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ORIGINAL ARTICLE

Biomechanical Factors That Influence the Bone-Implant-Interface

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Introduction

Osseointegration or osteointegration refers to a direct bone-to-metal interface without interposition of non-bone tissue. This concept has been described by Branemark, as consisting of a highly-differentiated tissue making "a direct structural and functional connection between ordered, living bone and the surface of a load-carrying implant [1,2]. Through his initial observations on osseointegration, Branemark showed that titanium implants could become permanently incorporated within bone that is, the living bone could become so fused with the titanium oxide layer of the implant that the two could not be separated without fracture.

Bone healing around implants involves a cascade of cellular and extracellular biological events that take place at the bone-implant interface until the implant surface appears finally covered with a newly formed bone [3]. These biological events include the activation of osteogenic processes similar to those of the bone healing process, at least in terms of initial host response [4-6]. This cascade of biological events is regulated by growth and differentiation factors released by the activated blood cells at the bone-implant interface [7].

Factors affecting osseointegration

Factors enhancing osseointegration include biological and biomechanical factors, biological factors such as, the status of the host bone bed and its intrinsic healing potential, the biomechanical factors which will be discussed in this research divided into three main factors.

Host related factors:

a) Bone density and its rule in withstanding the stresses, as the differences in bone density make differences in bone to implant contact.

b) Available remaining bone after extraction, which has a direct influence in choosing the width and the length of the implant, affecting the surface area and the bone to implant contact.

c) Parafunctional habits, which increase time, magnitude, direction, and distribution of the forces affecting bone to implant contact.

Forces and loading conditions applied on the implant

Implant-related factors:

a) Implant macro design (implant body, length and diameter, threads shape, pitch, lead, depth and width, and crest module), implant design is mainly responsible for 1- increase the surface area of the implant, 2- decrease the stress in addition to 3-distributing the forces on the bone and convert the stresses into favorable compressive stresses.

b) Chemical composition and biomaterial of the implant and its relation to biocompatibility, enhancing healing, modulus of elasticity.

c) Implant surface treatment and coatings (surface topography), responsible for increase the surface area of the BIC, decrease the stresses, enhance adhesion qualities to the bone-implant interface at initial healing.

Other factors:

a) Implant tilting

b) Prosthetic passive fit

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c) Cantilever, crown high and occlusal table

d) Loading time

Host Related Factors

Bone density

Mish density bone classification: Four bone densities found in the edentulous regions of the maxilla and mandible. D1 bone is primarily dense cortical bone; D2 bone has dense to thick porous cortical bone on the crest and coarse trabecular bone underneath; D3 bone has a thinner porous cortical crest and fine trabecular bone within; and D4 bone has almost no crestal cortical bone. The fine trabecular bone composes almost all of the total volume of bone.

Four facts form the basis for treatment plan modification in function of the bone quality: 1) Each bone density has a different strength; 2) Bone density affects the elastic modulus; 3) Bone density differences result in different amounts of bone-implant contact percent; and 4) Bone density differences result with a different stress-strain distribution at the implant-bone interface. As the bone density decreases, the strength of the bone also decreases. To decrease the incidence of micro-fracture of bone, the strain to the bone should be reduced. Strain is directly related to stress. Consequently, the stress to the implant system should also be reduced as the bone density decreases. One way to reduce the biomechanical loads on implants is by prosthesis design to decrease force [8]. Bone is an organ that is able to change in relation to a number of factors, including hormones, vitamins, and mechanical influences. However, biomechanical parameters, such as duration of edentulous state, are predominant. [9,10]. Awareness of this adaptability has been reported for more than a century Wolff, in 1892, published, “Every change in the form and function of bone or of its function alone is followed by certain definite changes in the internal architecture, and equally definite alteration in its external conformation, in accordance with mathematical laws”. [11].

The modified function of bone and the definite changes in the internal and external formation of the vertebral skeleton as influenced by mechanical load were reported by Murry [12].

Cortical and trabecular bone throughout the body is constantly modified by either modeling or remodeling [13]: Modeling has independent sites of formation and resorption and results in the change of the shape or size of bone. Remodeling is a process of resorption and formation at the same site that replaces previously existing bone and primarily affects the internal turnover of bone, including that region where teeth are lost or the bone next to an endosteal implant [14,15].

These adaptive phenomena have been associated with the alteration of the mechanical stress and strain environment within the host bone [15,16]. Stress is determined by the magnitude of force divided by the functional area over which it is applied. The greater the magnitude of stress applied to the bone, the greater the strain observed in the bone.

Frost proposed a model of four histologic patterns for compact bone as it relates to mechanical adaptation to strain [17]. Acute disuse window, adapted window, mild overload zone and the pathologic overload zone.

a) The bone in the **acute disuse window** loses mineral density, and disuse atrophy occurs because modeling for new bone. The microstrain of bone for trivial loading is reported to be 0 to 50 microstrain.

b) **The adapted window** (50 to 1500 microstrain) represents an equilibrium of modeling and remodeling, and bone conditions are maintained at this level. Bone in this strain environment remains in a steady state, and this may be considered the homeostatic window of health. The histologic description of this bone is primarily lamellar or load-bearing bone. Approximately 18% of trabecular bone and 2% to 5% of cortical bone is remodeled each 25 year in the physiologic loading zone, which corresponds to the adapted window. This is the range of strain ideally desired around an endosteal implant.

c) **The mild overload zone** (1500 to 3000 macrostrain) causes a greater rate of fatigue microfracture and increase in the cellular turnover rate of bone. As a result, the bone strength and density may eventually decrease. The histologic description of bone in this range is usually woven or repair bone. This may be the state for bone when an endosteal implant is overloaded and the bone interface attempts to change the strain environment.

d) **Pathologic overload zones** are reached when microstrains greater than 3000 units [14].

