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Surface Roughness of Implants: A Review

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For centuries, clinicians have been attempting to replace missing teeth with suitable synthetic materials. Dental implants are fixtures that serve as replacements for the root of the missing natural tooth and becoming popular in the current day dental practice. Success or failure of the dental implant treatment is mainly based on the principles of creating and maintaining an interface between the implant and surrounding bone. This can be achieved by a phenomenon called *osseointegration*, which is the direct and stable anchorage of an implant due to the formation of bony tissue around the implant. A number of systemic and local factors influence the production of an osseointegrated interface and therefore the stability of the implant. However, surface characteristics of the implant materials in general and surface roughness in particular have received a great deal attention in the recent years to help achieve favourable interaction between the implant and biological tissues. Present article is a review of surface roughness characteristics and its effect on the osseointegration of dental implant materials.

Introdcution

Aim of the any dental treatment is to replace the damaged or lost part of the tooth structure with a suitable material so as to restore the function and aesthetics' of the tooth. Over the years, several materials have been employed for this purpose. Among the various categories of materials used, implants have become more popular in the recent past. An implant is defined as a biomaterial which is inserted either partially or completely, into the body for therapeutic, diagnostic, or prosthetic purposes [1,2]. In other words, they serve as replacements for the root of the missing natural tooth. After the insertion, they integrate with the bone over time and serve as an anchor for the dental prosthesis [3].

History

Dental implants were used as early as by Greeks, Etruscans and Egyptians. Archaeological findings showed that materials used to replace missing human teeth include ox teeth, sea shells, coral, ivory (elephant tusk), stones, wood, human teeth from corpses, jade, and metals (gold or silver) [4–6].

The first documented placement of implant was from Albucasis de Condue, who used ox bone to replace missing teeth and the same was followed for centuries for a series of tooth transplants using either human or animal teeth [5]. In 18th century, Pierre Fauchard and John Hunter documented tooth transplantation with conditions for its success and claimed that the success was greater with anterior teeth or premolar replacement and in young people with healthy tooth sockets. Failures of these transplants are attributed to the incompatibility of the type of tooth used or lack of fitting of the transplant into the socket [5]. The increased failure rate of transplants stimulated interest in using artificial materials as tooth roots. In 1809, Maggiolo designed and implanted gold tooth roots into fresh extraction sockets. A crown was attached to the root only after they were allowed to heal [3,5]. Although implants failed after a period of time, this made the researchers to experiment and use various metals and alloplastic materials to replace the missing teeth. In the late 1800's various researchers implanted different metals including platinum posts coated with lead, gold or iridium tubes, silver capsules, etc [5]. The success rates of all these implants were limited. As scientific experimentation on tissue biocompatibility and bone material interaction continued, Vitallium (cobalt–chromium–molybdenum alloy) was proposed by Venable, Strock and Beach and became the first long–term successful implant material in 1930's [3,5]. Vitallium was considered to be inert, compatible with living tissues, and resistance corrosion in the body fluids [5]. However, the success rate and service life for these implants were greatly variable and unpredictable [3]. Many other materials and designs followed, including the use of stainless steel, porcelain, high-density aluminium oxide (alumina), sapphire (alpha alumina), hydroxyl apatite, bioactive glass (bioglass), and carbon [5].

In 1952, a Swedish orthopaedic surgeon, Per-Ingvar Bränemark, developed a threaded implant design made of titanium that increased the popularity of implants to a new level. Bränemark studied aspects of implant design, including biological, mechanical, physiological and functional phenomena relative to the success of the endosteal implants [3,5]. Since then the ongoing research with endosseous dental implants has revolutionized dental care. Present day dental implant treatment is much advanced than it was earlier and much of it clinical success is related to the improvements in surgical management, combined with greater understanding of biological responses and engineering of dental implants [7]. The outcome and the rate of success of dental implants is mainly based on the principles of creating and maintaining an interface between the implant and surrounding bone [8] which is capable of load transmission, associated with healthy adjacent tissues. This outcome proved elusive until the discovery of the phenomenon of “***osseointegration***” [6].

