotation.

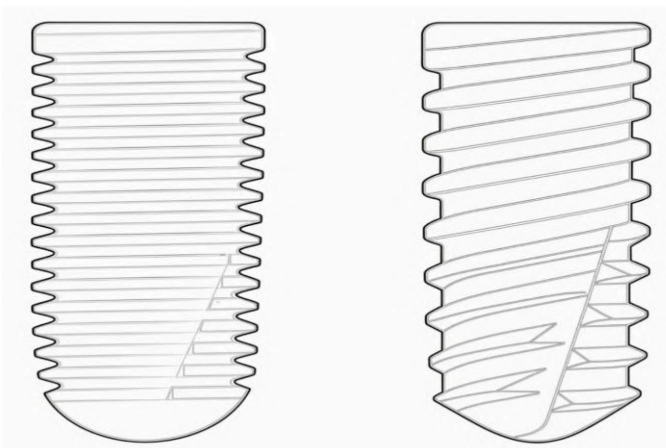

Fig 4 : Implant Apical Geometry [92]

2.4 Implant Surface Materials

The main purpose of implant surface coatings is to combine optimal bulk properties of the implant (such as tensile strength or stiffness) with desired surface characteristics (such as biocompatibility, wettability, roughness, etc.). Several functional coatings are used to improve the biocompatibility, bioactivity, resistance to corrosion, and antimicrobial properties of dental bio materials. [40]

Hydroxyapatite:

One widely adopted technique involves applying a hydroxyapatite (HA) coating to implants[41]. HA is one of the stable forms of calcium phosphate that integrates with organic matrix, enhancing its strength.[42] The bioactivity and osteoconductivity of the titanium substrate can be enhanced by the HA layer. By using a micro-arc oxidation process to create a porous hydroxyapatite-coated titanium alloy surface, it is possible to enhance the mechanical properties and promote bone formation by increasing the interface contact rate and bone to-implant contact.[43]

Growth Factors as Coatings on Implants:

Bone morphogenetic proteins (BMPs) are pivotal in cartilage and bone development, influencing osteogenic cells and promoting bone mesenchymal stem cell (MSC) differentiation.[44,45] Studies have demonstrated that titanium implants incorporating BMP-2 exhibit superior bone-to-implant contact, enhanced new bone formation, and increased bone density compared to acid-etched implants.[46,47] Additionally, local administration of BMP-7 has been shown to improve osseointegration through the formation of poly-ethyl acrylate.[48,49] Electrospraying polylactic-co-glycolic acid (PLGA) and basic fibroblast growth factor (bFGF) onto titanium implants can promote bone formation adjacent to the implant surface.[50] Furthermore, coating implants with vascular endothelial growth factor (VEGF) can stimulate the growth of osteoblasts and endothelial cells.[51]

Extra Cellular Matrix Coating:

Another method to increase dental implant biocompatibility is by accumulating extra cellular matrix (ECM) proteins on implant surfaces, which control cell-matrix adhesion.[52] These extracellular matrix proteins help in the early stages of bone healing by rearranging intracellular microfilaments and microtubules. The tetrapeptide Gly-Arg-Gly-Asp (GRGD) plays a crucial role in enhancing osteoprogenitor cell migration to implant surfaces by facilitating cell adhesion through integrin binding. This interaction promotes osteoblast differentiation and osseointegration.[53] Surface modification of titanium implants with GRGD peptides influences their topography, hemocompatibility, and wettability.[54]

Coatings With Antibacterial Property:

To address implant-associated infections, researchers have developed certain implants with coatings that specifically target and eliminate bacteria. They may also prevent bacterial adherence and biofilm formation.[55, 56]

a) Drug Coating on Dental Implant: Antibiotics like Simvastatin and Bisphosphonates are used to coat the dental implant surfaces.[57] These coatings aim to prevent infections and enhance bone healing.[58, 59]

Dental implants can be classified into two types based on the nature of the antimicrobial coating:

1. **Type I Surfaces**: These surfaces actively release antimicrobial agents to inhibit bacterial adherence and promote bacterial killing.

2. **Type II Surfaces**: In contrast, these surfaces have antimicrobial agents permanently bound to them, providing sustained protection against bacterial adhesion and biofilm formation. For example, applying tetracycline permanently on implant surfaces efficiently eradicate the bacteria.[60, 61]

To address the limitations of Type I surfaces, different chemicals have been permanently added to implant surfaces to generate a Type II surface that prevents the formation of biofilm around dental implants.[62] Applying tetracycline permanently on implant surfaces efficiently eliminates bacteria.[63]

b) Antimicrobial Peptide Coating: Antimicrobial peptides (AMPs) are naturally occurring, positively charged peptides that exhibit broad-spectrum antibacterial activity against both Gram-positive and Gram- negative bacteria. These peptides also play a role in reducing the development of bacterial resistance.[64,65] One notable application of AMPs is their use in biofunctionalizing titanium surfaces to impart antibacterial properties.[66] This is particularly relevant in preventing infections associated with dental and orthopedic implants, such as peri-implantitis. A prominent example is the GL13K peptide, derived from human parotid secretory protein. GL13K has demonstrated bactericidal effects against various pathogens commonly associated with implant-related infections. When covalently immobilized onto titanium surfaces, GL13K coatings have shown resistance to mechanical, thermochemical, and enzymatic degradation, maintaining their antimicrobial efficacy over time.[67]

c) Polysaccharide Antibacterial Coating: Chitosan-immobilized implant surfaces have demonstrated effective antibacterial properties.[68, 69] Triethoxysilylpropyl Succinic Anhydride (TESPSA) facilitates the formation of stable bonds between chitosan and implant surfaces.[70] Additionally, polyelectrolyte multilayers composed of chitosan and hyaluronic acid have shown antibacterial activity against Staphylococcus aureus.[71] Furthermore, chitosan nanoparticle coatings conjugated with silver (Ag) on titanium surfaces have exhibited potential in inhibiting the growth of Streptococcus mutans and Porphyromonas gingivalis, as well as reducing biofilm formation and bacterial adhesion.[72, 73]

d) Antibacterial Properties of Metal-Element Components: The use of antibiotic coatings on dental implants has limitations, including the development of antibiotic resistance and a narrow spectrum of antibacterial activity.[74] To address these issues, alternative coatings incorporating metals such as silver, copper, and zinc have been explored. These metals possess inherent antibacterial properties and, when utilized in nanoparticulate forms, offer advantages suitable for implant applications.[75,76] Their unique mechanisms of action—such as disrupting bacterial cell membranes, generating reactive oxygen species, and interfering with metabolic pathways—enhance their effectiveness against a broad range of pathogens, including those resistant to conventional antibiotics. Cerium Oxide is a newly developed biomaterial that has demonstrated strong antibacterial activity potentially useful for dental implant applications.[77] It is a rare earth oxide showing free radical scavenging activity. Plasma-sprayed CeO2 coating on Ti-6Al-4V surfaces with high composition of Ce4+ valences significantly enhance antibacterial activity towards oral microbiota, along with increased osteogenic activity.[78]

2.5 Implant Surface Properties

These properties include surface roughness, wettability, yield strength, and fatigue strength, all of which are essential for the functionality of oral implants.[97] The biomechanical performance of dental implants is crucial for their clinical success. Implants should be designed to endure functional loads and possess adequate strength to prevent significant deformation or fracture. They must effectively transfer stress to the surrounding bone tissue to prevent bone atrophy. Additionally, the stress transferred should not exceed physiological limits to avoid bone resorption and fracture.[98, 99]

Elastic Modulus:

This is defined as the measurement of resistance offered by a material against the forces acting on it. The elastic modulus of a suitable dental implant material should be close to that of the bone (18 GPa). This minimizes the relative movement at implant bone interface, and also ensures a uniform distribution of stress at implant and bone interface.

