Review in Advance first posted online on May 29, 2014. (Changes may still occur before final publication online and in print.)

Effects of Biomechanical Properties of the Bone–Implant Interface on Dental Implant Stability: From In Silico Approaches to the Patient's Mouth

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Annu. Rev. Biomed. Eng. 2014. 16:187–213

The *Annual Review of Biomedical Engineering* is online at bioeng.annualreviews.org

This article's doi: 10.1146/annurev-bioeng-071813-104854

Keywords

bone, bone–implant interface, dental implant, osseointegration, biomechanical properties, stability, bone remodeling

Abstract

Dental implants have become a routinely used technique in dentistry for replacing teeth. However, risks of failure are still experienced and remain difficult to anticipate. Multiscale phenomena occurring around the implant interface determine the implant outcome. The aim of this review is to provide an understanding of the biomechanical behavior of the interface between a dental implant and the region of bone adjacent to it (the bone–implant interface) as a function of the interface's environment. First, we describe the determinants of implant stability in relation to the different multiscale simulation approaches used to model the evolution of the bone–implant interface. Then, we review the various aspects of osseointegration in relation to implant stability. Next, we describe the different approaches used in the literature to measure implant stability in vitro and in vivo. Last, we review various factors affecting the evolution of the bone–implant interface properties.

Contents

1. INTRODUCTION

Tooth replacement by means of dental implants has become a widely popular procedure in contemporary dentistry. Dental implants differ from natural teeth because of the absence of periodontal ligament (PDL), and as a result implants are osseointegrated to the bone with limited movement. The PDL acts as a cushion around teeth and is composed of multioriented collagen fiber groups. It not only acts as a shock absorber but also has neurophysiological receptors that can detect changes in occlusal forces encountered. Implants, by contrast, are integrated to the bone and thus are sensitive in detecting even a small occlusal load (1). Implants, unlike teeth, undergo stresses that are concentrated at the crestal bone region instead of evenly distributed along the implant's surface (2). Although dental implants have relatively high survival rates (3), a trend toward higher incidences of complications in implant prostheses, compared with tooth-supported fixed prosthodontics, has been observed (3). A meta-analysis of studies reporting on the survival and complication rates of tooth-supported fixed dental prostheses (FDPs), implant-supported FDPs, and single crowns showed a significantly higher complication rate for implant-supported fixed restorations (3). Over a 5-year period, implant-supported FDPs had a cumulative complication rate of 38.7%, whereas tooth-supported conventional and cantilever FDPs had complication rates of only 15.7% and 20.6%, respectively (3). To avoid increased complications, clinicians must have a good understanding of the biomechanical aspect of implant prostheses.

This article reviews different types of work, including fundamental, animal, and clinical studies. We first focus on aspects related to the multiscale and multitime nature of dental implant stability with a particular focus on the biomechanical properties of the interface between the dental implant and the region of bone adjacent to it (the bone–implant interface). We then review the main determinants of osseointegration from a biomechanical point of view. The different experimental modalities allowing assessment of implant stability are investigated, and, eventually, the main factors affecting the evolution of implant status are tackled. Purely biochemical or biological factors (4) are beyond the scope of the present article, as we focus on biomechanical aspects because they are still poorly understood. Similarly, this article is dedicated to aspects related to bone biomechanics rather than to the mechanical strength (usually assessed in fatigue) of the implant material (5).

2. TIME EVOLUTION OF DENTAL IMPLANT STABILITY

Implant stability (a notion relevant to orthopedic as well as dental surgery) is defined as the capacity of an implant to carry loads in the axial and lateral directions as well as in rotation. Two different kinds of stability are distinguished: primary and secondary stabilities. From a bone structural and morphological point of view, implant stability results from the contact between bone and the implant surface.

2.1. Primary Stability

Primary stability occurs just after the surgical intervention and is a phenomenon of a purely biomechanical nature, related mostly to bone quality at the implant site. The implant design, the surgical protocol, and the congruence between bone and the implant play an important role in an implant's primary stability (see Section 6). Friction phenomena at the bone–implant interface is one of the main determinants of primary stability (6).