Osseointegration

Goal of the research and development in the implant materials is to achieve physical and biological compatibility with alveolar bone. Ideally, bone should integrate with the implant material rather than responding to the material as a foreign substance by encapsulating it with fibrous tissue [3]. Bränemark observed the fusion of bone with titanium chambers when he had placed them into the femurs of rabbits. The word “***osseointegration***” is coined by him for such a phenomena [3,6,9], which is defined as the apparent direct attachment or connection of osseous tissue to an inert, alloplastic material without intervening connective tissue [10-13]. During osseointegration, osteoblasts and mineralized matrix contacts the implant surface even when loads are applied. The first practical application of osseointegration was the implantation of new titanium roots in an edentulous patient in 1965 and the first ground breaking study was published 16 years later by Adell et al. in the *International Journal of Oral Surgery*. This is considered to be the beginning of the birth of modern implantology and its acceptance worldwide [9]. Since then titanium and its alloys have been extensively used as implant materials for over 40 years due to their excellent biocompatibility, mechanical properties and high corrosion resistance. Biocompatibility of implants is very important with the physiological environment in which they are placed. Osseointegration provides a stable bone-implant connection that can support a dental prosthesis and transfer applied loads without concentrating stresses at the interface between the bone and the implant. Osseointegration takes place when the bone is viable and space between the bone and implant must be less than 10 nm without any fibrous tissue [3]. Both the aspects of osseointegration, maintenance of present bone (remodelling) and new bone formation (modelling), determine the fate of implant healing [12] and implant stability. Before introduction of the Bränemark protocol, dental implants were commonly loaded at placement because immediate bone stimulation was considered to avoid crestal bone loss. Bränemark et al. in 1969 showed that direct bone apposition at the implant surface was possible and lasts under loading at the condition that implants were left to heal in a submerged way [14].

For many years, researchers in the field of dental implantology have investigated several materials such as metals, polymers, ceramics and composites as potential candidates for implants**.** Amongst these, dental implants made from commercially pure titanium (CPTi) have shown excellent success rates in long-term studies. Biocompatibility of CPTi has been demonstrated in *in vitro* studies as well as *in vivo* studies in animals and humans. High clinical success rates with these implants are not only due to the nature of CPTi but also due to the proper selection of patients and adherence to strict protocols during the operative and postoperative follow-ups. In addition, a number of systemic and local factors have been identified as being associated with the production of an osseointegrated interface. These include type of material being implanted, surface composition, structure, heat generated during surgery, contamination, initial stability of implant, bone quality, and time of loading the implants [6,11,15,16]. The role of material properties for achieving a successful long-term clinical performance is related to the type of local tissue conditions and clinical needs. For the majority of long-term implanted materials, inertness of the material is usually the preferred characteristic [11]. Recently, there has been an increasing interest in the chemical modification of the surface of such metallic implants to allow attachment of bioactive biomolecules, which may enhance their osseointegration.

Surface roughness

Response of the tissues to the implant is largely controlled by the nature and texture of the surface of the implant. Compared to smooth surfaces, textured implants surfaces exhibit more surface area for integrating with bone via osseointegration process. Textured surface also allows ingrowth of the tissues [17,18]. The role of surface topography has been the interesting area of investigation in implant dentistry for several years. Several types of implant surface textures are currently available for clinical use. Some of these have the ability to enhance and direct the growth of bone and achieve osseointegration when implanted in osseous sites [19]. Endosseous dental implants are available commercially with many different surface configurations. Most implant systems of this category are based on the fact that bone tissue can adapt to surface irregularities in the 1 – 100 micron range, and that altering the surface topography of an implant can greatly improve its stability [18].

Ability of textured implant, with higher surface area, to achieve better bone-to-implant contact has motivated many researchers around the world to carry out research on the effect of different types of textures and various methods of achieving the textured surfaces. Goal of various surface textures and techniques is to enhancing bone growth towards the implant surface. A number of *in vivo* studies have demonstrated that increased surface area on the implant improves bone-to-implant contact after the implant placement [20–22].

Primary aim of the surface texturing or treating the implant surface is to enhance cellular activity and improve bone apposition [23,24]. Studies using endosseous dental implants in human clinical trials indicated that rough surfaces integrate better with the bone than those materials with relatively smooth surfaces [25]. With few animal studies, low success rates were observed when implantation was done in the posterior maxilla (with cancellous bone) as compared to denser bone elsewhere in the mouth [23]. When the bone volume and the quality are poor, an implant with greater surface roughness is indicated. Based on the scale of the features, the surface roughness of implants can be divided into macro, micro, and nano-sized topologies [26,27]. Several methods have been employed to alter the surface topography and surface chemistry of the of the implant materials [7].