Implants possessing optimal biomechanical properties ensure long-term success.[100] Upon implantation, the implant and the surrounding bone tissue form an interface, that establishes a bond that facilitates load transfer. This interface experiences various stresses, including compressive, tensile, and shear forces. While compressive strength in bone is significantly higher than tensile and shear strengths, excessive tensile and shear stresses can lead to fractures at the bone-implant interface.[101] The elastic modulus is a key factor in stress transfer between the implant and bone. A suitable elastic modulus ensures efficient stress distribution, preventing the "stress shielding" effect— a phenomenon where bone atrophy occurs due to reduced mechanical stimulation.[102] Commercial titanium implants typically have an elastic modulus around 110 GPa,

whereas human cortical bone ranges from 12 to 18 GPa. The range for cancellous bone

is between 0.1 to 4.5 GPa. This disparity can lead to stress concentration and hinder effective load transfer. [103, 104, 105]

Fatigue Strength:

Fatigue is the process by which a metal material undergoes changes in its properties due to repeated or cyclic stress. Fatigue life refers to the number of cycles or the duration a material or component can withstand such stress before failure occurs. The fatigue life of a material is influenced by factors such as the stress level, its resistance to damage, structural design, and surface characteristics.[106] Dental implants are subjected to repetitive stresses from chewing, which can lead to fatigue and potential fracture, thereby affecting their longevity. Mechanical fatigue may lead to implant and central screw loosening.[107] Therefore, studying the fatigue properties of dental implants is essential.

The primary factors influencing the fatigue failure of dental implants include the implant material, its size, and structural design. The fatigue strength of an implant is based on its dimensions and design. Implants with diameters more than 4.5 mm are considered large-diameter implants, while those with diameters less than 3.5 mm are small-diameter implants.[108] The stress level, resistance to damage, structure, and surface morphology of the material or part all affect its fatigue life.[109] Cylindrical threaded implants show a pitch of 0.8–1.2 mm.[110]

Surface Wettability:

Wettability is quantified by the contact angle (CA), which is the angle between the tangent line to a liquid drop's surface at the three-phase boundary and the horizontal solid's surface. In principle, the CA can range from 0 to 180 degrees (°). Water CAs lower than 90° designate surfaces as hydrophilic, and CAs very close to 0° as super hydrophilic. Surfaces with water CAs above 90° are considered hydrophobic, and those with CAs above 150° are termed superhydrophobic. Two primary techniques are employed to quantify surface wettability using contact angles (CAs):

a) Sessile Drop Method: A droplet of liquid is placed on a solid surface, and the contact angle is measured based on the droplet's shape. This method provides a direct assessment of the surface's wettability.

b) Tensiometry (Wilhelmy Plate Method): This indirect approach involves immersing a flat, vertically oriented sample into a liquid and measuring the force exerted on the sample. The contact angle is then calculated from this force measurement.[111]

Surface wettability significantly influences several biological interactions:

a) Protein and Macromolecule Adhesion: The initial conditioning of a surface by proteins and other macromolecules affects subsequent biological responses.

b) Cell-Surface Interactions: The interaction of hard and soft tissue cells with pre-conditioned surfaces is influenced by the surface's wettability.

c) Bacterial Adhesion and Biofilm Formation: The tendency of bacteria to adhere to surfaces and form biofilms is affected by the surface's wettability.

d) Osseointegration: The rate at which bone integrates with an implant is influenced by the wettability.

Surface Roughness:

Surface roughness is very important for the osseointegration of dental implants, influencing bone-to-implant contact and cellular behavior during healing.[112] Rough surfaces enhance bone cell adhesion, growth, and differentiation, thereby promoting successful osseointegration.[113]

Since the 1990s, blasting, oxidation and acid etching have been employed to achieve moderately rough surfaces on dental implants. These techniques have replaced the original smooth surfaces with moderately rough ones, which have shown improved bone anchorage due to enhanced biochemical bonding during osseointegration.

Nanotechnology involves the design and application of materials with structures at the nano meter scale.[114]

Nanostructured surfaces on titanium implants, such as those incorporating nanosized hydroxyapatite or TiO_2 particles, have been found to induce bone cell adhesion, growth, and differentiation by promoting specific protein interactions. These nanostructures can be applied through various methods, including the spontaneous formation of titanium oxide nanotubes on titanium surfaces.

The incorporation of nano roughness into titanium dental implants offers several advantages:

a) Increased Surface Area: Nano features enhance the surface area, providing more space for cell attachment and proliferation.

b) Improved Cell Attachment: Nanostructures facilitate better adhesion of bone cells to the implant surface, promoting early osseointegration.

c) Enhanced Biomechanical Interface: The nano roughness contributes to a stronger mechanical interlock, improving the overall stability

Table 3: Classification of Rough Surfaces[115]

Surface Roughness Category	Sa Range
Smooth	0-0.4 µm
Minimally rough	0.5-1 µm
Moderately rough	1-2 µm
Maximally rough	>2 µm

2.6 Implant Surface Microtopography

A good osseointegration between the dental implant and the bone is paramount to the success of the dental prosthesis that it would support, both in terms of quality as well as the expected years of functioning. It is known that titanium is a biocompatible material and osteoblasts readily attach to the surface, favouring osseointegration. However, several studies have shown that modification of the dental implant surface by means of additive or subtractive techniques favourably influence the osseointegration process by affecting the surface roughness. This has led to the advent of implants with different types of microtopographic features.

Subtractive Alterations on Titanium Surface

This form of implant surface modification boasts of enhancing effective osseointegration by means of increasing the mean surface area on a microscopic level as also the surface free energy. Various modalities are employed to this effect utilizing mechanical, chemical or electro chemical means to achieve a micro- or nano- rough texture. Sand blasting; acid etching; combined sand blasting – acid etching technique; anodization; as well as the use of laser irradiation and more recently, hydrothermal methods have been explored. Sand Blasting technique involves the use of compressed air to propel micro particles of aluminium oxide, silicon oxide, titanium oxide or hydroxyapatite onto the titanium implant surface leading to surface abrasion that causes surface roughness ranging from 0.3 to 3 microns. It is interesting to note that the surface roughness of polished titanium is 0.04 microns. Increase in roughness enhances attachment of cells and proteins to facilitate new bone formation. At the same time, this also leads to enhanced hydrophobicity apart from diminished friction, attrition, and friction coefficient of $TiAl6V4$ in saliva. Sand blasted acid etch technique involves the use of acids, including hydrochloric acid, nitric acid, sulphuric acid and hydrofluoric acid to create micro pits on the titanium implant surface by virtue of chemical reaction. It is often used in conjunction with sand blasting to remove the residual layer and achieve superior properties as compared to sand blasting treatment done alone. In the anodization technique, pure titanium is used as anode while an inert material is used as cathode, the two being immersed in an electrolyte solution, and with the use of specific current and voltage, oxidation is induced on the titanium surface. This results in the formation of a stable titanium oxide layer which is porous, thus facilitating the attachment of stem cells, favouring osseointegration. Laser radiation technique involves the use of laser beam to create micro and nano relief structures on the surface of the titanium oxide layer by directing a laser on the surface, that causes on-site melting. Laser machining also minimizes the viability of surface contaminants.

Additive Alterations on Titanium Surface

This involves the deposition of particles of varying size on titanium dental implant in order to enhance the surface area as well as the biochemical properties of the implant for enhancement in osseointegration and the overall performance of the implant. HA is the bioceramic substance of choice.[40] Plasma spraying is the most common method for applying HA coatings on the surfaces of titanium implants. This increases the roughness of dental implant.

2.7 Implant Surface Alterations Treatments / Classification

Osseointegration is essential for the long-term success of endosseous dental implants. The process is influenced by various factors, including the implant's surface characteristics such as energy, roughness and composition.[116] Surface modification techniques are employed to enhance these characteristics, thereby optimizing osseointegration.