Primary stabilization of the implant in bone tissue is a necessary condition to obtain implant osseointegration. If the implant's primary stability is not sufficient, micromovements may occur, preventing optimal healing conditions and leading to the formation of fibrous tissue and to surgical failure. Adequate primary stability reduces interfacial displacements of the implant and newly formed bone tissue, thus allowing bone apposition on the implant surface. However, excessive stresses acting on the bone–implant interface after surgery may prevent nutrient circulation and may cause bone necrosis around the interface, which may decrease the quality of osseointegration (see Section 4). An appropriate balance between reasonable initial stresses to allow bone healing and adequate primary stability to minimize micromotion at the bone–implant interface is the key determinant for the long-term success of implant surgery.

2.2. Secondary Stability

Secondary stability occurs after a certain healing period (typically several months) and corresponds to the initial stability reinforced by new bone formation and maturation at the bone–implant interface. The quality and duration of bone formation and maturation are related to several factors, such as the implant surface and bone quality at the implant site. It is generally assumed that osteoclastic activity undermines primary stability before new bone formation prevents implant micromotion. Therefore, a decrease in stability may take place soon (around 1–3 weeks) after surgery.

Secondary stability is achieved by bone remodeling and subsequent bone apposition bridging old bone and the implant surface. After surgical intervention, loads are transmitted directly to the peri-implant bone.

3. MODELING OF A MULTISCALE PROBLEM

Phenomena involving load transfers between bone and an implant are intrinsically of a multiscale and multitime nature because various scales (from the nanometric scale up to the organ scale) and various durations (from seconds up to several months) play a role in the clinical outcome of an implant. The difficulty of ensuring implant stability derives from (*a*) the complex multiscale nature of bone tissue; (*b*) the time-evolving nature of bone tissue (typical times of bone evolution are of the order of several weeks, whereas the mechanical stresses are applied with typical time constants of around several seconds) (see Section 4); and (*c*) the presence of an interface that presents mechanical issues, in particular those arising from contact interactions between bone and the rough implant surface.

At the scale of the implant (from several millimeters up to a centimeter), load is transferred from the prosthesis (i.e., the part emerging from the implant and replacing the teeth) to the implant and then to the bone–implant interface. When considering the problem at a scale of around 200 μm, the types of cortical and trabecular bone microstructure around the bone–implant interface play an important role in nutriment circulation and stress distribution around the implant. The roughness of the implant surface, which is on a scale of 1–50 μm, constitutes an important factor in stimulating the production and maturation of newly formed bone tissue (see Section 4.3). At the scale of several hundred nanometers, adhesion phenomena between bone and the implant surface occur through complex cascades of biochemical phenomena (7), which can be enhanced using nanoscale modifications of the implant surface (8).

More specifically, implant surgery usually affects bone tissue located within about several hundred micrometers around the implant (9); as a result, subsequent healing and remodeling in this region will affect the evolution of the local interfacial bone properties. Detailed models of events at the hundreds-of-micrometers scale should take into account these phenomena, and results should be used in macroscopic mechanical models of the entire bone–implant system. More complex phenomena occur at the cellular or molecular level, and the validity of continuum mechanics may become questionable at that scale. At the nanoscale, substrate elasticity and cell strain are important parameters influencing stem cell fate decisions (8, 10, 11).

Multiscale modeling (MSM) approaches, such as homogenization methods, aim at determining the mechanical properties of a system at the macroscopic scale based on knowledge of the geometrical and mechanical properties of each subsystem at lower scales. Modern MSM approaches take into account the coupling of the different phenomena influencing the macroscopic mechanical behavior of the modeled object. To do so, ordinary MSM approaches consider different successive interconnected scale changes (corresponding to elementary bricks) in order to link the nanoscale up to the macroscopic scale (see Section 3.1). MSM models derive from concepts developed in micromechanics and constitute powerful tools to understand the multiscale problem of estimating the determinants of dental implant stability. MSM approaches can also be used to solve the inverse problem—that is, determining the micro- or nanoscopic bone properties from results of experiments carried out at the macroscopic scale. Such approaches are widely used in mechanical engineering, and some authors have started applying these kind of models in the bone field, determining the macroscopic properties of cancellous (12) and cortical bone (13) based on knowledge of its properties at the microscopic scale.