Macro-topographic profiles of dental implants have a surface roughness in the range of millimetres to microns. Because the size of the topography is large (roughness more than 10 ìm), it is directly related to implant geometry, (example: threaded screw, solid body press-fit designs and/or sintered bead technologies) [7,26,27]. Screw threaded implants are designed to achieve a compressive loading of the surrounding cortical or cancellous bone. Sintering technologies are used to create mesh or sintered beads on the surface of the implant to facilitate the growth of bone. Success rate of short implants (< 10 mm in length) with sintered bead technology was found to be superior [26]. Macro-sized topographies with high rough surfaces help in initial implant stability and provide volumetric spaces for growth of bone [26,27]. However; high surface roughness may result in an increase in ionic leakage as well as peri-implantitis [27].

The microtopographic profiles of dental implants have a surface roughness in the range of 1 □ ",)"□ 10 ìm. Micro-surface roughness attempts to enhance the osteoconduction (in-migration of new bone) through changes in surface topography, and osteoinduction (new bone differentiation) along the implant surface by utilizing implant as a vehicle for local delivery of bioactive agents (adhesion matrix or growth factor such as BMP [Bone Morphogenic Protein]) [26]. Implant surfaces with microtopographies have shown greater percentage of bone-to-implant contact when compared with machined or polished titanium surfaces. Plasma etched surface also shows similar results however they are no better than surface topographies created by sand blasting or acid etching [28].

Improved bone bonding and accelerated bone formation appears to be possible with roughened surfaces modified with certain acid treatments. Sandblasted and acid-etched surfaces have shown improved bone apposition in histomorphometric analyses. These studies indicate that surface modification improves osseointegration of the implant surface with the bone and suggests a synergistic mechanism to enhance bone formation involved between the macro-topography (due to the sandblasting procedure) and micro-texture (due to acid etching) of the implant [7].

Recently methods have also been proposed to create nano features on titanium implant surfaces. Physical approach by compaction of nanoparticles (such as titanium dioxide [TiO_2]), molecular self-assembly method, chemical modification by acid/alkaline treatment or peroxidation, nanoparticle deposition (such as sol-gel and discrete crystalline deposition) [29] have been used to create nanotopographies on the implant surfaces. Nano-sized topographic profiles on the implant surface may play a role in the adsorption of proteins, adhesion of osteoblastic and thus the rate of osseointegration [30,31]. Acid etching of the grit blasted implant surface increases the surface roughness by creating a nanotopography that allows bone ingrowth [26]. One of the drawbacks of the nanotopographies is the reproducibility in the roughness values [27]. Primary goal of current strategies is to provide an enhanced osseous stability through micro and nano-surface features. These strategies can be divided into those that attempt to enhance bone ingrowth (e.g. osteoconduction), through changes in surface topography (e.g. surface roughness) and the biological means to manipulate the type of cells that grow onto the implant surface.

Surface topography of an implant can be designed by making porous and/or by coating the implant surface with other suitable materials to increase bone-implant contact since the anatomic surface of bone cannot be controlled [23]. A number of surface treatments are available to create controlled roughness on the surface of the implants. Roughness can be produced on the implant surfaces through the addition or subtraction procedures. A plasma arc is a kind of addition process, which involves the deposition of bioactive hydroxyapatite material on the surface of the implants. Polishing, machining, and acid-etching, on the other hand, are subtraction procedures [23]. These treatments may also be classified into mechanical, chemical, electrochemical, electropolishing, vacuum, thermal and laser methods [11]. In addition to creating surface topography on the implant, some of these methods also produce sterile surfaces on the implant surfaces [2].

Mechanical treatments

Mechanical methods involve treatment, shaping or removal of the material surface by means of physical forces [11]. Mechanical treatments involve either removal of surface material by cutting or abrasive action, or the surface of the implant is deformed (and/or partially removed) by particle blasting [2]. The most commonly employed mechanical techniques are machining, polishing, and blasting [11].