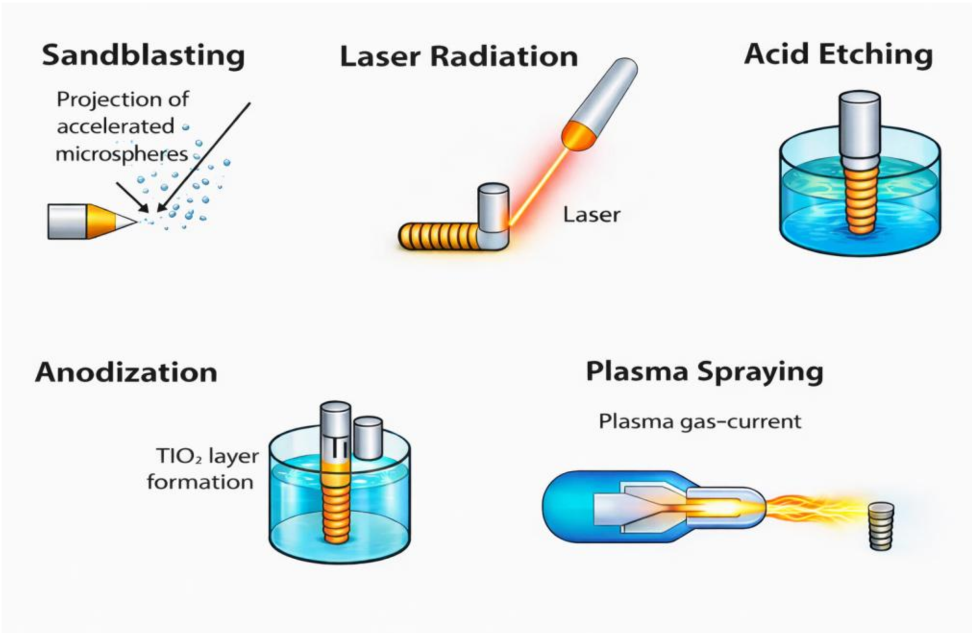

Fig 5 Implant Surface Modification Methods [117]

Table 4: Surface Treatments and Various Implant Systems Available Commercially[115]

Surface Treatment	Description	Implant System/Surface
Blasted and acid washed/etched	Implants undergo a blasting process. Afterward, the surface is either washed with non-etching acid or etched with strong acids. RBM-treated implants like the Hahn Tapered Implants have the advantage of resorbable, biocompatible blast media.	Hahn Tapered Implants, DENTSPLY Implants FRI-ALIT and FRIADENT plus, Straumann SLA, Inclusive Tapered Implants
Anodized	This electrochemical process thickens and roughens the titanium oxide layer on the surface of implants.	Nobel Biocare TiUnite

Acid etched	Etching with strong acids increases the surface roughness and the surface area of titanium implants.	BIOMET 3i OSSEOTITE and NanoTite
Blasted	Particles are projected through a nozzle at a high velocity onto the implant. Various materials such as titanium dioxide, aluminum dioxide, and HA are often used.	DENTSPLY Implants ASTRA TECH TiObiast, Zimmer Dental MTX
HA coated	HA is an osteoconductive material that has the ability to form a strong bond between the bone and the implant.	Implant Direct (various), Zimmer Dental MP-1
Laser ablation	High-intensity pulses of a laser beam strike a protective layer that coats the metallic surface. As a result, implants demonstrate a honeycomb pattern with small pores.	BioHorizons Laser-Lok
Titanium plasma sprayed	Powdery forms of titanium are injected into a plasma torch at elevated temperatures.	Straumann ITI titanium plasma-sprayed

Subtractive Modifications:

Here, techniques are employed to influence the surface continuity of the implant. It includes the following:

a) Sand Blasting: Sandblasting is a surface treatment technique for dental implants that involves bombarding the implant surface with abrasive particles such as alumina (Al_2O_3) or titanium oxide (TiO_2) under high pressure. This process increases the surface roughness and area, enhancing mechanical interlocking between the implant and bone, which promotes osseointegration—the integration of the implant with the bone.[118] It also increases contact angle for improving hydrophilicity.[132] The effectiveness of the technique is influenced by factors, including type and particle size of abrasives, the pressure applied, and rotation speed during the process. While sandblasting enhances osseointegration, it may also introduce challenges. Residual abrasive particles, particularly Al_2O_3, can remain on the implant surface even after cleaning procedures. These residues have been linked to potential adverse effects, including inhibition of osteoblast differentiation and bone mineralization. It can also lead to contamination by microbes and subsequent peri implantitis.[132]

b) Acid Etching: The proposed method aims to modify the surface of implants without leaving the residues commonly associated with sandblasting, ensuring a uniform treatment across the implant's surface. This technique creates micro-pits on the implant surfaces. The etching process is influenced by factors such as the acid mixture used, the duration of etching, and the temperature of the bath. This mainly uses hydrochloric acid (HCl), sulfuric acid (H_2SO_4), hydrofluoric acid (HF), or nitric acid (HNO_3). These may be used individually or in various combinations. These factors—acid composition, etching time, and bath temperature—affect the rate of the etching process.[138] Implants treated with acid etching have shown a greater bone-to-implant contact.[139]

c) Sand Blasted Large Grit Acid Etched (SLA) Surface: This involves a dual-step process. Initially, the implant surfaces are subjected to sandblasting using large grit particles (0.25–0.5 mm) to create macro-roughness. This is followed by an acid-etching treatment with a mixture of hydrochloric (HCl) and sulfuric (H_2SO_4) acids at elevated temperatures to introduce micro-roughness. This combination enhances the surface area and promotes better cell adhesion, facilitating improved osseointegration. Implants treated with this Sandblasted, Large grit, Acid- etched (SLA) method demonstrate superior biomechanical stability compared to those with acid-etched-only surfaces. For instance, in a study involving miniature pigs, SLA implants displayed significantly higher removal torque values—186.8 N·cm— compared to 95.7 N·cm for acid-etched-only implants, indicating stronger bone- implant integration .[119] Examples of dental implants utilizing this SLA surface treatment include Straumann SLA, Hahn Tapered Implants, DENTSPLY Implants FRIALIT and FRIADENT plus, and Inclusive Tapered Implants. These implants are designed to achieve rapid and stable osseointegration, making them suitable for early loading protocols in dental rehabilitation.

d) Anodization: The anodization process involves treating titanium implant surfaces using acids such as phosphoric acid (H_3PO_4), sulfuric acid (H_2SO_4), nitric acid (HNO_3), and hydrofluoric acid (HF) under high current densities (e.g., 200 A/m^2) or voltage potentials (e.g., 100 V). This treatment causes formation of a thick oxide layer exceeding 1,000 nm in thickness. The anodization process alters the crystalline microstructure of the titanium dioxide (TiO_2) layer, leading to the creation of micropores.[120,121] These micropores facilitate the formation of the anatase phase of TiO_2, which is considered one of the most important phases of TiO_2.[145] Additionally, anodization enhances blood-clot retention, promoting favorable osseointegration in dental implants.[147, 148]

e) Laser Peening: Laser peening is a surface modification technique that employs high-intensity nanosecond laser pulses (5–15 GW/cm^2, 10–30 ns) to generate shock waves on a material's surface. This process induces compressive residual stresses, enhancing the material's resistance to fatigue, wear, and stress corrosion cracking. When applied to commercially pure titanium (cpTi) biomedical implants, laser peening creates micro molecular patterns approximately 20 μm in width and 7 μm in depth. These microstructures improve the implant's surface properties, including corrosion resistance, mechanical strength, and fatigue resistance.[122]

Additive Modifications:

These modifications focus on the additional application of different particles in micro or nano sizes or coating of drugs or biomimetic coatings that work to enhance the osseointegration of dental implants. These include the following:

a) Hydroxyapatite Coating: The release of calcium phosphate from hydroxyapatite (HA)-coated implant surfaces into surrounding tissues increases the saturation of local fluids, promoting the formation of a biological apatite layer on the implant. This layer, enriched with endogenous proteins, acts as a matrix that supports osteogenic cell attachment and proliferation.[123] Various methods have been developed to apply HA coatings to endosseous implants, including plasma spraying, vacuum deposition, sol-gel and dip coating techniques, electrolytic processes, hot isostatic pressing, high-velocity oxygen fuel spraying, frit enameling, ion-assisted deposition, sputter coating, and nano-HA coatings.[124] However, HA-coated implants may be more susceptible to bacterial colonization compared to uncoated implants or natural teeth. The increased surface roughness of HA coatings can promote enhanced biofilm growth, potentially leading to peri- implantitis.[115]

b) Biomimetic Coatings: The biomimetic technique involves submerging a pretreated implant into a supersaturated calcium phosphate solution under physiological conditions— specifically at 37°C and a pH of 7.4.[125] This method enables the formation of a calcium phosphate coating that can integrate osteogenic agents such as Bone Morphogenetic Protein-2 (BMP-2), Bone Morphogenetic Proteins (BMPs), and Growth and Differentiation Factors (GDFs). These agents are co-precipitated into the coating, allowing for their gradual release upon exposure to the biological environment, thereby enhancing osteogenic activity. Biomimetic agents are materials designed to elicit specific cellular responses through interactions with scaffold-tethered peptides derived from extracellular matrix (ECM) proteins. These interactions can be achieved by incorporating cell-binding peptides into biomaterials via chemical or physical modifications. Such strategies aim to replicate natural cellular environments, thereby promoting desired biological outcomes.