Critical engineering thinking is needed to estimate the determinants of dental implant stability because strong coupling of various multiphysical phenomena should be taken into account in order to relate dental implant stability and its evolution to implant properties and environments. Moreover, empirical or purely phenomenological studies cannot be adapted to tackle the problem of estimating dental implant stability because the number of parameters playing a role in implant

outcome is too important to be explored only with animal or clinical approaches. Therefore, modeling and simulation studies are necessary. Ethical issues require a drastic reduction in the number of animals used in such protocols. Additionally, different bone properties, such as microstructure and material properties, evolve simultaneously, so it is not possible to determine the influence of each parameter independently. Engineering science (both theoretical and experimental) is the only path that will allow us to tackle this problem in an efficient and rigorous manner.

3.1. Homogenization Approaches

Different studies have aimed at climbing the hierarchy of scales, from the nano- up to the macroscopic level. The first step is determining tissue-independent building blocks at the micro- and nanolevels, which can then be used to predict, through their interaction, bone elasticity at the macroscopic scale.

At the organ level, models are often continuum based and describe the variation of bone apparent density as a function of biological and mechanical stimuli. At the scale of bone tissue, models account for the bone microarchitecture and remodeling properties. At the cellular level, partial differential equations describe cellular interactions in the temporal domain (14), and molecular mechanics approaches are employed to model the elastic properties (15). In addition to the adaptation of modulating cells, mechanochemical signals (which are still poorly understood) also influence cell proliferation, migration, and differentiation. However, although they are informative, most models consider only limited scale changes and do not take into account the interaction between the nanoscale and the organ scale.

Interestingly, multiscale approaches developed in the literature that use concepts derived from continuum micromechanics allow the modeling of bone elastic behavior (16). Models considering mechanotransduction as well as the evolution of bone mechanical properties should account for the flow channels (17) that provide conduits for fluid flow, enhancing molecular and cellular transport and inducing shear stresses via fluid drag at the cell surfaces.

3.2. Using High-Resolution Medical Imaging Techniques

Modern medical imaging technology allows in vivo scanning of large specimens at the microscale level. Coupling three dimensional (3D) high-resolution reconstructions with numerical simulation tools has become a common approach to determine bone mechanical properties (12) because of the difficulty of achieving reliable estimations of stress and strain fields in vivo. A critical issue is accounting for anisotropy and heterogeneity of the bone material properties. When heterogeneity is accounted for, empirical regression analyses between tissue mineral density and Young's modulus have often been employed. Recently, a homogenization model was developed to infer from the nanoscale up to the organ scale using data obtained from synchrotron radiation microcomputed tomography (μCT) (19, 20). The approach used in this three-step method is illustrated in **Figure 1**. Each scale change is based on continuum micromechanics, which provides estimates of the effective elasticity tensor for materials with matrix-inclusion microstructures (21–23). Continuum micromechanics stems from Eshelby's (24) solution of the elastic problem of an isolated inclusion embedded in an infinitely extended matrix. Eshelby's solution describes the elastic fields (strain, stress) engendered by such an inclusion under homogeneous strain/stress boundary conditions at infinity. C_{hom} is the homogenized elasticity tensor describing the effective elastic properties of a material with microstructure. In continuum micromechanics theory, C_{hom} is represented as follows:

$$C_{\text{hom}} = \sum_r f_r c_r : A_r,$$

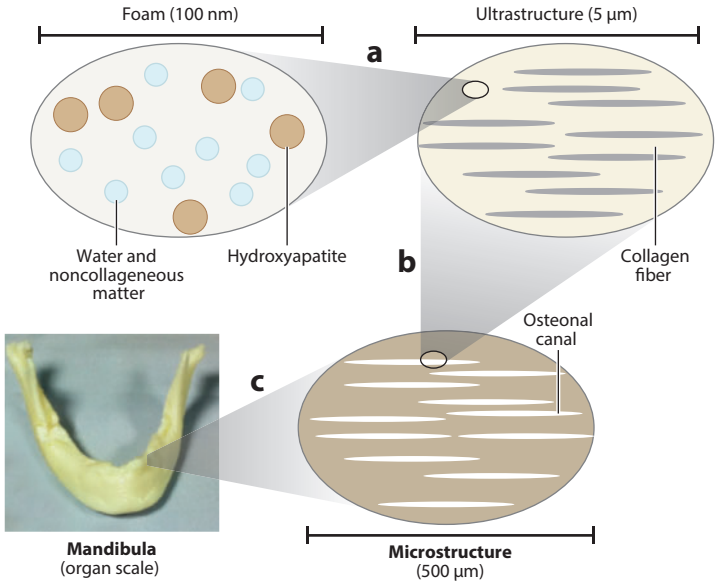


Figure 1

Schematic description of the three-step homogenization method applied to cortical bone using the continuum micromechanics theory developed in References 19 and 20. The scale changes are labeled and reflect (*a*) moving from the nanoscale up to the ultrastructure, (*b*) moving from the ultrastructure to the microstructure, and (*c*) moving from the microstructure to the scale of the organ. In the microstructure (*bottom right*), the white regions schematically represent Haversian canals or different mesoscopic pores of the bone tissue. Material properties of the ultrastructure (*top right*) and of the mineral foam (*top left*) are homogenized within each region of interest. (Figure courtesy of Dr. Romain Bosc.)

where f_r is the volume fraction of phase r, and c_r and A_r are the elasticity and localization tensors of phase r, respectively. It can be shown that the expression of A_r reads (25)

$$A_r = (I + P_r^0 : (c_r - C^0))^{-1} : \left(\sum_s f_r (I + P_s^0 : (c_s - C^0))^{-1} \right)^{-1},$$

where I is the fourth-order symmetric identity tensor, C^0 is the elasticity tensor of the effective matrix when the phases are embedded, and P_r^0 is the Hill tensor of phase/inclusion r embedded in the effective matrix. The expression of the Hill tensor, P_r^0, depends on the shape of the inclusions representing the phase r and on the elastic tensor of the surrounding effective matrix.

To the best of our knowledge, existing multiscale models do not account for the temporal bone evolution due to remodeling phenomena, which prevents one from understanding aspects related to bone regulation and regeneration. Many challenges have to be faced in modeling bone regeneration, including a need for a model that provides a reliable description of the system evolution of bone tissue around a bone–implant interface.

3.3. Modeling the Multiscale and Multitime Bone–Implant Interface

Different approaches have been developed to model the biomechanical behavior of the bone–implant interface. As the biological threshold for micromovements is in the range of 100 to 200 μm (see Section 6), modeling is difficult because accuracy must be lower than around 10–20 μm (26). Finite element models (FEM) have been widely employed to simulate the mechanical

behavior of an implant at the organ scale, using, for example, large sliding contact elements in the case of primary stability (i.e., just after surgery) (27). To tackle this problem, different teams have considered Coulomb's law of friction (28), empirical remodeling laws (29), mesh morphing (30), nonlinear anisotropic FEM (31), or coupling of FEM with statistical techniques (32) to account for the uncertainty related to the in vivo situation. However, the existing approaches for modeling bone–implant interfaces do not account for the coupling of the multiscale and evolving natures of bone tissue nor for the influence of adhesion phenomena. Instead, most models assume homogeneous bone properties around the implant. Indeed, existing biomechanical modeling of the bone–implant interface often remains simplistic owing to a lack of experimental data at the scale of 1 to 100 μm.