Machining (lathing, milling, threading) is not really a surface treatment method, but on the other hand it can be used to produce specific surface topographies and surface compositions. The properties of machined surfaces mainly depend on the work-piece speed, tool pressure and choice of lubricant. Machined implant surface is generally characterized by grooves and valleys more or less oriented along the machining direction [11] and the surface layers are plastically deformed. Depending on the machining parameters, surface roughness values (Ra □ mean arithmetic roughness) may range between 0.3 to 0.6 ìm when measured by optical or stylus profilometry [32,33]. Grinding and mechanical polishing are identical methods in that they remove some of the surface material by using a hard abrasive [11,34]. Grinding involves use of coarse particles as abrasive medium to remove the surface at a faster rate. Grinding creates relatively rough surface topographies. Grinding with an abrasive grade 60 leads to Ra values around 1μm, and with the coarsest grade the surface roughnesses of up to 5–6 μm can be achieved [34].

Polishing of the implant surface involves use of a fine abrasive material that is applied to a flexible wheel or a belt and then the implant is brought into direct contact with the abrasive surface. Polishing is always carried out in the presence of lubricant. During the initial process coarse abrasive paper (50 220 grit) is employed followed by a finer abrasive (about 600 grit) at a speed of 10 30 m/s [35]. Polishing is generally carried out using SiC, alumina or diamond to produce extremely smooth and mirror like surface with Ra values of 0.1 μm or less [2].

Grit blasting, also known as abrasive blasting, is another technique which is used to create surface topographies on the implant surfaces. In grit blasting, surface of the implant is bombarded with hard dry particle or particles suspended in a liquid at high velocity. Various types of ceramic particles such as alumina, silica, etc. of dif–ferent sizes can be used for grit blasting of titanium [2,23]. This technique is generally employed for descaling and surface roughening of commercial implants there by increasing the surface area of the implant for better osseointegration. Shot peening is a modified method of grit blasting and is used primarily for introducing compressive stresses in the material's surface. It is most commonly used for producing specific surface topographies on various biomaterials surfaces [2]. Surface topography achieved by shot peening depends greatly on the size of the particle used [11]. Alumina particles in the size range of 25–75 μm result in mean surface roughness in the range 0.5–1.5 μm, [33,35,36] where as roughness in the range of 2–6 μm are reported for surfaces blasted with particles of size between 200–600 μm [20,30]. Use of fine particle size glass particles of 150–230 ìm results in relatively smooth surface with Ra value of 1.36 ìm where as use of coarse alumina particles of 200 ”@3” 500 ìm provides a much rougher surface with Ra value of 5.09 ìm [37].

Chemical treatments

A variey of chemical treatments such as solvet cleaning, wet chemical etching, and passivation treatments have been employed for modifying the implant surfaces.

Solvent cleaning is mainly aimed at cleaning the surface of the implant from oils, greases and fatty surface con–taminants remaining after manufacturing process by using organic solvents (aliphatic hydrocarbons, alcohols, ketones or chlorinated hydrocarbons), surface active detergents and alkaline cleaning solutions. For affective cleaning the process may be carried out at elevated temperatures with or without the use of ultrasonication [2]. This process does not have any affect on the surface of the implant. Selection of a solvent is based on the type of material to be cleaned and type of contamination to be removed from the material.

Wet chemical etching dissolves the native surface layer of the implant material including the oxide layer and parts of the underlying metal. Chemical etching is also used to improve the surface roughening as well as for producing an aes–thetically favorable surface finish. Because the titanium dioxide on the surface of the implant is a stable one, choice of etchants is limited to few acids and alkaline solutions.

Acid etching or pickling is used for removing oxide layer to obtain clean and uniform surface finish. An aqueous mixture of 10–30 volume % of nitric acid; (69 mass%) and 1–3 volume% of hydrofluoric acid (60 mass%) [2,23,30] is the most commonly used ecthing solution. Relative proportion of nitric acid to hydrofluoric acid is critical as it minimizes the formation of free hydrogen on reaction with titanium [23]. Formation of free hydrogen on the surface of the implant embrittles the implants. Mixture of 100 ml hydroclhoric acid (18 mass%) and l00 ml sulfuric acid (48 mass%) [23,32,38] can also be used as an alternative etchant to produce a significant surface roughness with micropits of 1 10 ìm and large valleys of 20 30 ìm [39]. Degree of pickling/etching is dependent on the acid concentration, temperature, and treatment time (typically in the range 1 60 min). Surfaces which have been blasted prior to acid etching will generally show irregular surface topography [31,32]. Sur–face roughness in the range from 0.1 μm to several microns have been reported with this treatment [30,32].