c) Bioactive Glasses: Bioactive glass coatings possess osteoinductive properties, promoting bone formation at the interface of implant and adjacent bone.[28] Porosity and surface roughness of these coatings, influenced by deposition methods, are critical factors affecting their bioactivity.[126] Based on their chemical composition, bioactive glasses are categorized into silicate-, borate-, and phosphate-based types.[168] Among these, silicate-based bioactive glasses, such as BG 45S5 and 1393, are the most commonly utilized due to their proven effectiveness in clinical applications.

d) Coating Of Bioactive Drugs: Pharmacological agents, such as bisphosphonates, are applied to implant surfaces to enhance bone density in areas with highly cancellous bone.[127] Implants treated with specific drugs exhibit improved osseointegration and possess antibacterial properties against pathogens at the implantation site.

These drugs get leached by the following primary mechanisms:

1) Diffusion-Controlled Release: Drug molecules move from areas of higher concentration to lower concentration through the implant material.

2) Solvent-Controlled Release: This includes osmotic or swelling phenomena where the presence of solvents causes the implant to release drugs.

3) Chemically Controlled Release: Involves processes like biodegradation of polymers and cleavage of the bonds between the biomaterial and drug, leading to drug release.

4) pH-Sensitive Release: Drug release is triggered by changes in pH levels at the implant site.

2.8 Subtractive Surface Alterations

Subtractive surface alterations on dental implants involve techniques that remove material to modify the implant's surface properties—such as roughness, wettability, topography, or chemistry—to enhance bone-to-implant contact and improve osseointegration. These modifications are particularly beneficial for titanium implants, which are commonly used in dental applications.

Sand Blasting:

Sandblasting is a widely used technique for modifying the surface of titanium implants to enhance osseointegration. This process involves propelling abrasive particles, such as alumina (Al_2O_3) or titanium dioxide (TiO_2), onto the implant surface at high velocities, resulting in a micro-roughened texture. [128] The effectiveness of sandblasting depends on several factors, including the type, size, and shape of the abrasive particles, distance between the projection nozzle and the implant surface, the applied pressure, the duration of exposure, and the size of the projection area.[129] These parameters collectively influence the surface roughness, which plays a crucial role in biological interactions.[130]

Research has demonstrated that titanium implants subjected to sandblasting exhibit improved clinical outcomes compared to those with smooth surfaces. Specifically, sandblasted implants have shown higher bone-to-implant contact (BIC), indicating better integration with the surrounding bone tissue.[131] The increased surface roughness enhances the hydrophilicity of the implant, which is beneficial for osteoblast adhesion and proliferation.[132] However, it's important to note that while sandblasting can improve osteoblast activity, it may also lead to microbial contamination if residual abrasive particles remain on the implant surface.[133, 134]

Moreover, the technique of sandblasting can introduce sharp edges on the implant surface, which may adversely affect the adhesion of bone tissue. [135, 136] To mitigate these issues, additional treatments, such as acid etching, are often employed. Acid etching can remove residual particles and smooth out sharp edges, thereby enhancing the overall biocompatibility of the implant surface. Higher torque application is required for removal compared to turned implants.[137]

Acid Etching:

Acid etching is a technique that employs acidic solutions to chemically alter or remove material from a surface, resulting in a textured or patterned appearance. This method was introduced for modifying the surfaces of implants, without leaving behind any residues typically associated with sandblasting. The process involves immersing the implant in acids such as hydrochloric acid (HCl), sulfuric acid (H_2SO_4), hydrofluoric acid (HF), or nitric acid (HNO_3), either individually or in various combinations. This treatment creates micro-pits on the implant surface. Factors like the acid mixture, etching duration, and bath temperature influence the etching rate. Implants treated with HCl and H_2SO_4 demonstrated a higher removal torque force (20.5 Ncm) compared to turned implants (4.95 Ncm).[138] Dual etching, particularly the combination of HF and HCl, is more effective in producing rough surfaces on implants. Etched implants achieved a bone-to-implant contact (BIC) of 62.5%, surpassing the 39.5% BIC observed in turned implants.[139] Commercial implants like Osseotite® (Zimmer Biomet, Warsaw, IN, USA) and Steri-Oss Etched® (Nobel Biocare, Zürich-Flughafen, Switzerland) utilize the acid etching method. Osseotite® has shown a success rate exceeding 96%.[140,141] Surface characteristics may be modified by adjusting the concentration and type of acid used, as well as its exposure time and temperature, which affect the creation of irregularities of varying sizes.[142] Implants subjected to this technique exhibit enhanced roughness and osteogenic responses, including increased proliferation, adhesion, and differentiation of osteogenic cells.[143] Moreso, uncontrolled etching, whether excessive or insufficient, can lead to variations regarding the surface properties.[144]

Anodization:

Anodization is a technique that thickens the layer of titanium oxide (TiO_2) on the implant surface, creating a micro- or nano-porous structure. Anatase and rutile are two important phases of TiO2.[145] This modification enhances surface roughness and bioactivity, promoting better osseointegration. Factors such as applied voltage, electrolyte composition, temperature, and anodization time influence the outcome.[146] It also favours blood clot retention.[147, 148] While anodization improves biological outcomes, challenges remain regarding the mechanical stability of anodized implants.[149]

Laser Peening:

Laser peening, or laser shock peening, uses high-energy, short-duration laser pulses to induce compressive residual stresses in metals. This process refines grain structures, increases surface hardness, and improves fatigue strength and resistance to stress corrosion cracking.[150, 151] Laser peening produces a controlled surface roughness and has shown to stimulate bone growth at the surface. It has been reported to achieve more significant surface enhancement than grit blasting.[152] Adsorption of radiation causes on-site melting of the metal.[153] Thereby, microchannels are formed that hasten bone healing.[154]

Thus, subtractive surface alterations, including sandblasting, acid etching, anodization, and laser peening, are effective techniques. Each method has its advantages and considerations, and the choice of technique depends on factors such as desired surface characteristics, implant material, and clinical requirements.

2.9 Additive Surface Alterations

Surface modifications of dental implants involve applying materials often in nanoparticle form to enhance the chemical and physical properties of the implant, including wettability, surface roughness, biocompatibility, and osseointegration. These modifications aim to improve the implant's overall survival and the performance rates.[154,155]

Inorganic Coatings:

a) Hydroxyapatite-Based Coatings: HA, a stable biological mineral form of calcium apatite, is non-inflammatory and non- immunogenic. It stimulates osteoblast activity for bone formation. Micro-arc oxidation techniques can create porous HA coatings on titanium alloys, enhancing biomechanical properties and bone formation.[156]

Nanotechnology applied to HA particles can improve surface area and adsorption features, leading to better interaction with host bone.[157]

b) Calcium Phosphate-Based Biomimetic Coatings: Calcium and phosphate ions released from bioactive coatings play a significant role in promoting bone healing and tissue integration.[158] These ions enhance the mineralization of interface tissues and stimulate the expression of osteogenic factors, leading to improved adhesion, proliferation, and overall bone regeneration. The combination of acid-etching techniques with discrete crystalline deposition facilitates the impregnation of these ions during the sol-gel process, thereby promoting osteoconduction and exhibiting long-term durability with minimal bone resorption.[159]

Table 5: Different Techniques to Deposit Hydroxyapatite Coating [3]