4. OSSEOINTEGRATION AND BONE REMODELING

4.1. What Is Osseointegration in Relation to Implant Stability?

According to Brånemark & Skalak (33), an oral implant is osseointegrated if "it provides a stable and apparently immobile support of a prosthesis under functional loads without pain, inflammation or loosening." Furthermore, per these authors, an implant is osseointegrated if "there is no progressive relative motion between the implant and surrounding living bone and marrow under functional levels and types of loading for the entire life of the patient." Others define osseointegration in histological terms—for example, as "the formation of a direct interface between an orthopedic or dental implant and bone, without intervening soft tissue" (34). However, a histological definition is silent about the interface's ability to endure functional biomechanical loading, which is a key part of Brånemark's original definition. So although osseointegration has been defined by some as direct bone–implant contact, this review adopts Brånemark & Skalak's definition.

Real implants and bone are deformable materials, even though the deformations turn out to be relatively small under functional levels and types of loadings. Thus, use of the term immobile, in the context of osseointegration, does not mean completely immobile—a point also made in literature on implant stability (see Section 6.1). Indeed, Brånemark & Skalak's (33) warning that there should be "no progressive relative motion between the implant and surrounding living bone" indicates that some relative motion (often called micromotion) is inevitable between implant and bone owing to the deformability of the materials involved.

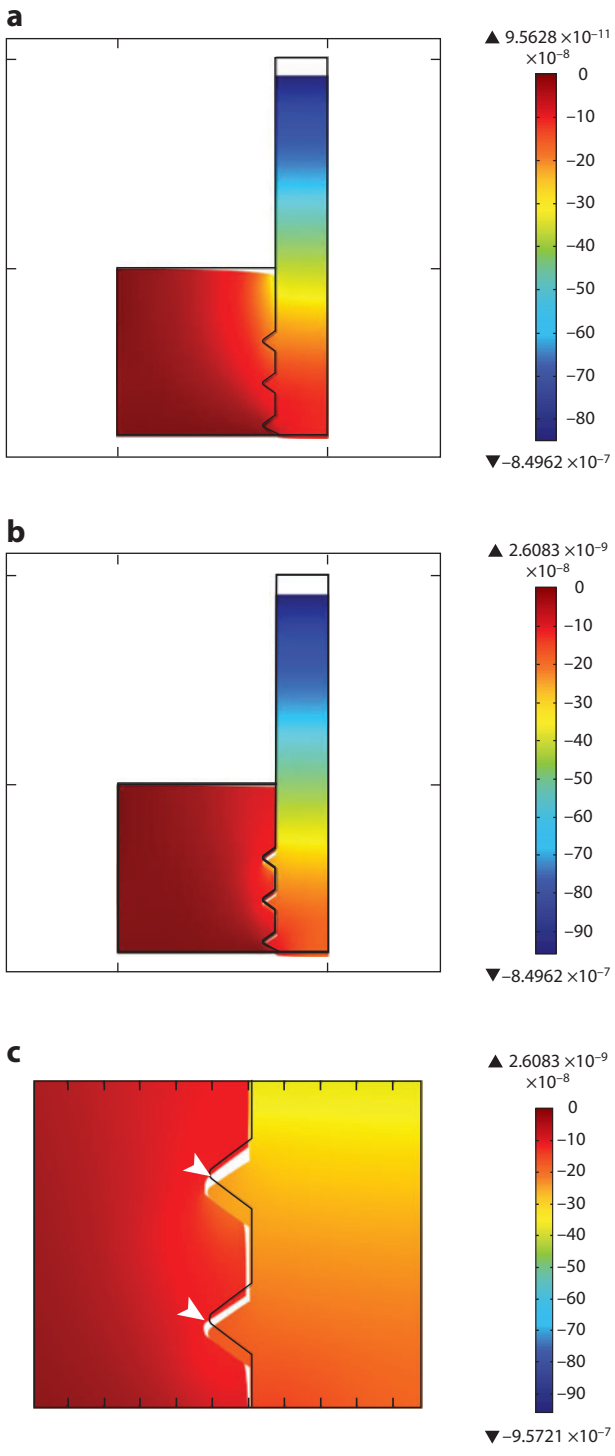
A working definition of relative motion comes from illustrative axisymmetric FEM of a titanium implant in bone that explore bonded versus nonbonded conditions right at the bone-implant boundary (see **Figure 2**). When a titanium implant is loaded by a downward force, the bonded and nonbonded FEMs share one similarity: Points inside the bone and implant move relative to one another, and there is relative motion between implant and bone. Examining just the interface between implant and bone, when the implant is bonded to bone, equal displacements occur for points at the bone–implant boundary, so there is no relative motion. However, with nonbonding along the bone–implant boundary (**Figure 2*b***), points on the implant surface do move relative to points on the bone surface, gaps open between the implant and bone (**Figure 2*c***, *arrows*), and relative motion occurs.

