Clinical problems requiring implants for solution

Implant design refers to the three-dimensional structure of the implant, with all the elements and characteristics that compose it. Dental implants are subjected to various force magnitudes and directions during function. Because implants function to transfer occlusal loads to the surrounding biologic tissues, functional design objectives should aim to manage biomechanical loads to optimize the implant-supported prosthesis function. Thus, the primary functional design objective is to manage biomechanical loads to optimize the implant-supported prosthesis function. An implant has a macroscopic body design and a microscopic component of implant design.

The microscopic features are most important during initial implant healing and the initial loading period. The macroscopic implant body design is most important during early loading and mature loading periods. The product used by the implant team may increase or decrease the risk of screw loosening, crestal bone loss, implant body bone loss, peri-implantitis, esthetics of soft tissue drape, implant failure, and implant body fracture.

INTRODUCTION

Early implants with documented success were fabricated from noble metals or base metals shaped in either basic or pin designs that attempted to create natural teeth, which could then be connected to transmucosal prosthesis. Failures were believed to be caused by poor biomechanics, especially poor stabilization.

These implants had limited success, and mechanical and biological failures prompted dentists to create design that, in many instances, had no semblance to tooth morphology. The most successful designs of this type are staple, subperiosteal, and blade form implants.

A favorable implant design may compensate for risk of occlusal loads in excess of normal, poor bone densities, less than ideal implant positions and number, or less than an ideal implant size. In the past, implant body design was driven by the surgical ease of placement.

A surgically driven implant design will tend to have a tapered, short implant body or a press fit insertion. These features permit the implant to be surgically placed most easily.

A cylinder or press fit implant has a friction fit insertion and may have less risk of pressure necrosis to too tight an insertion pressure, has no need to bone tap, and may have the cover screw already in place because no rotational force is required to insert the implant.

As a result, cylinder or press fit implants are the easiest to insert. After 5 years of loading, reports of the cylinder implants including loss of crestal bone and implant failure are most often observed. This is related to a fatigue overload condition and harmful shear loads on the bone causing large bone turnover rates and ultimately less bone implant contact percent and higher risk of overload failure. Another focus of several implant designs is to reduce the plaque-related complications.

With this concept in mind, one consistent implant body design features smooth metal surfaces at the crestal portions of the implant. A smooth crest module of the implant is easier to clean related to oral hygiene methods and collects less plaque than the rougher surfaces.

Therefore, the rationale is that if bone loss occurs at the marginal regions of the implants, the smooth implant surface will harbor less plaque and be Implant Design and Stress Distribution International Journal of Oral Implantology and Clinical Research, easier to clean.

The problem with this philosophy is the smooth crest module is initially placed below the crest of the bone and is a design that encourages marginal bone loss from the extension of a biological width after implant uncovery and from shear forces after occlusal loading. As a result, this design feature increases the peri-implant sulcus depth.

Most implant body complications are related to early implant failure after loading, marginal bone loss before loading but after exposure of the implant, and marginal bone loss after the loading of the implant bone interface. Implant failures are most often observed as early loading failures in softer bone types or shorter implant length.

Thus, implant body designs should attempt to primarily address the primary causes of complication, i.e., the factors that address the loading conditions of the implant after the implants are placed in function.

Failure of osseo integrated implants is generally not related to mechanical failure of the load-bearing artificial structure, but is induced by bone weakening or loss at the peri-implant region. Bone resorption can be activated by surgical trauma or bacterial infection as well as by overloading at the bone–implant interface.

Under functional forces, overloading of the peri-implant bone can be induced by a shortcoming in the load transfer mechanism, primarily due to improper occlusion, prosthesis and/or implant design, and surgical placement. As a consequence, high stress concentration at the bone–implant interface may arise and, according to well-supported hypothesis, related strain fields in the bone tissue may stimulate biological bone resorption jeopardizing implant effectiveness. As far as implant shape is concerned, design parameters that primarily affect load transfer characteristics, i.e., the stress/ strain distribution in the bone include implant diameter and the length of the bone–implant interface, as well as in the case of threaded implants, thread pitch, shape, and depth. To increase the surface area for osseous integration, threaded implants are generally preferred to smooth cylindrical ones. Depending on bone quality, surface treatments and thread geometry can significantly influence implant effectiveness, in terms of both initial stability and the biomechanical nature of the bone–implant interface after the healing process. Smooth-sided, cylindrical implants provide ease in surgical placements; however, the bone–implant interface is subject to significantly larger shear conditions. In contrast, a smooth-sided, cylindrical, tapered implant provides for a component of compressive load delivered to the bone– implant interface, depending on the degree of taper.6 The greater the taper, the greater is the component of compressive force delivered to the interface. As a negative feature, the greater the taper of a smooth-sided implant, the less the overall surface area of the implant body under load and the less initial stability provided by that implant at an immediate extraction and implant insertion. Implant body designs with threaded features have the ability to convert occlusal loads into more favorable compressive loads at the bone interface; therefore, thread shape is particularly important when considering long-term load transfer to the surrounding bone interface. Under axial loads to a dental implant, a buttress or square-shaped thread would transmit compressive forces to the bone.