Technique	Thickness	Advantages	Disadvantages
Plasma Spraying	<20 µm	Rapid deposition; sufficiently low cost; lost bone	Poor adhesion; non-uniformity in coating density; extreme high temperature up to 1200^0C phase transformation; increase in residual stress
Thermal Spraying	30 - 200 µm	High deposition rates; low cost[1]	Line-of-sight technique; high temperatures induce decomposition; rapid cooling produces amorphous coatings; lack of uniformity; crack appearance; low porosity; coating spalling and interface separation between the coating and the substrate[2]
Sputter Coating[3]	0.5 - 3 µm	Uniform coating thickness on flat substrates; dense coating; homogenous coating; high adhesion	Line-of-sight technique; expensive and time-consuming; produces amorphous coatings; low crystallite accelerates the dissolution of the film in the body

Pulsed Laser deposition	0.05 – 5 mm	Coating that is crystalline and amorphous; dense and porous; ability to produce wide range of multilayer coatings from different materials; high crystalline HA coating; ability to restore complex stoichiometry; high degrees of control on deposition parameters	Line-of-sight technique; expensive and time-consuming; produces amorphous coating; low crystalline; needs surface pretreatment; lack of uniformity
Dip Coating	<1 μm	Inexpensive; coatings applied quickly; can coat complex substrates; high surface uniformity; good speed of coating	Requires high sintering temperatures; thermal expansion mismatch; crack appearance
Sol-Gel	0.1 - 2.0 μm	Can coat complex shapes; low processing temperatures; relatively cheap because coatings are very thin; simple deposition method; high purity; high corrosion resistance; fairly good adhesion	Some processes require controlled atmosphere processing; expensive raw materials; not suitable for industrial scale; high permeability; low wear resistance; difficult to control the porosity
Electrophoretic Deposition	0.2 - 2.0 mm	Uniform coating thickness; rapid deposition rates; can coat complex substrates; simple setup; low cast; high degree of control on coating morphology and thickness; good mechanical strength; high adhesion for n-HA	Difficult to produce crack-free coatings; requires high sintering temperatures; HA decomposition during sintering stage
Hot Isostatic Pressing	0.2 - 2.0 mm	Produces dense coatings; produces net-shape ceramics; good temperature control; homogeneous structure; high uniformity; high	Cannot coat complex substrates; high temperature required; thermal expansion mismatch; elastic property differences; expensive;

		precision; no dimensional or shape limitation	removal/interaction of encapsulation material
Ion Beam-Assisted Deposition	<0.03 µm	Low temperature process; high reproducibility and reliability; high adhesion; wide atomic intermix zones are coating-to-substrate interface	Crack appearance on the coated surface

c) Magnesium-Based Biomimetic Coatings: Magnesium alloys, known for their biomimetic and biodegradable properties, have been explored for dental implants. Magnesium phosphate salts promote osteoblast function and faster bone formation.[160,161] Techniques like microwave-assisted preparation and laser peening have been used to produce magnesium-doped coatings, encouraging bone formation and increasing bone volume.[162,163,164]

d) Carbon Nanotube-Based Coatings: Carbon nanotubes (CNTs) can serve as an external coating for implants. This leads to an increase in the roughness and also promotes cell adhesion and proliferation. However, their interaction with host cells and nucleic acids raises concerns about cytocompatibility.[165]

e) Silver-Based Coatings: Silver possesses oligodynamic antimicrobial properties, exhibiting bactericidal and bacteriostatic effects at low concentrations. Silver ions (Ag^+) disrupt cell wall permeability, and induce condensation of the DNA, demonstrating broad-spectrum activity against both Gram-positive and Gram-negative bacteria. Coatings incorporating hydroxyapatite (HA) and silver, such as HA–silver composites, plasma- sprayed silver-implanted HA, and silver-loaded gelatin microspheres within porous titanium, have been developed to control in situ metal release.[166] Silver nanoparticles offer a promising alternative, facilitating prolonged Ag^+ ion release through chemical binding after a silanization reaction on hydroxylated surfaces. These nanoparticles also exhibit osteoinductive properties at the implant-device interface.[167]

f) Bioactive Glasses: Bioactive glasses are specialized glasses capable of bonding to bone. Upon exposure to human fluids, they can form hydroxyapatite (HA) precipitates on the glass substrate, facilitating bonding to host bone. The porosity and roughness of bioactive glasses, influenced by deposition techniques, significantly affect their bioactivity. Depending on their chemical composition, bioactive glasses can be categorized into silicate-, borate-, and phosphate-based types.[168] Silicate-based bioactive glasses, such as BG 45S5 and 1393, are among the most commonly used.

Organic Coatings:

a) Growth Factors-Based Coatings: Important growth factors used as a coating on implants to favour osseointegration are the bone morphogenetic protein and vascular endothelial growth factor. VEGF favours angiogenesis, enhances osteoblast proliferation and increases alkaline phosphatase (ALP) activity. BMPs influence the alveolar bone regeneration and enhanced bone- implant contact by affecting the osteogenic and mesenchymal stem cells.[169]

b) Extracellular Matrix Proteins and Polysaccharides: Proteins like collagen, elastin, and fibronectin, along with polysaccharides such as hyaluronic acid, can be applied as implant coatings. These coatings promote bone formation at the implant–bone interface, enhancing osseointegration.[170] Mussel adhesive proteins have also been utilized to promote differentiation of bone-forming cells and favor cell adhesion and proliferation.[171]

c) Drug-Releasing Coatings: Implants coated with drugs can improve osseointegration and provide antibacterial effects. Drug release mechanisms include diffusion-controlled, solvent-controlled, chemically controlled, and pH-sensitive mechanisms. In order to incorporate antibiotics into implant coatings, techniques like spray coating or dipping may be used. This helps to replace systemic agents and reduce toxic effects. For instance, gentamycin-loaded coatings have shown favourable activity against implant-related infections.[172] Vancomycin and doxycycline also produce antibacterial effect. [173, 174] Thus, surface modifications of dental implants using various inorganic and organic coatings aim to enhance their physical, chemical, and biological properties, leading to improved osseointegration, reduced infection rates, and overall better performance and longevity of the implants.

2.10 Biomimetic Surface Alterations

The term "biomimetics" introduced by Otto Schmitt in the 1950s, refers to the practice of emulating biological systems and processes to develop innovative technologies. In the context of biomaterials, biomimetic agents are engineered substances designed to interact with the extracellular matrix (ECM) and elicit specific cellular responses. These agents often incorporate peptides that mimic those found in the ECM, facilitating cellular adhesion and promoting tissue regeneration. Biomimetic materials aim to replicate the natural structure and function of dental tissues. By integrating bioactive components that encourage remineralization and mimic the mechanical properties of natural teeth, these materials enhance the longevity and effectiveness of dental restorations. Biomimetics encompasses the development of materials and systems that imitate natural biological functions, with applications ranging from tissue engineering to dental restorative practices.

Desirable properties of a biomimetic agent, for use in dental implantology[175]

a) Ability to induce differentiation of the appropriate cells for enhancing new bone formation

b) Ease of processing

c) Resorbability in response to osteogenic action

d) Should not produce any untoward immune response in the host tissues

e) Should be cost-effective

Examples of currently available biomimetic materials:

a) Bioceramics / Hydroxyapatite – Based Biomimetic Coatings

b) Calcium Phosphate – Based Biomimetic Coatings

c) Bioactive Proteins

d) Magnesium - Based Biomimetic Coatings

e) Growth factors

f) Extracellular Matrix Proteins

g) Graphene - Based Biomimetic Coatings

h) Nano- Diamond Coatings

i) Fluoride Ions

j) Chitosan

Bioceramics / Hydroxyapatite-Based Biomimetic Coatings:

Hydroxyapatite (HA) is a mineral derivative of calcium apatite, known for its stability and biocompatibility. It is non-inflammatory and non-immunogenic, promoting osteoblast activity essential for bone formation. Enhancing its bioactivity, micro-arc oxidation (MAO) techniques can create a porous HA coating on titanium alloys, improving biomechanical properties and supporting bone integration. Nanotechnology has been employed to modify HA particles, facilitating the development of a biomimetic osteomatrix by combining HA with collagen and titanium dioxide. The nanoscale modifications increase surface area and adsorption capabilities, leading to improved interaction with host bone tissue.[178]