Figure 2 also illuminates three points about osseointegration. First, some relative motion of an implant relative to bone will usually occur owing to differing mechanical properties of implant and bone. Second, if a strong enough bond exists at the bone–implant boundary,[1] then that bond may

[1]The reader is directed to Reference 18 for a discussion of whether bone–biomaterial interfaces are strong enough in shear and tensile loading to carry interfacial stresses that develop during functional loading in vivo.

Figure 2

Axisymmetric finite element simulation illustrating bone–implant micromotion (relative motion)—that is, axial (z) displacements of the implant and cylinder of uniform bone when the top of the implant (0.5 mm in diameter) is loaded in the negative z direction by 10 N while the bottom of the bone cylinder remains fixed. The color scales on the right of each panel indicate z displacements in meters. (*a*) A bond exists at the bone–implant boundary. (*b*) No bond exists at that boundary. (*c*) A magnified view of the lower region in panel *b*, showing the gaps (*arrows*) that open along the bone–implant boundary without bonding.



prevent relative motion and gaps from opening at the bone–implant boundary (see Section 5.4). Third, per Brånemark & Skalak's (33) warning, any relative motion that does occur should not be so large that it causes biological damage (Section 6).

Some studies define osseointegration in terms of percentage contact of bone to implant (35). However, as bone heals over time around an implant, the amount of bone–implant contact may change, which, in turn, implies that the level of osseointegration changes with time. Likewise, implant A might have different percentage contact with bone compared with implant B. As a result, this use of the term osseointegration can be confusing when compared with the original, functionally based definition of osseointegration. Nevertheless, exploring changes in tissue contact and mechanical properties in a small region around an implant can give insight into local contributions to stability (36).

4.2. Biological Aspects of Bone Remodeling

One of the major challenges in implant dentistry today is to preserve peri-implant tissues and to maintain long-term implant esthetics and function. After an implant is successfully integrated, some marginal bone loss may occur, which causes many concerns to patients and their clinicians. This bone loss is often attributed to either biological or biomechanical factors (37). Recently, implant design has been shown to be a major factor in marginal bone loss due to biological bone remodeling (38) (see Section 6.3). The amount of bone remodeling was determined by the location of the rough/smooth border of a one-piece implant (39) and the microgap of a two-piece implant (40). In general, placing an implant too deep may cause the formation of deep probing depths as well as marginal bone loss, promoting the occurrence of peri-implant diseases.

Biological bone remodeling is defined as a process by which bone gradually alters its morphology in an attempt to adapt to any new external environment/load. It has been demonstrated that, just like all other bones in the human body, the jaw bone around implants can remodel (41), with most of the peri-implant bone remodeling occurring in the first 6 months (42). After 12 months, bone loss stabilizes at approximately 0.2–0.4 mm (42). It is essential for clinicians to understand the process of biological bone remodeling so they can properly control this phenomenon; otherwise, it may trigger deeper probing pocket depths around dental implants, which often lead to peri-implantitis (peri-implant diseases with significant bone loss). For this reason, it is crucial to take into account the effect of bone remodeling during the design of dental implants and presurgical assessment to maximize the success of implant therapy.