The stress contours in the bone are different for each bone density: In D1 bone, highest strains are concentrated around the implant near the crest, and the stress in the region is of lesser magnitude. D2 bone, with the same load, sustains a slightly greater crestal strain, and the intensity of the stress extends farther apically along the implant body. D4 bone exhibits the greatest crestal strains, and the magnitude of the stress on the implant proceeds farthest apically along the implant body. As a result, the magnitude of a prosthetic load may remain similar and yet give one of the following three different clinical situations at the bone-implant interface, based on bone density: 1) Physiologic bone loads in the adapted window zone and no marginal bone loss; 2) Mild overload to pathologic overload bone loss and crestal bone loss; or 3) Generalized pathologic overload and implant failure.

Bone density and bone-implant contact

The initial bone density not only provides mechanical immobilization of the implant during healing, but after healing also permits distribution and transmission of stresses from the prosthesis to the implant-bone interface. The mechanical distribution of stress occurs primarily where bone is in contact with the implant. Misch noted in 1990 that the bone density influences the amount of bone in contact with the implant surface, not only at first stage surgery, but also at the second stage and early prosthetic loading [18]. The bone implant contact (BIC) percentage is significantly greater in cortical bone than in trabecular bone.

Available bone after extraction

Available bone after extraction significance to bone -implant- contact: With a greater surface area of implant-bone contact, less stress is transmitted to the bone, improving the implant prognosis. For a generic root-form implant design, each 0.25-mm increase in diameter corresponds to a surface area increase of 5% to 8%. Therefore, a cylinder root-form implant 1 mm greater in diameter will have a total surface area increase of 20% to 30%. (S = F/A), the greater diameter decreases the amount of stress at the crestal bone-implant interface. Because early bone loss and complications relate to the crestal bone regions, the width of the implant is much more critical than its height, once a minimum height has been obtained.

Maxillary and mandibular atrophy followed by tooth extraction, classification according to Carl E Mish:

a) **Division A (Abundant Bone):** Division A abundant bone forms soon after the tooth is extracted. The abundant bone volume remains for a few years, although the interseptal bone height is reduced and the original crestal width usually is reduced by more than 30% within 2 years [19]. Division A bone corresponds to abundant available bone in all dimensions

b) **Division B:** As the bone resorbs, the width of available bone first decreases at the expense of the facial cortical plate because the cortical bone is thicker on the lingual aspect of the alveolar bone, especially in the maxilla. A 25% decrease in bone width occurs the first year, and 40% decrease in bone width occurs within the first 1 to 3 years after tooth extraction [19,20]. The resulting narrower ridge is often inadequate for many 4-mm diameter root-form implants. Slight to moderate atrophy often is used to describe this clinical condition. Once the bone reaches this Division B bone volume, it may remain at this level for more than 15 years in the anterior mandible [21].

c) **Division C (Compromised Bone):** The Division C available bone is deficient in one or more dimensions (height, length, width, angulation, or crown height/bone height ratio) Therefore, the width may be less than 2.5 mm, the crown height more than 15 mm, and the bone angulation greater than 30 degrees.

d) **Edentulous Division D (Deficient Bone):** Long-term bone resorption may result in the complete loss of the alveolar process, accompanied by basal bone atrophy. Severe atrophy describes the clinical condition of the Division D ridges site.

Bone availability dimensions

a) **Available bone height:** The height of available bone is measured from the crest of the edentulous ridge to the opposing landmark. The anterior regions of the jaws have the greatest bone heights because the maxillary sinus and inferior alveolar nerve limit this dimension in the posterior regions.

The minimum height of available bone necessary for long-term survival of endosteal implants is related in part to the density of bone. The denser bone may accommodate a shorter implant (i.e., 8 mm), and the least dense, weaker bone requires a longer implant (i.e., 12 mm). Once the minimum implant height is established for each implant design and bone density, the width is more important than additional length. The height of the implant also affects its total surface are an initial stability of the implant, the overall amount of bone-implant interface, and a greater resistance to rotational torque during abutment screw tightening. The increased height of an implant in the immediate extraction site also facilitates healing with a decreased risk of movement at the interface. In addition, the crestal bone and opposing anatomical landmark often are composed of cortical bone, which is denser and stronger than trabecular bone. As a result, these may help stabilize the implant while the trabecular woven bone forms. This process encourages a direct bone-implant interface and may be of particular advantage when an immediate loading protocol of implants is used for a transitional prosthesis. However, once the implant has healed, the crestal region is the zone that receives the most stress. As a result, implant length is not an effective way to decrease crestal loads around an implant.

b) **Available bone width:** Once adequate height is available for implants, the next most significant criterion affecting long-term survival of endosteal implants is the width of available bone.

The available bone height in an edentulous site is the most important dimension for implant consideration because it affects crown height, force factors and esthetics and because bone augmentation is more predictable in width than height. Hence even when the width is inadequate for implant placement, bone grafting may be indicated to create a site ideal for the restorative requirements of implant insertion.

c) **Available bone length:** The mesio distal length of available bone in an edentulous area often is limited by adjacent teeth or implants. As a rule, the implant should be at least 1.5 mm from an adjacent tooth.