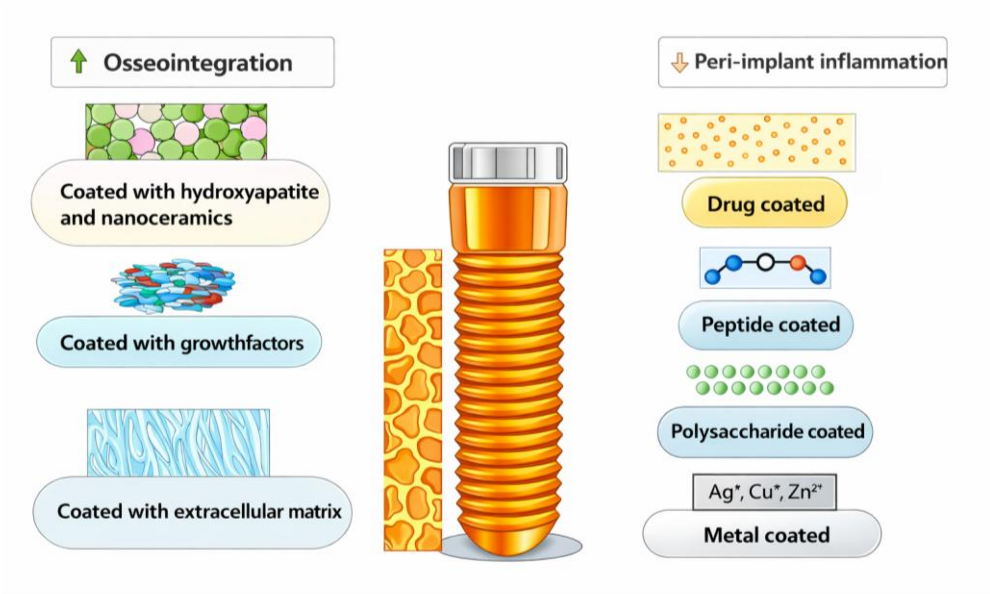

Fig 6: Biomimetic Coatings[176]

Calcium Phosphate-Based Biomimetic Coatings:

Calcium phosphate (CaP)-based coatings on implants release calcium and phosphate ions upon implantation, leading to the formation of an apatite layer. These coatings have been associated with enhanced clinical outcomes, including increased bone-to- implant contact and long-term implant success. Inorganic nano topography on implant surfaces influences cell behaviour during healing, enhancing gene expression related to osteogenesis, mineralization, adhesion, and proliferation.[179]

Bioactive Proteins (Bmp):

Bone Morphogenetic Proteins (BMPs), first identified by Urist in the 1960s, are proteins that induce undifferentiated mesenchymal cells to differentiate into osteoblasts, facilitating de novo bone formation at implantation sites. Recombinant human BMP-2 (rhBMP-2) has demonstrated effectiveness in promoting the initial integration of dental implants.[180]

Magnesium-Based Biomimetic Coatings:

Magnesium (Mg)-based biomimetic coatings are biodegradable and have been explored for their osteoinductive properties. Magnesium phosphate salts exhibit favorable rate of resorption and dissolution, compared to calcium salts.[60] These coatings enhance osteoblast function and accelerate bone formation.[161] Techniques such as microwave-assisted preparation and selective laser melting (SLM) have been utilized to create Mg-doped HA coatings, demonstrating increased bone volume compared to control groups.[162163, 164]

Growth Factors:

Growth factors are a diverse group of proteins that regulate various biological processes, including cell proliferation, differentiation, and chemotaxis. Platelet-Rich Plasma (PRP), rich in growth factors like TGF-β, platelet-derived growth factor, and insulin-like growth factor-1, has been proposed as a coating for implants. However, challenges such as the short half-life of these proteins complicate their application.[181]

Extracellular Matrix Proteins:

Type I collagen serves as a biomimetic agent for implant surface coating. The RGD peptide sequence, present in several extracellular matrix proteins, plays a crucial role in cell adhesion by interacting with integrins. Coating implants with RGD peptides has been shown to enhance osteoblast adhesion, thereby promoting early osseointegration.[182]

Graphene-Based Biomimetic Coatings:

Graphene oxide (GO) and reduced graphene oxide (rGO) have been explored as implant coatings due to their two-dimensional structure and functional groups, which facilitate surface modification. Coating titanium surfaces with GO has been shown to stimulate cell proliferation and adhesion.[183] Additionally, GO exhibits antibacterial properties, and when combined with silver nanoparticles, it demonstrates activity against various bacterial strains.[184] Furthermore, GO coatings have been utilized to deliver osteoinductive agents like minocycline, enhancing both antibacterial and osteogenic activities.[185]

Nano - Diamond Coatings:

Oxygenated and boron-doped nanocrystalline diamond coatings on titanium implants have been shown to increase cellular proliferation and adhesion. Larger particles (approximately 100 nm) tend to yield better results compared to smaller ones.[186]

Fluoride Ions:

Fluoride treatment of dental implants involves immersing them in a fluoride-containing solution, typically hydrofluoric acid, to create a fluoride layer on the surface. This layer increases surface energy and roughness, promoting better cell attachment and proliferation, leading to enhanced osseointegration. Fluoride ions can interact with hydroxyapatite to improve its crystallinity and reduce dissolution rates.[187]

Chitosan:

Chitosan, a natural polymer derived from glucosamine and N-acetylglucosamine, serves as an effective scaffold for osteoblasts. It facilitates the deposition of the extracellular matrix and enhances the differentiation of pre-osteoblasts into mature osteoblasts, thereby supporting bone regeneration. Chitosan's osteoconductive properties make it a promising biomimetic agent for coating titanium implants.[188]

2.11 Implant Surface Contamination and Decontamination

Exposure of dental implant surface to the oral cavity may occur either by iatrogenic means or following the inflammation of the peri implant mucosa. The oral cavity is teaming with microorganisms that form the adherent biofilm on hard surfaces present. Thereby, these implant surfaces provide a niche for the microorganisms to adhere and proliferate. This may be termed as contamination. Contaminants may be classified into inorganic and organic types.

Inorganic Contaminants:

a) Reactive Oxygen & Nitrogen Species: These are free radicals derived from nitrogen and oxygen species. They act as intercellular and intracellular messengers. RONS have beneficial effects at low or moderate concentrations which results in the angiogenesis, proliferation and re- epithelialisation of cells in the gingival tissues. Very high levels of reactive oxygen species (ROS), may cause adverse effects, such as mitochondrial dysfunction.[189]

b) Calcium: Treatment of titanium with sodium hydroxide reagent and heat treatment induces apatite formation, that improves the osseointegration properties of titanium. However, calcium contamination may occur from the calcium in the reagent, which inhibits the formation of apatite by suppressing the release of sodium ion into the surrounding. Calcium contamination level of 0.0005% was also sufficient to inhibit the apatite formation. [190]

c) Phosphorus: Treatment of titanium with sodium hydroxide reagent and heat treatment induces apatite formation, that improves the osseointegration properties of titanium. However, calcium contamination may occur from the calcium in the reagent, which inhibits the formation of apatite by suppressing the release of sodium ion into the surrounding. Calcium contamination level of 0.0005% was also sufficient to inhibit the apatite formation. [190]

d) Chlorine: Several ions such as potassium (K), sodium (Na), nitrogen (N), and chloride ions, as well as proteins are present in the saliva. Chloride ions can compromise the oxidation layer of dental implants, resulting in corrosion at the implant-abutment interface.[192]

e) Sulphur: Residual contaminants such as sulphates, fluorides, magnesium oxides, silicates, and calcium oxides can remain on titanium implant surfaces after sandblasting and etching procedures.[193] A dual acid etching method involving hydrofluoric acid followed by sulfuric acid treatment can produce a titanium surface with dual roughness. This enhanced surface topography promotes osteoblast adhesion, proliferation, and differentiation, thereby improving osseointegration. Additionally, non-thermal plasma treatment has been shown to effectively remove sulphur contaminants from titanium surfaces, enhancing their biological compatibility.[194]

f) Sodium: Titanium implants treated with saline and sodium hypochlorite (NaOCl) often retain trace amounts of sodium. These residual sodium ions can interfere with the oxygen cathodic reaction by obstructing active sites on the implant surface. This blockage prevents foreign ions, such as iron or chromium, from catalyzing the oxygen reduction reaction, leading to an increased dissolution rate of the titanium implant. Consequently, this dissolution can hinder the re-osseointegration process, compromising the implant's stability and long-term success.[195]

g) Aluminium: During sandblasting and acid etching, aluminium and fluoride ions are deposited onto surfaces. The oxidized form of aluminium, known as alumina, is stable in physiological fluids and elicits minimal tissue response. Consequently, a coating of alumina may improve the resistance to corrosion in dental implants.[196] However, elevated concentrations of residual aluminium oxide (Al_2O_3) may adversely affect the osseointegration process.