Recently, a new surface treatment has been developed and employed on Avantblast® (Impladent, Sentmenat, Spain). It improves osteoblast response with the advantages of an increased thickness and crystallinity of the titanium oxide layer. Surface roughness of about 1 ìm is attained with homogenization of surface stresses and chemical etching of the surface with an aqueous solution of hydrofluoric and sulphuric acids, while the increase in thickness and crystallinity of the oxide layer is due to a thermal treatment [40].

Alkaline etching is a simple technique to modify the titanium surfaces. Treatment of titanium in 4–5 M sodium hydroxide at 600 °C for 24 hours has been shown to pro–duce sodium titanate gel of 1 ìm thick, with an irregular topography with high degree of open porosity. Composition and structure of this layer can be furhter modified by proper heat treatment. Alternatively, boiling alkali solution (0.2 M sodium hydroxide, 1400 °C for 5 h) can be used to produce a high density of nanoscale pits on the titanium. When the alkali treatment is preceded by etching in hydrochloric acid/sulfuric acid, porosity of the final surface is found to increase [38].

A variety of other wet chemical surface modification methods have been applied to titanium and tita–nium alloy surfaces such as deposition of apatite coatings or organic and biological molecular films, treatment with different types of ions [2], exposure to UV/ozone [41], or immersion in hydrogen peroxide solutions [42–44]. However, these procedures may not show significant changes in the surface topography.

Passivation treatments are used for obtaining a uniformly oxidized surface to improve corrosion resistance. It is often the last step in the surface preparation of the implants. Immersion of the titanium for a minimum of 30 minutes in 20–40 vol% solution of nitric acid at room temperature is the most commonly employed method. After the passivation, surface of the implant should be neutralized, thoroughly rinsed and dried. Nitric acid passivation has no major influence on the overall surface topography of titanium surfaces [30].

In addition to nitric acid passivation, heating in air at 400–600 °C or ageing in boil–ing deionized water for several hours can be used as an alternative passivation treatments (heat treatment) for Ti–6A1–4V alloys [4547]. These treatments do not show any major changes in the overall surface topography, as compared to nitric acid passivation.

Electrochemical treatments

Electropolishing and anodic oxidation, also known as anodizing, are the most commonly used methods for titanium surface modification. They are based on different chemical reac–tions occurring at an electrically energized surface (electrode) placed in an electro–lyte. The specimen to be treated is made the anode and by controlling the variables such as choice of electrolyte and other processing parameters such as electrode potential, temperature, current etc., to obtain different effects on the sample (anode) surface [48].

In electro–polishing technique, a controlled dissolution of the surface takes place under the influence of electrochemical reaction. Choice of the electrolyte is generally a mixture of an acid and alcohols (60 ml perchloric acid and 350 ml n––butanol, and 540 ml methanol held at 25 °C or lower) for titanium. Rate of removal of the surface is dependent on many variable process parameters and is generally in the range of 1–10 µm per minute for titanium. Surface of the electro–polished titanium appears to be very smooth except for occasional pits that are preferentially located at grain boundaries. Although most of the surfaces have shown smooth surface, few materials have shown rough surface due to the differences in removal rate between different phases present in titanium alloy. In atomic force microscopy reveals that the surfaces are granular in appearance with granule size of few nanometers [49,50]. Typical surface roughness value (Ra) of electro–polished titanium is < 10 nm.

Anodization of titanium surfaces at high voltages causes crystallization of surface oxide and there by produces desired roughness and porosity. In anodic oxidation electrode reactions in combination with electrical–field driven by metal and oxygen ion diffusion lead to the formation of an oxide film at the anode surface. Type of oxide formed during anodization and its properties are influenced by anode potential, electrolyte composition, temperature and current used during the process. Diluted acids such as sulfuric acid, phosphoric acid, acetic acid etc, are commonly used as electrolytes for anodization of titanium [48].