This dimension not only allows surgical error but also compensates for the width of an implant or tooth defect, which is usually less than 1.4 mm.

d) **Available bone angulation:** Bone angulation is the fourth determinant for available bone. Ideally, the bone is perpendicular to the plane of occlusion, aligned with the forces of occlusion, and is parallel to the long axis of the prosthodontic restoration. The incisal and occlusal surfaces of the teeth follow the curve of Wilson and curve of Spee. As such, the roots of the maxillary teeth are angled toward a common point about 4 inches away. The mandibular roots flare, so the anatomical crowns are more lingually inclined in the posterior regions and labially inclined in the anterior area compared with the underlying roots. The first premolar cusp tip is usually vertical to its root apex.

Para functional habits

Para functional forces on teeth or implants are characterized by repeated or sustained occlusion [22-24]. The most common cause of implant failure after successful surgical fixation or early loss of rigid fixation during the first year of implant loading is the result of parafunction. Such complications occur with greater frequency in the maxilla, because of a decrease in bone density and an increase in the moment of force [25].

In implant dentistry Nadler 21 has classified the causes of parafunction into the following six categories:

1. Local
2. Systemic
3. Psychological
4. Occupational
5. Involuntary
6. Voluntary

The parafunctional groups presented are

1. Bruxism
2. Clenching
3. Tongue thrust or size

The parafunction may be categorized as:

1. Absent
2. Mild
3. Moderate
4. Severe

Bruxism: Is the vertical or horizontal, nonfunctional grinding of teeth. The forces involved are in significant excess of normal physiologic masticatory loads. Bruxism may affect the teeth, muscles, joints, bone, implants, and prostheses. These forces may occur while the patient is awake or asleep and may generate several hours per day of increased force on the teeth. Bruxism is the most common oral habit [24]. Sleep clinic studies have evaluated nocturnal bruxism and have found that about 10% of those observed had obvious movement of the mandible with occlusal contacts [26,27]. More than half of these patients had tooth wear affecting esthetics. Only 8% of these patients were aware of their nocturnal bruxism. One quarter of the patients' spouses was aware of the condition. Muscle tenderness in the morning was observed less than 10% of the time [28]. Nocturnal bruxism is often a difficult disorder to diagnose. The maximum biting force of those with bruxism is greater than average. Just as an experienced weight lifter can lift more weight, the patient constantly exercising the muscles of mastication develops greater bite force. A male chewing paraffin wax for an hour each day for a month can increase the bite force from 118 psi to 140 psi within 1 week. A 37-year-old patient with a long history of bruxism recorded a maximum bite force of more than 990 psi (4 to 7 times normal) [29].

Clenching: Clenching is a habit that generates a constant force exerted from one occlusal surface to the other without any lateral movement. The habitual clench position does not necessarily correspond to centric occlusion. The jaw may be positioned in any direction before the static load. Hence a bruxing and clenching combination may exist. The clench position most often is in the same repeated position and rarely changes from one period to another. The forces involved are in significant excess of normal physiologic loads and are similar to bruxism in amount and duration.

Tongue thrust and size: Parafunctional tongue thrust is the unnatural force of the tongue against the teeth during swallowing [30]. A force of approximately 41 to 709 g/cm^2 on the anterior and lateral areas of the palate has been recorded during swallowing [31]. In orthodontic movement a few grams per square centimeter of constant force are sufficient to displace teeth. At least five different types of tongue thrust have been identified; anterior, intermediate, posterior, unilateral, and bilateral may be found and in most any combination. This condition can contribute to complications of implant healing and prosthesis durability. Although the force of tongue thrust is of lesser intensity than in other parafunctional forces, it is horizontal and can increase stress at the per mucosal site of the implant. Addressing this force is most critical for one-stage surgical approaches in which the implant resides in an elevated position at initial placement and the implant interface is in an early healing phase.

Forces and Loading Conditions Applied to the Implant

Forces applied to the implant may be evaluated in **type, direction, magnitude** and **duration.**

The surface area over which the forces are applied is also relevant and is inversely proportional to the stress observed within the implant system (stress = force ÷ surface area). It can be clearly seen from this basic engineering equation that, to reduce stress, the force must decrease or the surface area must increase .For a given bone volume, implant surface area should be optimized for functional loads.

Force type

Forces may be described as compressive, tensile, or shear. Compressive forces attempt to push masses toward each other. Tensile forces pull objects apart. Shear forces on implants cause sliding. Compressive forces tend to maintain the integrity of a bone-implant interface, whereas tensile and shear forces tend to distract or disrupt such an interface. Shear forces are most destructive to implants and bone compared with other load modalities. In general, compressive forces are accommodated best by the complete implant-prosthesis system. Thus, an important distinction is made between theoretical total surface area and functional surface area of an implant. Because 1) Bone is 65% weaker to shear forces and 35% weaker to tensile forces, functional surface area is defined as the area that actively serves to dissipate compressive loads to the implant-bone interface [32]. 2) Cortical bone is strongest in compression and weakest in shear [33]. 3) Cements and retention screws implant components, and bone-implant interfaces all accommodate greater compressive forces than tensile or shear.