4.3. Newly Formed Bone Tissue Properties

Several in vitro approaches have employed animal models to measure the biomechanical properties of newly formed bone tissue, which can be distinguished from mature bone tissue in animal models using various fluorochrome labels (43) or backscattered electron imaging analysis (44). For example, small-angle X-ray scattering (SAXS) has been used to assess the thickness, orientation, and shape/arrangement of the mineral crystals in bone tissue (45). SAXS measurements were carried out near the implant surface (46) and showed that (*a*) the mineral crystals tended to be aligned with the surface of the implants and (*b*) the mineral crystal thickness increased linearly with distance from the implant. However, SAXS does not allow direct retrieval of mechanical properties.

Similarly, nanoindentation has been used, revealing lower values of Young's modulus and hardness in the vicinity of the implant (47). Scanning acoustic microscopy (SAM) has been used for measurements of bone acoustic properties in vitro (48), with a resolution of around 50 μm.

Table 1 Current methods used to assess implant stability along with their reliability and feasibility and the major concerns associated with them

	Reliability	Feasibility	Major concern
Traditional clinical methods			
Percussion	Low	Good	No actual value
Two instruments	Medium	Good	Low sensitivity
Radiograph	Low	Poor	Low sensitivity
Vibration analysis			
Periotest® (damping effect)	Low	Low	Not reliable
Resonance frequency analysis	Medium	Good	Low specificity
Torque test			
Insertion torque	High	Good	One-time assessment
Reverse torque	High	Fair	Too destructive

SAM was also applied for the qualitative assessment of biomechanical microstructural properties of the bone–implant interface (49). However, it remains difficult to clearly distinguish mature from newly formed bone tissue around dental implants because both tissues are usually spatially interconnected.

In order to overcome this difficulty, an approach coupling experimental surgery [using a dedicated animal model adapted from (50, 51)] with a multimodal experimental approach using nanoindentation (52, 53) and micro-Brillouin scattering (54) has been developed. This animal model allowed measurement of the biomechanical properties of newly formed bone tissue during bone healing thanks to the presence of a bone chamber, which contains newly formed bone tissue only. The ultrasonic velocity measured with the micro-Brillouin scattering device in newly formed bone tissue (4,970 $\pm$ 140 m/s, 7 weeks of healing time) was significantly lower than that in mature bone tissue (5,310 $\pm$ 40 m/s) for the same bone sample. Similarly, the Young's modulus measured with the nanoindentation device in newly formed bone tissue (15.85 $\pm$ 1.55 GPa) was significantly lower than that in mature bone tissue (20.66 $\pm$ 2.75 GPa). These results might be explained by the higher mineral content in mature bone than in newly formed bone tissue. Comparing the aforementioned findings allows the determination of the relative variation of mass density between newly formed and mature bone tissue, which is of the order of 13.5% (55). However, despite this progress, the time evolution of the biomechanical properties of newly formed bone tissue around implants remains unclear and should be investigated in more detail in the future.

5. MULTIMODAL APPROACHES FOR IMPLANT STABILITY ASSESSMENT

Implant stability, an indirect indication of osseointegration and a measure of clinical immobility of an implant, is a prerequisite for functional dental implants. As primary stability decreases, the possibility of micromotion between the implant surface and bone increases, which may result in implant failure (56). Hence, it is important to estimate implant stability before loading. The best way to assess osseointegration is through microscopy or histology, which is not clinically feasible. As a result, many methods/tools have been developed to measure implant mobility or stability (57). **Table 1** outlines the methods that can be used to assess implant stability as well as their reliability and feasibility and the major concerns associated with them.

5.1. Traditional Clinical Methods

5.1.1. Empirical percussion test. A percussion test is based upon vibrational-acoustic resonance theory. A metal instrument is tapped against an implant to determine if it is integrated. A crystal or ringing sound usually indicates successful implant integration, whereas a dull sound may suggest otherwise. However, this subjective method relies largely on the clinician's experience level. Hence, it cannot be used as a standardized tool for determining implant stability.