EFFECT OF THREAD DESIGN/GEOMETRY

Threads are used to maximize initial contact, improve initial stability, enlarge implant surface area, and favor dissipation of interfacial stress. Threaded implants have been shown to play an important role in increasing mechanical osseointegration7,8 and influencing stress around implants during loading.9 Huang et al10 reported that "threaded implants could reduce both bone stress and the implant–bone sliding distance, thus potentially improving initial implant stability and long-term survival." Chun et al11 indicated that a square thread shape with a small radius distributes stress more effectively. Thread depth, thread thickness, thread pitch, thread face angle, and thread helix angle are varying geometric parameters that determine the functional thread surface and affect the biomechanical load distribution of the implant. Thread Pitch Thread pitch is the distance measured parallel between adjacent thread form features of an implant.12 The height of the threaded portion of the implant body divided by the pitch equals the threads per unit length. The smaller (or finer) the pitch, the more threads on the implant body, if all other factors are equal. Of all the design variables, pitch has the most significant effect on changing the surface area on a threaded implant. The thread pitch may be used to help resist the forces to bone with poorer quality.13 Because the softest bone types are 58% weaker than ideal bone quality, the implant thread number may be increased to increase the overall surface area and reduce the amount of stress to the weaker bone trabeculae. Therefore, if force magnitude is increased, implant length is decreased, or bone density decreased, the thread pitch may be decrease to increase the thread number and increase the functional surface area. Thread Shape Thread shapes in dental implant designs include square, V-shaped, and buttress and reverse buttress. The face angle of the thread can change the direction of load from the prosthesis to a different force direction at the bone. Preeti Yadav et al 36 Under axial loads to a dental implant, a V-shaped thread face is comparable to the buttress thread when the face angle is similar and is usually 30°. A square thread design has been suggested to reduce the shear component of force by taking the axial load of the prosthesis and transferring a more axial load along the implant body to compress the bone. The thread shape has primarily design applications for loading conditions, but may also contribute to the initial healing stage for the direct bone interface. The face angle of the thread or plateau in an implant body can modify the direction of the occlusal load imposed on the prosthesis and abutment connection to a different direction at the bone interface. The face angle of the V-shaped thread is 30° off the long axis, whereas a square thread may be perpendicular to the long axis. As a result, occlusal loads in an axial direction of an implant body may be compressive at the bone interface when the implant body incorporates square-shaped threads, but can be converted to higher shear loads at the bone interface when the implant body incorporates V-shaped threads.12 A shear force in a V-thread and reverse buttress thread is 10 times greater than the shear force on a square thread.1 The reduction in shear loading at the thread bone interface provides for more compressive load transfer, which is particularly important in compromised bone density, short implant lengths, or higher force magnitudes. Different thread shapes with the same pitch indicate that implant with different total contact areas at the implant–bone interface affects the primary stability. Previous research has revealed that stress loading of threaded implants is maximal at the interface between the first pitch of the implant and the cortical bone.14 The thickness of the cortical bone ranges between 0.8 and 2.0 mm on average, with thicker bone having a higher load-bearing capacity.15-18 Kong et al19 emphasized that thread pitches exceeding 0.8 mm were optimal selections for a screwed implant by biomechanical consideration. Interestingly, Lee et al20 pointed out that square thread with a 0.6 mm pitch has optimal contact area and stress values. Chung et al21 found that implants with a pitch distance of 0.6 mm exhibited more crestal bone loss as compared with the implants with pitch distance of 0.5 mm. Lan et al22 found that the loading type is the main factor of influence on stress distribution, and that in biomechanical consideration, thread pitches exceeding 0.8 mm are more appropriate for screwed implants. Each type of thread form has its optimal thread pitch with regard to lower concentration of bone stress. Thread Depth The thread depth is the distance between the major and minor diameters of the thread.12 Conventional implants provide uniform thread depth throughout the length of the implant. A straight minor diameter results in uniform cross-sectional area throughout a parallel-walled implant length. A tapered implant often has a similar minor diameter, but the outer diameter decreases in relation to the taper, so the thread depth decreases to the apical region. As a result, this implant design has overall less surface area, which is more critical in shorter implant lengths. Thus, the implant body taper may result in higher stresses, especially in shorter implant lengths.23 The greater the thread depth, the greater is the surface area of the implant, if all other factors are equal. The implant increases surface area by 15 to 25% for every 1 mm increase in diameter.24 However, as an implant becomes wider, the depth of the thread may be deeper without decreasing the body wall thickness between the inner diameter and the abutment screw space within the implant. Thus, the thread depth may be modified relative to the diameter of the implant and, thereby, the overall surface area may be increased by 150% for every 1-mm increase in diameter. Another recent approach has been the introduction of a rounded thread design that claims to induce “osteocompression”. In dentistry, controlled functional osteocompression is the compaction created by the tapping procedure.