Force direction

Bone is weaker when loaded under an angled force [34]. The greater the angle of load, the greater the stresses to the implant-bone interface. The noxious effect of angled loads to bone is further exacerbated because of the anisotropy of bone [35]. Anisotropy refers to how the character of bone's mechanical properties, including ultimate strength, depends on the direction in which the bone is loaded. A 30-degree angled load will increase the overall stress by 50% compared with a long axis load, especially around the crestal portion of the implant [36]. Therefore, under ideal conditions, the implant body long axis should be perpendicular to the curve of Wilson and curve of Spee to apply a long axis load to the implant during occlusal load in centric occlusion (where the occlusal forces are usually the greatest). As the angle of load to the implant-bone interface increases, the stresses around the implant increase. As a result, virtually all implants are designed for placement perpendicular to the occlusal plane. Additionally, axial alignment places less shear stress on the overall implant and decreases the risk of complications, as screw loosening and fatigue fractures. A force applied to a dental implant rarely is directed longitudinally along a single axis. In fact, three dominant clinical loading axes exist in implant dentistry: 1) Mesiodistal; 2) Labiolingual; and 3) Occlusoapical. The process by which three-dimensional forces are broken down into their component parts is referred to as vector resolution and may be used routinely in clinical practice for enhanced implant longevity.

Force magnitude

Normal physiology imposes constraints on the magnitude of forces that must be withstood by engineering designs in the oral environment. The magnitude of bite force varies as a function of anatomical region and state of the dentition. Average bite forces can range from 10 to 350 lb [29]. The magnitude of force is greater in the molar region (200 lb), less in the canine area (100 lb), and least in the anterior incisor region (25 to 35 lb) [37]. Many biocompatible materials are unable to with stand the type and magnitude of parafunctional loads that may be imposed on dental implants. As an example, ceramic, which has excellent biocompatibility, is very susceptible to tension and bending loads. Such loads are commonly applied to dental implants and render this material unsuitable in many implant body applications.

Typical **maximum bite force magnitudes** exhibited by adults are affected by age, sex, degree of edentulism, bite location, and especially parafunction [38,39].

Force duration

The duration of bite forces on the dentition has a wide range. Under ideal conditions, the teeth come together during swallowing and eating for only brief contacts. The total time of those brief episodes is less than 30 minutes per day [40]. Patients who exhibit bruxism, clenching, or other parafunctional habits, however, may have their teeth in contact several hours each day. This force duration essentially is able to create a fatigue load on the implant.

Stress

S = f/A, The question arises as to what are the peak stresses or maximum stresses that an implant and the surrounding interfacial tissues experience. Peak stresses occur when the stress element is positioned in a orientation (or geometric configuration) in which all shear stress components are zero. When an element is in this configuration, the normal stresses are given a name, principal stresses, and are indicated as s1, s2, and s3. By convention, maximum principal (s1) stresses represent the most positive stresses (typically peak tensile stresses) in an implant or tissue region and minimum principal. (s3) stresses, the most negative stresses (typically peak compressive stresses). Sigma 2 (s2) represents a value intermediate between s1 and s3. Determination of these peak normal stresses in a dental implant system and tissues may give valuable insights regarding sites of potential implant fracture and bone atrophy.

Deformation and strain

A load applied to a dental implant may induce deformation of the implant and surrounding tissues. Biological tissues may be able to interpret deformation or a manifestation thereof and respond with the initiation of remodeling activity. The deformation and stiffness characteristics of the materials used in implant dentistry, particularly the implant materials, may influence interfacial tissues.

Force delivery and failure mechanisms

The way forces are applied to implant restorations within the oral environment dictates the likelihood of system failure. The duration of a force may affect the ultimate outcome of an implant system.

Relatively low-magnitude forces, applied repetitively over a long time, may result in fatigue failure of an implant or prosthesis. Stress concentrations and, ultimately, failure may develop if insufficient cross-sectional area is present to dissipate high-magnitude forces adequately.

Implant Related Factors

Implant macro design

Thread shape and stress distribution: Most dental implants can be found in various thread shapes developed for effective inserting and force transmission. Threaded implants are inserted into the osteotomy site by creating linear motion through rotation. Thread shapes available for screw-retained implants include square shape, V-shape, buttress and reverse buttress threads, which are defined by the thread thickness and face angle [41]. Once an implant is inserted, bone undergoes constant remodeling against external stress, called bone homeostasis. When an implant receives optimal functional load, the surrounding bone experiences remodeling and produces woven bone. However, under extreme adverse stresses, microfractures occur in the alveolar bone inducing "osteoclast genesis" [42]. Since bone formation is not fast enough to fill in the damage, the defect becomes worse, resulting in severe bone loss and ultimately implant failure [43,44]. However, the optimal stress distribution is difficult to achieve, and too little or much stress can induce bone resorption [42,45]. Therefore, implant threads should be fabricated to increase surface contact area and favorable forces while reducing adverse stimuli. Recently, the finite element analysis (FEA) has been utilized to understand the effect of those geometric parameters on the load distribution at the surrounding regions.

Using FEA, Chang and his colleagues evaluated the pattern of micro motion within implants and surrounding bone with different thread designs (trapezoidal, buttress, square, and standard V-thread) under immediate loading of 300 N axial loads [46]. The results revealed that all micro motion was located near the interface of cortical and cancellous bone and the square thread profile had the most favorable micro motion value. Eraslan, et al. [47] performed the similar study with four different thread forms under a static axial load of 100 N. The study reported that maximum stress was concentrated at the cervical cortical regions around the first thread and the stress value was lowest in the square thread type. Likewise, other previous FEA and animal studies showed the most effective stress distribution and bone-to implant contact area (BIC) in the square thread shape [48,49].