When anodizing process is carried out below 100 V in sulfuric acid, phosphoric acid, or acetic acid, it produces microporous surfaces [49,51]. This process is used to produce Nanoporous (approximately 10 100 nm) surfaces by using chromic acid with or without HF at 10 ”8/” 40 V. Rough and microporous (approximately 0.1 ”8/” 1 ìm) surfaces can also be obtained in spark anodizing in sulfuric acid, phosphoric acid or mixtures of these at above 100 V or spark anodization in Calcium and Phosphorus based electrolytes [2].

Vacuum treatments

Vacuum treatment offers superior control on the processing conditions, especially with respect to cleanliness. Glow–discharge treatment, also known as cold plasma treatment, is based on the action of a low–pressure electrical discharge on the surface of the implant. Two different types of plasma treatments are available such as plasma deposition method and plasma surface modification. In plasma deposition, glow discharge is used to deposit the coating material from a separate solid target (sputter deposition) and/or by reactions in the gas phase (reactive sputtering or plasma polymerization). Plasma surface modification, on the other hand, is based on the exposure of sample surface to a glow dis–charge in order to obtain a specific modification of surface properties. Surface modification of inorganic materials by cold plasma is achieved by bombardment of energetic ions, leading to removal of atoms and molecules from the surface (sputtering), and reactions between gas or plasma phase and surface atoms. Plasma treatment increases the surface energy of the implant and there by improves the wetting characteristics as compared to conventional implant surfaces cleaned by using solvents or autoclaving [52,53].

In Ion implantation method, surface of the implant is bombarded with high energy ions (approximately 100 KcV to 1 McV range). Ions will penetrate the surface of implant to typical depths of approximately 0.1–1 µm [17]. Ion implantation is controlled by varying the concentration of ions and their energy [2]. Ion implantation is most commonly used on those surfaces of implants which are subjected to high wear conditions such as orthopaedic devices to increase surface hardness and reduce the generation of wear debris. This process is also used on some of the dental implants to increase the corrosion resistance by forming Ti–N surface [17].

Further, this technique is also used to produce antimicrobial surfaces on the implants. Plasma–based ion implantation (PBII) and plasma–based ion implantation and deposition (PBII–D) are the two methods used for this purpose. Ions like F and Ag with antibacterial property can be implanted and deposited on the surface of stainless steel implants with no toxic effect [54].

Thermal treatments

Commercially pure titanium was thermally annealed up to 1000°C to form oxide layer composed of anatase and rutile structures of TiO_2 on the surface which is crack–free and uniformly rough. The average roughness of the oxidized surface observed when the titanium is annealed at 600 °C and 650 °C for 48 hours was 0.90 and 1.30 ìm, respectively where as the average roughness of untreated sample was 0.08 ìm. Thermal treatment at 600 °C and 650 °C for 48 hours is considered appropriate for implanted materials [55].

Laser treatments

Implant surface roughening using the previously discussed methods would cause surface contamination. Laser techniques have recently been developed as an alternative to these techniques. Laser enables implant surface treatment without direct contact and provides better control on the micro–topography of the implant. Laser treatments are clean and easy method to perform. The average surface roughness of the laser treated acid–etched implant was 2.28 ìm [56]. Clinical studies have indicated more bone formation around the laser treated implants [57,58]. This observation can be due to the formation of TiN on the surface that improves biocompatibility [58].

Conclusion

Various methods of surface modification or rough surface preparation in titanium and its alloys for implants were discussed with an emphasis on the methods based on the mechanical, thermal, chemical, electrochemical and laser methods. Although mechanical methods can be used to produce roughness on the titanium implant surfaces, the properties of the surface oxide layer are more difficult to control. Chemical treatments of implant surface are mainly employed to improve the oxide layer thickness required for the passivation of the metal. Several alternative methods have been discussed which are used to produce surface films on titanium implants with varying morphology, thickness, microstructure, and chemical composition. In thermal treatments, surface roughness and amount of oxide layer formation are temperature and time dependent. Laser surface treatment, on the other hand, can used to produce desired surface roughness without any contamination of the implant surfaces. Methods discussed are well established and are the methods that are widely used by the manufacturers of current day dental implants. Although these methods have been successfully developed and employed to produce dental implants with varying surface topographies, the effect of the surface topographies on the long term biological compatibility and osseointegration has not been established very well. However, research in this area is very much active and several new technologies and methods will be introduced in near future to produce various surface topographies on the implants surfaces.

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