CREST MODULE CONSIDERATIONS

The crest module of an implant body is the transosteal region, which extends from the implant body and often incorporates the antirotation components of the abutment implant connection. 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), and a prosthetic influence. Therefore, this area of the implant body is a determinant for the overall implant body design. Bozkaya et al25 compared implant systems with different thread profiles and crestal modules. They found that moderate occlusal load did not change the compact bone. However, when extreme occlusal loads were applied, overloading occurred near the superior region of the compact bone. Hence, the authors concluded that the crestal module may play a role in minimizing stresses to bone. Schrotenboer et al26 compared the effect of microthreads vs. smooth neck and platform switching vs. equal diameter abutment on crestal module. They concluded that stress was concentrated on the coronol portion of the bone crest. The crest module of an implant should be slightly larger than the outer thread diameter of the implant body to completely seal the osteotomy, providing a barrier and deterrent for the ingress of bacteria or fibrous tissue during initial healing. It also provides greater initial stability, Implant Design and Stress Distribution International Journal of Oral Implantology and Clinical Research, May-August 2016;7(2):34-39 37 IJOICR especially, in softer unprepared bone as it compresses the crestal bone region.24 A larger crest module diameter increases the surface area, which can decrease stress at the crestal region. Because the stresses are highest in this region, the greater surface area decreases stress to the bone and increases the strength of the implant body. 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 antirotational hex of the abutment to implant body connection.27 Most of the occlusal stresses occur at the crestal region of an implant design.28,29 A smooth, parallel-sided crest module will increase the risk of bone loss after loading. Smooth metal promotes shear stresses in the adjacent bone interface.30 Any crest module design that incorporates an angled geometry or grooves to the crest module, coupled with a surface texture that increases bone contact, will impose a beneficial compressive component to the contiguous bone and decrease the risk of bone loss. The prosthetic features of the crest module may affect the implant design. In an internal hex implant, the antirotational feature of the abutment is designed within the implant body. As a result, the implant body is lower in profile and easier to cover with soft tissues during surgery. In addition, the antirotational feature is often deeper within the body compared with external hex implants. However, because the antirotation feature is wider than an abutment screw, the wider body diameter at the crest module is reduced. As a result, the threads on the outside of the implant body cannot be designed at or above the antirotational feature of the implants. Therefore, greater smooth metal and shear forces are observed above the first implant body thread compared with an implant with an external hex. Apical Design Considerations Most root form implants are circular in cross-section. Round cross-sections, however, do not resist torsion/ shear forces when abutment screws are tightened or when freestanding, single-tooth implants receive a rotational (torsional) force. As a result, an antirotational feature is incorporated into the implant body, usually in the apical region. The most common design is a hole or vent. Bone can grow through the apical hole and resist torsional loads applied to the implant. The apical hole region may also increase the surface area available to transmit compressive loads to the bone. A disadvantage of the apical hole occurs when the implant is placed through the sinus floor or becomes exposed through a cortical plate. The apical hole may fill with mucus and become a source of retrograde contamination or will likely fill through fibrous tissue. Another antirotational feature of an implant body may be flat sides or grooves along the body or apical region of the implant body. Bone grows against the flat or grooved region and helps resist torsional loading. In addition, the grooves or recessed areas of the apical portion of the implant help to enhance the "self-tapping" aspect of an implant design. The recess areas may be designed to decrease the angle of the cutting thread along the apical portion of the implant. As a result, less torque is required to thread the implant into the bone. Also, the apical end of each implant should be flat rather than pointed. Pointed geometry has less surface area, thereby raising the stress level in that region of bone. Additionally, a V-shaped apex may irritate or inflame the soft tissues, if any movement occurs.