Thread pitch and lead: Thread design should maximize implant surface area and create a better spreading of stress and primary stability [50]. Like thread shape, **pitch** is another important geometric factor that determines the bone-to-implant contact and the biomechanical load distribution. Thread pitch is defined as the distance between two neighboring threads, measured on the same side of the axis [51], It also refers to the number of threads per unit length [51]. Therefore, when implants have the same length, smaller pitch indicates more threads, leading to greater surface area. Another geometric parameter related to thread pitch is lead. **Lead** is the distance within the same thread before and after one complete rotation in the axial direction. That is, for single, double, and triple-threaded implants, lead increases by one, two, and three times the pitch respectively. Since lead indicates the distance that an implant would move after one turn, it plays an important role on determining the speed of implant insertion. Hence, thread pitch is clinically significant due to its effect on surface area and insertion speed [52]. Multiple studies demonstrated that implants with smaller pitch showed the greater surface area and better stress distribution particularly in low-density bone, thread pitch plays a more critical role in enhancing the primary stability in low-density bone than in high-density bone.

Thread depth and width: In addition to thread shape and pitch, its depth and width are important design parameters that affect the stress distribution around endosteal implants. According to Misch, **thread depth** is the distance from the outermost tip to the innermost body of the thread [53]. The same author defines **thread width** as the distance between the superior most and inferior-most tip of a single thread measured axially In other words, **thread depth** can also be calculated by difference between the major and minor diameter of the thread [53]. Like the previously discussed geometric variables, thread depth and width clinically influence implant insertion and surface area. The shallower **the thread depth**, the easier the implantation procedure, especially in the high-density bone [53]. It may be able to eliminate the need for tapping during the surgery. On the contrary, deep threads increase the functional surface area at the bone-implant interface, which can improve primary stability in the low-density bone or the region with high occlusal load.

Various implant systems are available using progressive threads; for example, Ankylos (Dentsply Friadent, Mannheim, Germany) [54]. In this thread form, thread depth gradually decreases from the apical end to the coronal neck of the implant. It is allegedly claimed that it may transfer the stress away from the crestal cortical region, preventing possible bone resorption [54].

Implant size (length and width): An increased **implant length** is usually not significant at the crestal bone interface, but it is a benefit for initial stability and the overall amount of bone-implant interface. The increased length also provides resistance to torque or shear forces when abutments are screwed into place. However, the increased length does little to decrease the stress that occurs at the transosteal region around the implant at the crest of the ridge during occlusal loading [55,56]. Hence increasing implant size is not an effective method to decrease stress from force factors. The surface area of each implant is related directly to the **width** of the implant. Wider root form implants have a greater area of bone contact than narrow implants (of similar design) resulting from their increased circumferential bone contact areas. Each 0.25 mm increase in implant diameter may increase the overall surface area 5% to 10% in a cylinder implant body.

Although past theories suggested that implant height increase was more important than width increase, the given occlusal load to the implant causes the most stress at the crest of the ridge, where initial bone loss occurs. The crest of the bone is where the forces are applied to the abutment screws and is the greatest stress to the entire system. As a result, width is more important than height (once a minimum height has been obtained for initial fixation and resistance to torque) [57,58]. Bone augmentation in width may be indicated to increase implant diameter by 1 mm when force factors are greater than ideal.

In addition, an increase in implant diameter has been suggested to be more effective than implant staggering to reduce stress [59]. An interesting note is that the natural teeth are narrower in the anterior regions of the mouth, where the amount of force generated is less. The natural teeth increase in diameter in the premolar region and again in the molar region as the amount of force increases, with a total 300% surface area increase from the lower anterior teeth to the maxillary molars. The length of natural teeth roots does not increase from anterior to posterior regions of the arch but their cross section does. The greater diameter not only decreases stress but also decreases the likelihood of implant fracture.

Crest module considerations: The crest module of an implant body is the transosteal region, the crest module of the implant has a surgical influence, a biological width influence, a loading profile consideration (characterized as a region of highly concentrated mechanical stress).Therefore, and this area of the implant body is a determinant for the overall implant body design. During the surgical phase the crest module design primarily benefits the crestal implant interface. The crest module of an implant should be slightly larger than the outer thread diameter of the implant body. In this way, the crest module seals completely the osteotomy, providing a barrier and deterrent for the ingress of bacteria or fibrous tissue during initial healing [60].

The larger crest module diameter also increases surface area, which can further decrease stress at the crestal region. The increase in crest module diameter increases the platform of the abutment connection with a stress reduction to the abutment screw during lateral loading. In fact, the platform dimension is more critical to reduce the stress applied to the abutment screw than is the height (or depth) of the anti-rotational hex of the abutment to implant body connection [44].

The concept of designing an implant crest module with a smooth collar is for a reduction of plaque accumulation and improved hygiene. However, the crest module is initially placed below the bone in most two-piece implant designs. Therefore the need for daily hygiene and plaque control is not relevant, unless crestal bone loss occurs .There are at least six causes of marginal bone loss at the crestal bone region of implants, including the formation of a "biological width" and occlusal overload after the implant is in function [61,62]. The biological width of an implant is related to marginal bone loss before, or after, occlusal loading bone loss occurred to the smooth region 1.5 mm below the crestal region. No bone loss occurred when the implant crest module was rough and placed at the level of the crestal bone. Therefore, the crest module of the implant should not have an extended smooth area placed below the bone. The microgap between the implant crest module and abutment may also be reduced when polished surfaces are approximated.