h) Silicon: Silicon plays a crucial role in the metabolism of bone. It promotes osteoblast differentiation, synthesis of type I collagen, and also facilitates mineralization.[197] Incorporating tetraethyl orthosilicate (TEOS) into sol–gel-derived silicon coatings leads to hydrolytic degradation, releasing silicon compounds that enhance osteoinductive properties and promote direct contact between new bone and titanium implants.[198] Additionally, silicon-based coatings possess properties that prevent bacterial infections post-implantation, thereby improving patient outcomes.[17]

i) Zinc: Zinc is an essential trace element that stimulates osteoblast proliferation, bone formation, and biomineralization. It also exhibits antibacterial properties, making it an attractive option for incorporation into titanium surfaces in dental implants to enhance bioactivity. Studies have shown that zinc-modified titanium surfaces promote osteoblast differentiation and mineralization in vitro, suggesting their potential for improving bone regeneration therapy.[199]

j) Flouride: Traces of fluoride can be introduced during the acid-etching process. However, high concentrations of fluoride ions can cause degradation of titanium oxide layer, leading to localized corrosion.[200] The concentration of fluoride ions and pH of the environment determine the severity of corrosion. For instance, at a concentration of 3 ppm of fluoride ions, titanium alloy may undergo discoloration, and at concentrations exceeding 20 ppm, titanium oxide may degrade, compromising the implant's integrity.

k) Hydrogen: Acid combinations, such as hydrofluoric acid and nitric acid, are commonly used to remove the oxide layer from titanium surfaces. In the process of hydrofluoric acid pretreatment, the oxide layer is attacked, and titanium reacts to form soluble titanium fluorides and hydrogen ions. When the concentration of free hydrogen ions becomes saturated, titanium hydride forms. Titanium hydride can significantly alter the mechanical properties of titanium, leading to surface embrittlement. However, hydrogen-rich films have been shown to increase the adhesion of keratinocytes, thereby improving the osseointegration.[201]

Organic Contaminants

a) Hydrocarbons: Titanium implant surfaces, whether newly photo-functionalized or previously treated, tend to accumulate hydrocarbons within four weeks, regardless of the surface treatment applied.[202] This hydrocarbon presence can hinder osteoblast attachment, as cell adhesion proteins struggle to bind to such surfaces.[203] Hydrocarbons from air, water, or cleaning solutions may easily adsorb onto the titanium surface. This reduces its hydrophilicity. Studies have shown that increased carbon content on titanium surfaces suppresses osteoblast functions like cell attachment, spreading, and mineralization.[204]

b) Carboxylates: Coating titanium with carboxylated carbon nanotubes (MWCNT-COOH) enhances osteoblast adhesion, proliferation, and differentiation.[205] However, contaminants containing carboxyl groups can increase the hydrophilicity of titanium, potentially reducing bonding with proteins and decreasing cell attachment. High-energy photons, including non-thermal plasma and UV light, are effective in removing carboxylate contaminants.[206]

c) Salts Of Organic Acids: Accumulation of organic acids can lead to an acidic environment, inducing corrosion and discoloration of titanium implants.[207] Therapeutic treatments using solutions containing citric or phosphoric acids are often utilized to eradicate bacteria from titanium surfaces, but they can also decrease the pH and accelerate corrosion.

d) Nitrogen Residue: Nitrogen and other elemental residues may originate from bacterial plaque, bolus, soft tissue, or protein compounds in saliva adhering to the titanium surface. Treatments like nitrogen plasma have been shown to increase the content of –OH groups on the titanium surface, promoting cell attachment and enhancing osteogenic differentiation.[208]

e) Bacteria: Bacterial colonization can damage the titanium oxide layer, leading to microbial corrosion. The acidic waste products from microbes create an acidic environment, potentially resulting in inflammation and peri-implantitis. Contamination from bacteria can occur on implants during surgery, affecting osseointegration and prognosis.[209] Combined use of antiseptics like chlorhexidine digluconate (CHX) or hydrogen peroxide (H_2O_2) and laser photodynamic therapy (PDT) has been useful in eliminating bacterial biofilms, such as those formed by Staphylococcus aureus, from titanium surfaces.[212]

2.12 Effect of Surface Modifications on Titanium

Several types of modifications are carried out on titanium implant surfaces. They aim at improving the osseointegration or mechanical and biological properties for improved performance. Commercially pure titanium (CP Ti) is extensively utilized in dental and orthopaedic implants owing to its exceptional biocompatibility, superior corrosion resistance, favorable mechanical properties, and capacity for osseointegration.[211] To enhance osseointegration, various surface modification techniques are employed. However, techniques like grit-blasting can be imprecise and challenging to control, potentially introducing contaminant particles onto the implant surface.[212]

Effect of Acid Etching / Sand Blasting Acid Etching

Acid etching involves creating micron-level indentations on the implant surface by immersing it in a mixture of hydrochloric acid (HCl) and sulfuric acid (H_2SO_4) at temperatures exceeding 100°C. This process enhances the adhesion of osteoblasts and fibrin, promoting osseointegration.[213] The surface roughness of CP Ti is strongly influenced by factors such as acid temperature, etching time (ranging from 0.25 to 8.00 hours), weight loss, and the activation energy in concentrated sulfuric acid (48%).[214]

Effect of Anodization

Anodizing is an electrochemical process applied to increase the thickness of the oxide layer on titanium implants. Increasing the anodizing voltage, decreases the refractive index due to a reduction in the density of the anodic film.[215] The thickness of the oxide layer increases with higher anodizing voltages. Anodizing significantly decreases the corrosion rate of titanium samples; however, at higher voltages, the increased porosity

of the oxide layer can lead to an elevated corrosion rate. Anodization is an electrochemical process that enhances the surface properties of titanium by forming a protective oxide layer. This method enables the creation of a porous titanium oxide film through dielectric breakdown, resulting in a crystalline structure with a specific chemical composition. The anodized surface exhibits increased roughness and homogeneity, featuring numerous nanoscale pores. These structural modifications contribute to improved corrosion resistance and reduced ion release rates, making anodized titanium suitable for various applications, including biomedical implants.[216]

2.13 Effect of Surface Alterations on Cells and Tissues

Once a dental implant is placed into the osteotomy, a cascade of molecular and cellular events begins. This process starts with the formation of a fibrin clot and the recruitment of various cells that interact with the titanium surface or its coating, ultimately leading to osseointegration. The implant surface engages with a variety of proteins and cell types, but the key objective is to attract osteoblasts—cells responsible for producing bone extracellular matrix—to achieve optimal bone-to-implant contact. Cell adhesion marks the initial phase, which is critical for subsequent osteoblast proliferation and differentiation, and is influenced by surface roughness and energy. Most dental implants feature a moderately rough surface (Ra = 1.0–2.0), which helps trap fibrin, enhances the attachment of bone-forming cells, and improves mechanical stability. Surface irregularities also influence the activity of both osteoblasts and osteoclasts, thereby affecting bone remodelling.

Bone formation on an implant is a multi-step process: it starts with the recruitment of osteoblast precursors, followed by their differentiation into active osteoblasts that secrete unmineralized extracellular matrix (osteoid), which later mineralizes. This entire sequence is regulated by numerous cellular and molecular signals, including autocrine and paracrine factors. Mesenchymal stem cells at the implant site can differentiate based on the concentration and type of signalling molecules present during the early wound healing stages, impacting both the type of cells recruited and their characteristics. Research shows that rough surfaces enhance the expression of TGF-β1, which supports greater osteoblast differentiation.