EFFECT OF SHAPE DIAMETER AND LENGTH

The macrodesign or shape of an implant has an important bearing on the bone response; growing bone concentrates preferentially on protruding elements of the implant surface, such as ridges, crests, teeth, ribs, or the edge of the threads, which apparently act as stress risers when load is transferred. The shape of the implant determines the surface area available for stress transfer and governs the initial stability of the implant. Transforming shear forces into more resistant force types at the bone interface is the purpose of incorporating threads into the implant design as a surface feature. Implant Length Implant length is the dimension from the platform to the apex of implant. Most common lengths are between 8 and 13 mm, which corresponds quite closely to normal root length. The significance in increased implant length or its ability to achieve osseointegration is not found at the crestal bone interface, but rather in initial stability and the overall amount of bone–implant interface. The increased length can provide 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 or change its ability to achieve osseointegration. Implant Diameter Implant diameter is the dimension measured from the peak of the widest thread to the same point on the opposite side of the implant. It is considered to be more important than the implant length in the distribution of loads to the surrounding bone. At least 3.25 mm in diameter is required to ensure adequate implant strength and most implants are Preeti Yadav et al 38 approximately 4 mm in diameter. From a biomechanical standpoint, the use of wider implants allows an engagement of a maximal amount of bone, and a theoretically improved distribution of stress in the surrounding bone. It has been confirmed that more bone contact area provides increased initial stability and resistance to stresses. The increase in diameter will result in a higher percentage of bone contact by increasing the surface area of the implant. Previous research, done by Misch,33 shows that increasing the diameter in a 3 mm implant by 1 mm increases the surface area by 35% over the same length in overall surface. Balshi34 evaluated the causes of implant fractures, and indicated biomechanical or physiological overloads as the most common reason for implant fracture. The source of the overload is likely patient parafunction habits and incorrect prosthesis design, which might be responsible for the creation of undesired bending moments. He recommended the use of implants with larger diameters to provide larger metal bulk, therefore, increasing implant strength by decreasing the applied level of stress. The use of wider components also allows for the application of higher torque in the placement of prosthetic components. However, according to Okumura et al,35 from a biomechanical viewpoint, to improve implant success odds in the posterior maxilla, rather than implant selection (design or parameter), careful preoperative evaluation of the cortical bone thickness at the planned implant site is recommended. If this cortical bone is very thin or even lacking, implant treatment should be carried on with caution by progressive loading in the range of functional loads. Implant Shape The shape of dental implants has been one of the most contested aspects of design among the endosseous systems and may have an effect on implant biomechanics. Most current implant systems are available as solid or hollow screws or cylinders. Among screw type designs, considerable modification has been made to the crestal and apical portion of the implant to increase self-tapping and decrease heat generation. Other designs have been developed to imitate root anatomy and incorporate a stepped cylindrical design, analogous to the root at both cervical and apical ends. These stepped cylindrical implants show more even stress dissipation compared with cylindrical or tapered implants, and improved loading of the crestal bone supporting of the alveolar bone from the root analog shape of the implant. The following would be the design principles one would want to achieve through an implant design36: • Gain initial stability that wouldreduce the thresholdfor the "tolerated micromotion" and minimize the waiting period required for loading the implant. • Incorporate design factors that would diminish the effect of shear forces on the interface (such as surface roughness related and thread features), so that marginal bone is preserved. • Design features that would stimulate bone formation and/or facilitate bone healing (secondary osseointegration).

CONCLUSION

Stress and strain fields around osseointegrated dental implants are affected by a number of biomechanical factors, including the type of loading, material properties of the implant and the prosthesis, implant geometry, surface structure, quality and quantity of the surrounding bone, and the nature of the bone–implant interface. Several implant concepts have been developed, and many implant types are commercially available in different sizes, shapes, materials, and surfaces. To analyze the effectiveness and reliability of endosseous implants, revealing possible risks of implant failure and stress analysis of bone–implant mechanical interactions are important. The complex geometry of the coupled bone–implant biomechanical system prevents the use of closed-form approach for stress evaluation. Therefore, the behavior of endosteal dental implants can be investigated by numerical techniques like finite element analysis method to predict stress and strain distributions at peri-implant regions, investigating the influences of implant and prosthesis designs, the magnitude and direction of loads, and bone mechanical properties, as well as modeling different clinical scenarios.