The crest module is related to occlusal loading [62-65]. Most of the occlusal stress occurs at the crestal region of an implant design [62,65]. A smooth, parallel-sided crest module will increase the risk of bone loss after loading. As previously discussed, smooth metal promotes shear stresses in the adjacent bone interface [61]. Because bone is 65% weaker to shear loads, attempts to limit shear are prudent. In addition, smooth metal does not encourage bone cell contact before loading [66]. A roughened surface and a crest module designed to reduce shear loading. Roughened surface (but a shear load design crest module) maintains the bone through the biological width cycle, but may lose the marginal bone during the occlusal load conditions, a rough crest module may not be sufficient to stop crestal bone loss once the implant is loaded. It has been a common clinical observation that bone is often lost to the first thread after loading, regardless of the manufacturer type or design the bone loss often stops at the first thread because the first thread changes the shear load created by the crest module to a component of compressive loading [67,68].

Biomaterial and chemical composition of the implant

Appropriate selection of the implant biomaterial is a key factor for long term success of implants. The biologic environment does not accept completely any material so to optimize biologic performance; implants should be selected to reduce the negative biologic response while maintaining adequate function, the criteria of choosing a biomaterial for dental implant is divided into

1. Bulk properties
2. Surface properties
3. Biocompatibility

Bulk properties [69,70]

Modulus of elasticity: Implant material with modulus of elasticity comparable to bone (18 GPa) must be selected to ensure more uniform distribution of stress at implant and to minimize the relative movement at implant bone interface.

a) **Tensile, compressive and shear strength:** An implant material should have high tensile and compressive strength to prevent fractures and improve functional stability. Improved stress transfer from the implant to bone is reported interfacial shear strength is increased, and lower stresses in the implant.

b) **Yield strength, fatigue strength:** An implant material should have high yield strength and fatigue strength to prevent brittle fracture under cyclic loading.

c) **Ductility:** According to ADA a minimum ductility of 8% is required for dental implant. Ductility in implant is necessary for contouring and shaping of an implant.

d) **Hardness and Toughness:** Increase in hardness decreases the incidence of wear of implant material and increase in toughness prevents fracture of the implants.

Surface properties:

a) **Surface tension and surface energy**: It determines the wett ability of implant by wetting fluid (blood) and cleanliness of implant surface. Osteoblasts show improved adhesion on implant surface. Surface energy also affects adsorption of proteins [71].

b) **Surface roughness**: Alterations in the surface roughness of implants influence the response of cells and tissue by increasing the surface area of the implant adjacent to bone and thereby improving cell attachment to the bone. Implant surfaces have been classified on different criteria, such as roughness, texture and orientation of irregularities [72,73].

Biocompatibility: This is property of implant material to show favorable response in given biological environment in a particular function. It depends on the corrosion resistance and cytotoxicity of corrosion products.

a) **Corrosion and corrosion resistance** [74]**:** It is the loss of metallic ions from metal surface to the surrounding environment. Following types of corrosion are seen.

b) **Crevice corrosion**: It occurs in narrow region like implant screw-bone interface. When metallic ions dissolve, they can create a positively charged local environment in the crevice, which may provide opportunities for crevice corrosion.

c) **Pitting corrosion**: Pitting corrosion occurs in an implant with a small surface pit. In this the metal ions dissolve and combine with chloride ions. Pitting corrosion leads to roughening of the surface by formation of pits.

d) **Galvanic corrosion**: This occurs because of difference in the electrical gradients. Nickel and chrome ions from artificial prosthesis may pass to peri-implant tissues due to leakage of saliva between implant and superstructure. This may result in bone reabsorption and also affect the stability of the implant and eventually cause failure.

e) **Electrochemical corrosion**: In this anodic oxidation and cathodic reduction takes place resulting in metal deterioration as well as charge transfer via electrons. This type of corrosion can be prevented by presence of passive oxide layer on metal surface.

Clinical significance of corrosion: Implant bio-material should be corrosion resistant. Corrosion can result in roughening of the surface, weakening of the restoration, release of elements from the metal or alloy, toxic reactions. Adjacent tissues may be discolored and allergic reactions in patients may result due to release of elements.

Biomaterials:

A) Metals and metal alloys: Metals have biomechanical properties which made them suitable as an implant material. Titanium (Ti) and its alloys (mainly Ti-6Al-4V) have become the metals of choice for dental implants. However, prosthetic components of the implants are still made from gold alloys, stainless steel, and cobalt-chromium and nickel-chromium alloys [75].

a) **Titanium:** Titanium has a good record of being used successfully as an implant material and this success with titanium implants is credited to its excellent biocompatibility due to the formation of stable oxide layer on its surface [76]. The commercially pure titanium (cpTi) is classified into 4 grades which differ in their oxygen content. Grade 4 is having the most (0.4%) and grade 1 the least (0.18%) oxygen content. The mechanical differences that exist between the different grades of cpTi is primarily because of the contaminants that are present in minute quantities. Iron is added for corrosion resistance and aluminum is added for increased strength and decreased density, while vanadium acts as an aluminum scavenger to prevent corrosion. Ti is a dimorphic metal i.e. below 882.5 °C it exists as α-phase and above this temperature it changes form α- phase to β phase. Because of the high passivity, controlled thickness, rapid formation, ability to repair itself instantaneously if damaged, resistance to chemical attack, catalytic activity for a number of chemical reactions, and modulus of elasticity compatible with that of bone, Ti is the material of choice for intraosseous applications [77,78].

b) **Titanium alloys Ti6Al4V:** Titanium reacts with several other elements for e.g.: silver, Al, Ar, Cu, Fe, Ur, Va and Zn to form alloys. Titanium alloys exists in three forms alpha, beta and α-β. These types originate when pure titanium is heated with elements Al, Va in certain concentrations and cooled, the alloys most commonly used for dental implants are of the alpha-beta variety. The most common contains 6% Al and 4% Va. (Ti 6 Al 4V) [75,79].

c) **Titanium-zirconium alloy (Straumann Roxolid):** Titanium zirconium alloys with 13%-17% zirconium (TiZr1317) have better mechanical attributes, such as increased elongation and the fatigue strength than pure titanium. Growth of osteoblasts that are essential for osseointegration is not prevented by Titanium and Zirconium. Straumann developed Roxolid that fulfills requirements of dental implantologists and is 50% stronger than pure titanium.