Osteoblasts can detect nanoscale surface features and respond differently depending on specific surface properties. For example, the roughness of titanium affects the behavior

of MG-63 osteoblast-like cells, including their proliferation, differentiation, and extracellular matrix synthesis. The implant surface also influences the release of cytokines and growth factors, shaping the healing environment. On smoother surfaces, the response to $1\alpha,25(OH)_2D_3$ is minimal, promoting a fibroblastic phenotype. In contrast, rougher surfaces reduce cell proliferation but enhance differentiation and responsiveness to this hormone, indicating a mature osteoblast phenotype. These cells produce more TGF-β1, promoting bone formation and reducing osteoclast activity, thus creating an environment favorable to new bone growth.

Polished titanium surfaces, on the other hand, are biologically inert and delay bone cell adhesion. On macro- and microporous surfaces, cells are guided in specific directions, which can alter their shape and behavior, potentially leading to either proliferation or programmed cell death (apoptosis). Cells on smooth surfaces tend to spread out more and appear more cuboidal with fewer cellular extensions compared to those on rough surfaces. Titanium also attracts calcium and phosphate ions, readily binds proteins, and stimulates cells to secrete hydroxyapatite and matrix proteins, helping to establish a bone-forming environment.

While rough titanium surfaces generally perform better than smooth ones, excessively high roughness levels (Ra = 52–74 μm) are beyond the sensing ability of osteoblasts and may not support optimal implant performance. The type of implant material and its surface properties influence the nature and distribution of adsorbed proteins and hydroxyapatite, which in turn affect cell adhesion, growth, and differentiation—ultimately impacting the clinical success of the implant.

2.14 Effects Of Surface Alterations on Healing / Osseointegration

Dental implants surface characteristics are vital to their biological performance. While core material properties like Young's modulus and fatigue resistance govern mechanical strength, it is the surface features that largely determine chemical and biological interactions with surrounding tissues. These interactions commence immediately upon implantation, starting with the adsorption of water, ions, and biomolecules, followed by mineral accumulation. This forms a preliminary layer that conditions subsequent cellular responses, ultimately influencing how well tissue integrates with the implant.

Research has consistently shown that implants with rough surfaces tend to integrate with bone more efficiently and rapidly than smooth, machined surfaces. This is because smooth implants are more likely to develop fibrous encapsulation—a process where a collagen-rich but poorly vascularized layer forms around the implant, hindering bone contact. Factors such as ongoing inflammation, poor blood flow, and low osteoblast activity contribute to this issue. When fibrous tissue forms instead of bone, it creates a fluid-filled space that can become a breeding ground for bacteria, leading to infection and bone degradation due to chronic inflammation. In contrast, rough surfaces often possess a thicker titanium oxide layer, which is biologically active and supports tissue bonding.

Surface roughness is classified into three scales: macro, micro, and nano. Macroroughness (millimeter to tens of microns) typically includes structural features like threads or pores larger than 10 microns. Microroughness, in the range of 1 to 10 microns, helps create a stronger mechanical bond between bone and implant.

Nano roughness increases surface energy and can directly influence cellular behaviors such as growth and specialization. Studies indicate that nanoscale patterns can modulate how cells interact with the surface.

Various techniques are employed to modify implant surfaces. Grit blasting is a common method used to roughen titanium, and it has been found to enhance osteoblast function, including gene activity and mineral formation, compared to smoother surfaces. Clinical evidence also shows that grit-blasted implants integrate better with bone.

Another widely used technique is acid etching, often in combination with grit blasting. Acid etching can improve surface texture and support osseointegration without leaving debris, although the use of strong acids may cause hydrogen embrittlement in titanium, potentially leading to microcracks that weaken the implant over time.

Plasma-sprayed hydroxyapatite coatings represent another effective strategy for enhancing surface roughness. These coatings facilitate quicker bone ingrowth into the implant's porous structure, potentially shortening healing time and improving overall success rates.

2.15 Future Trends in Implant Surface Modifications

The critical role of dental implants in restoring aesthetics, structure, and function has driven ongoing innovation in implant materials, surface coatings, and design enhancements. This underscores the necessity for advanced and dependable evaluation techniques before the adoption of any new biomaterial. One significant area of focus is the development of analytical tools for characterizing polymer sequences—an ongoing challenge in polymer science as researchers aim to optimize material properties for medical applications.[232] In parallel, OMICS technologies are increasingly utilized to identify molecules and signalling pathways involved in the formation of bone and osseointegration, offering potential for personalized approaches in implant therapy.[233] Additionally, advances in mass spectrometry hold promise by enabling detailed analysis of molecular composition, thereby helping to decipher the complex processes involved in bone regeneration.[234]

Bisphosphonates, known for their antiresorptive action and common use in treating osteoporosis, have shown potential when applied to implant surfaces. Coating implants with bisphosphonates has been found to enhance osseointegration by increasing bone density in the surrounding area, with the drug's effects remaining localized. This approach could be particularly advantageous for patients with systemic health conditions.[115]

The use of biomarkers to evaluate bone healing and regeneration around various implant materials is another evolving area. The success of titanium implants, for example, may be gauged by monitoring biomarkers related to osseointegration, including osteocalcin (OCN) and collagen type I (COL-1).[235] Similarly, fluoride-coated implants have been linked to enhanced bone formation, as indicated by the presence of markers like OCN, RUNX-2, and COL-1. [236] Simvastatin-coated implants also show promise, promoting both angiogenesis and osseointegration through elevated expression of markers such as VEGF and alkaline phosphatase (ALP).[237] Clinical applications now include the use of ELISA tests to assess biomarkers associated with inflammation, healing, and osseointegration, allowing researchers to evaluate the biological impact and therapeutic potential of new biomaterials and interventions.[238]

A novel area of research into the causes of peri-implantitis has identified the degradation of implant surfaces and subsequent release of particles as a contributing factor.[239] Insights from this research may lead to the development of new coatings designed not only to enhance osseointegration but also to support other critical biological interactions. Genetically engineered implant surfaces are also being explored. These genes are involved in cell differentiation and function.[240]

Application of recombinant osteogenic proteins (OP-1), BMP-2, BMP-7, and platelet-derived growth factor to implant surfaces is also being widely explored with promising results regarding osteoconductive properties.[241]

3. Summary & Conclusion

Dental implant surface modifications present an important field of innovation in implantology, with the ultimate goal of achieving effective osseointegration and successful implant treatment outcome. Over the years, several surface modification strategies have been adopted in order to improve the biological, chemical, and mechanical properties of implants. These include macro topographical alterations in the implant geometry including crest module and thread patterns; modifications that influence the microtopography, such as grit blasting, acid etching, laser preening etc. Each technique impacts the implant surface properties in a unique way, influencing cell adhesion, proliferation, and differentiation.

The biological response to these altered surfaces has been described, revealing that specific surface characteristics can significantly influence protein adsorption, osteoblast activity, and bone healing. Moreover, the development of bioactive surfaces, particularly those incorporating calcium phosphate, hydroxyapatite, and biomimetic coatings, has contributed to an osteoconductive environment. Alongside, the integration of antimicrobial elements, such as silver ions, chitosan, or antibiotic-loaded coatings, has opened new possibilities in combating early-stage infection.

It is worthy to note that rougher surfaces have more surface area and microscopic crevices, providing more niches for bacterial adhesion in the form of biofilm. This may cause inflammation and gives rise to peri-implant disease, which comprises of peri-implant mucositis, a reversible inflammation of the soft tissues around an implant, without bone loss; and Peri-Implantitis, which is a destructive inflammatory process affecting soft tissues along with bone loss. Hence it may be understood that, implants with rough surfaces may show more aggressive and rapid progression of peri-implantitis once infection is established due to difficulty in maintaining oral hygiene. Thus, while surface roughness can enhance osseointegration, it also presents challenges in maintaining oral hygiene and may influence the progression of peri-implant diseases. Therefore, careful consideration of implant surface characteristics is essential in clinical practice. Some modern implants using a dual-surface design i.e., rough at the bone-contacting part and smooth at the collar/neck to reduce plaque accumulation near the gingival margin may be explored.

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