Sandblasting and acid-etching on, TiZr1317 with a monophasic a structure results in a topographically identical surface as on pure titanium implants. Because of its superior mechanical properties. Thin implants and implant components that can be subjected to high strains can be produced using TiZr1317 due to its better mechanical properties, provided that the material shows a similar good biocompatibility as pure titanium [80].

B) Ceramics: Ceramics were used for surgical implant devices because of their inert behavior and good strength and physical properties such as minimum thermal and electrical conductivity. Certain properties of ceramics like low ductility and brittleness has limited the use of ceramics [75].

a) **Zirconia:** Zirconia was used for dental prosthetic surgery with endosseous implants in early nineties. Cranin and coworkers published first research work on Zirconia in 1975. Ceramic implants were introduced for osteointegration, less plaque accumulation resulting in improvement of the soft tissue management, and aesthetic consideration as an alternative to titanium implants [81,82].

b) Monoclinic (M), cubic (C), and tetragonal (T) are the three crystal forms in which polymorphic Zirconia structure is present. The C-phase of pure Zirconia can be stabilized by adding CaO, MgO, and Y_2O_3 (Yttrium) resulting in multiphase material called partially stabilized zirconia (PSZ) combining cubic, monoclinic, and tetragonal phases in the order of importance. Tetragonal zirconia polycrystals (TZP), containing tetragonal phase only can be obtained by adding Yttrium at room temperature. Yttria stabilized TZP possesses low porosity, high density, high bending, and compression strength and is suitable for biomedical application [83].

C) Polymers

a) **PEEK (Polyetheretherketone):** Has excellent mechanical properties & low Young's modulus (3-4 GPa). Can be modified by addition of carbon fibers to increase the elastic modulus to 18 Gpa, which is nearly similar to bone modulus 14 Gpa.

Implant surface treatment and coatings (surface topography)

The bone response, which means rate, quantity and quality, are related to implant surface properties, the composition and charges are critical for protein adsorption and cell attachment [84]. Hydrophilic surfaces seem to favor the interactions with biological fluids and cells when compared to the hydrophobic ones [85,86] and hydrophilicity is affected by the surface chemical composition. Various techniques of surface treatments have been studied and applied to improved biological surface properties, which favors the mechanism of osseointegration [72,87] surface roughness is measured by (Ra)or (Sa)value, and classified:

1. Smooth 0.00-0.4 µm.
2. Minimally rough 0.5-1 µm.
3. Moderatly rough 1-2 µm (best bone response for osteointegration).
4. Rough > 2.0 µm (increased incidence for peri-implantitis.

Machined surface the first generation of Osseo integrated implants had a relatively smooth machined surface (Branemark, et al. 1969). The machined implant surface is solely turned and considered to be minimally rough Different roughness values have been published using different measuring techniques. Moreover, manufacturing tools, bulk material, lubricant and machining speed will influence the resulting surface topography. Typical Sa values for machined surfaces are 0.3-1.0 µm.

Surface roughness either additive or reductive:

1) Additive:

a) Titanium plasma-spraying: This method consists in injecting titanium powders into a plasma torch at high temperature. The titanium particles are projected onto the surface of the implants where they condense and fuse together, resulting in a titanium plasma sprayed (TPS) coating with an average roughness of around 7 µm. This procedure increases substantially the surface area of the implants [88].

b) Calcium phosphate coatings: Calcium phosphate (CaP) coatings, mainly composed by hydroxyapatite, has been used as a biocompatible, osteoconductive and resorbable blasting materials [89]. The idea behind the clinical use of hydroxyapatite is to use a compound with a similar chemical composition as the mineral phase of the bone to avoid connective tissue encapsulation and promote peri-implant bone apposition [90].

2) Reductive:

a) **Acid-etching and double acid etching:** The immersion of a titanium dental implant in strong acids such as hydrochloric acid, sulfuric acid, nitric acid and hydrogen fluoride is another method of surface modification which produces micro pits on titanium surfaces with sizes ranging from 0.5 to 2 µm in diameter [91]. The resulting surface shows an homogenous roughness, increased active surface area and improved adhesion of osteoblastic lineage cells [92]. Dual acid-etching consist in the immersion of titanium implants for several minutes in a mixture of concentrated HCl and H_2SO_4 heated above 100 ◦C to produce a micro-rough surface [93].

b) **Sand blasting:** Sandblasted surface Increased roughness of an implant could be achieved by blasting the surface by small particles, usually called sandblasting or grit blasting. (When the particles hit the implant surface it will create a crater. The surface roughness is hence dependent on the bulk material, the particle material, the particle size, the particle shape, the particle speed and the density of particles. The resulting surface roughness is usually anisotropic consisting of craters and ridges and occasionally particles embedded in the surface. The surface roughness increases with the size of the particles used (Wennerberg, et al., 1992) Typical Sa values are 0.5-2.0 µm.

c) **Sandblasted and acid etched surface (SLA), mod (SLA):** Commercially available dental implants are usually both blasted by particles and then subsequent etched by acids. This is performed to obtain a dual surface roughness as well as removal of embedded blasting particles. The etching reduces the highest peaks while smaller pits will be created and the average surface roughness will be reduced. Typical Sa values for blasted and acid etched implants are 1-2 µm (moderately roughness is the best for osteointegration). Mod (SLA) implants, after acid etching are rinsed under protective N_2 gas conditions and stored in isotonic saline solution. The process results in a more active hydrophilic surface.

d) **Electrochemical anodization:** Another method that has been shown to increase surface macrotexture and change surface chemistry is electrochemical anodization. The combination of potentiostatic or Galvano static anodization of titanium in strong acids at high current density or potential, results in thickening of the titanium oxide layer. Anodized surfaces interfere positively in bone response with higher values for biomechanical and histomorphometric tests when compared to machined surfaces [94,95].

Other Factors

Tilted implants

Presence of the maxillary sinus or the mental foramen may prevent implant treatment in the posterior maxilla or mandible. Tilting of distal implants supporting fixed restorations may be a valid treatment alternative. The stress at the most coronal bone-to-implant contact was identical irrespective of the angle of tilt, demonstrating that tilting of splinted implants does not result in increased stress [96].

Clinical moment arms

A total of six moments (rotations) may develop about the three clinical coordinate axes previously described (occlusoapical, labiolingual, and mesiodistal axes. Such moment loads induce microrotations and stress concentrations at the crest of the alveolar ridge at the implant-tissue interface, which lead to crestal bone loss.

Three clinical moment arms exist in implant dentistry: (1) Occlusal height, (2) Cantilever length, and (3) Occlusal width. Minimization of each of these moment arms is necessary to prevent unretained restorations, fracture of components, crestal bone loss, or complete implant system failure.

1) **Occlusal height:** The occlusal height serves as the moment arm for force components directed along the labiolingual axis-working or balancing occlusal contacts, tongue thrusts, or in passive loading by cheek and oral musculature, as well as force components directed along the mesiodistal axis.

2) In Division A bone, initial moment load at the crest is less than in Division C or D bone because the crown height is greater in C and D bone.

3) **Cantilever length:** Large moments may develop from vertical axis force components in prosthetic environments designed with cantilever extensions or offset loads from rigidly fixed implants. A lingual force component also may induce a twisting moment about the implant neck axis if applied through a cantilever length. An implant with a cantilevered bar extending 1, 2, and 3 cm has significant ranges of moment loads. A 100-N force applied directly over the implant does not induce a moment load or torque because no rotational forces are applied through an offset distance. The same 100-N force applied 1 cm from the implant results in a 100 N-cm moment load. Similarly, if the load is applied 2 cm from the

implant, then a 200 N-cm torque is applied to the implant-bone region, and at 3 cm a 300 N-cm moment load results.

4) **Occlusal width:** Wide occlusal tables increase the moment arm for any offset occlusal loads. Labiolingual tipping (rotation) can be reduced by narrowing the occlusal tables or adjusting the occlusion to provide more centric contacts.

Prosthesis fit

The non-passive restoration exerts continuous lateral forces on the implants, in case of screw retained fixed bridge or a hybrid bridge, the dentist can achieve passive fit by (single screw test), before final fixation of the bridge.

Loading time

Loading time of the prosthetic part of dental implants is as follows:

1) Immediate loading: the prosthesis is attached to the implants the same day the implants are placed.

2) Early loading: the prosthesis is attached at a second procedure, earlier than the conventional healing period of 3 to 6 months. The time of loading is started after some days/weeks.

3) Delayed loading the prosthesis is attached at a second procedure after a conventional healing period of 3 to 6 months.

Loading either functional, the crown/bridge is in contact with the opposing dentition in centric occlusion. Or, Nonfunctional, the crown/ bridge is not in contact in centric occlusion with the opposing dentition in natural jaw position.

Immediate loading is debated: For some researchers, the concept of immediate loading is satisfied as soon as the coronal portion of the prosthesis is inserted, even if it is kept out of occlusion [97]. For others, the term immediate loading can be applied only if the prosthesis is subjected to occlusal forces as soon as it is inserted [98].

Both the immediate and early functional loading of implants before lamellar bone formation carries an inherent biological risk. Shortened loading protocols may expose the healing bone to implant interface to mechanical overload as described in Wolffs Law and Frosts Mechanostat theory. Interfacial micromotion above the biological threshold can result in the subsequent loss of implant stability.

Conclusion

The operator should be aware of the biomechanical factors that affect the bone to implant contacts which have a direct effect on the success of the implant, he role of the beneficial biomechanical factors effect starts from the first day of placement and enhance, preserve the contact after loading, the operator will be able to choose the suitable implant from hundreds of implants in the market according to the category of the biomechanical factors when the biological factors are stable.

The biomechanical factors are

Host related factors

a) Bone density

b) Available remaining bone after extraction

c) Parafunctional habits

Forces and loading conditions applied on the implant

Implant-related factors

a) Implant macro design (implant body, length and diameter. Threads shape, pitch, lead, depth and width, and crest module.

b) Chemical composition and biomaterial of the implant.

c) Implant surface treatment and coatings (surface topography).

Other factors

a) Implant tilting.

b) Prosthetic passive fit.

c) Cantilever, crown high and occlusal table.

d) Loading time.

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