5.1.2. Instrument test. As in assessing tooth mobility, many clinicians use the back handle of two instruments to detect implant mobility. However, this approach is not very accurate and is also too subjective. Therefore, it is not often used today, especially in research, as it is nonquantifiable and empirical. Furthermore, an instrument test cannot be used to detect a failing implant (e.g., implant infection with bone loss).

5.1.3. Radiographic and magnetic resonance imaging interpretation. Radiographs have been widely used for detecting and monitoring peri-implant bone loss. Traditional radiographs such as peri-apical, bite-wing, and Panorex are utilized to aid in evaluating implant conditions, especially marginal bone loss (58). However, they can show only mesiodistal and not buccopalatal bone loss (59). Other shortcomings include difficulty in predicting implant stability and correlating the amount of marginal bone loss to the degree of implant stability. Furthermore, radiographs cannot be used to quantify bone quality or density. Recent research indicates that quantitative cone-beam computed tomography (CBCT) may have the capacity to detect bone density but that it falls short in providing information regarding implant stability (60). In particular, the resolution obtained with clinical X-ray-based techniques (clinical radiography or μCT) around the bone–implant interface is relatively low owing to distortion effects caused by the presence of titanium. The use of magnetic resonance imaging (MRI) (61) has been suggested, but such an approach remains of limited interest owing to magnetic fields disturbance due to the presence of titanium.

5.2. Biomechanical Elastoacoustic Analyses

5.2.1. Periotest®. The Periotest® (PT; Siemens AG Benshei, Germany) uses the damping effect, whereby impact force is applied as excitation force to detect implant stability (62, 63). The PT unit is composed of an electronically controlled device with a flexible connection to a tapping metallic rod. The rod taps the implant 20 times in 5 s, generating visual and acoustic signals. The signals are then converted to a unique value called PTV, which is dependent on the damping characteristics of tissues surrounding teeth or implants (64). Therefore, in the presence of impact force, the lower rigidity of the tested bone–implant system results in a longer contact duration. However, studies that measured implant mobility using the PT reported low sensitivity of the device (65) because of the wide value range (−8 to +50) used to reproduce the characteristics of the PDL (66) as well as the soft tissue surrounding teeth. In addition, PTV is significantly influenced by rod position along the intraoral height of the abutment and by direction (67). As a result, it is difficult to obtain a reliable reading in an oral cavity owing to the difficulty of placing the rod at a straight angle. Clinically, it is hard to control factors that may influence PTV [e.g., implant design, diameter, length, arch location, gender, bone quality, and quantity (68)]. In sum, the prognostic accuracy of the PT for implant stability cannot be ascertained owing to poor sensitivity, lack of resolution, and weak reproducibility (69).

5.2.2. Resonance frequency analysis. Resonance frequency analysis (RFA) works in a similar manner to the PT, except that it employs a small transducer that is screwed to the implant itself or an abutment. The transducer is composed of two piezoceramic elements: One serves as a vibrator to create a sinusoid signal, and the other acts as a signal receptor. When the implant is excited via wideband excitation, the system response is then amplified and recorded to measure implant stability (70). This is a noninvasive tool that provides information on the stiffness of the bone–implant interface and can be used to measure implant stability and bone density at different time points.

Osstell™ (formerly Integration Diagnostics Ltd, Göteborgsvägen, Sweden) makes the most commercially used RFA machine today. This newer version of the RFA system employs a SmartPeg screwed into the abutment, which is then vibrated when in proximity to a handheld wand. It uses implant stability quotient (ISQ) in place of Hertz as the unit of measurement (71). Osstell instruments convert resonance frequency values ranging from 3,500 to 8,500 Hz into ISQ values of between 0 and 100. The company suggests that a successful implant should have an ISQ greater than 65 and that ISQ less than 50 may indicate potential failure or increased risk of failure (72).

Many factors have been shown to affect RFA and ISQ readings. These include but are not limited to implant micro- and macrodesigns, implant length and diameter (73), implant positions (74), abutment length, exposed implant height above bone crest (73), bone quality and quantity, soft-tissue effects, stiffness of the bone–implant interface, implant healing time, tightness of the transducer rod, and the rod's position (57). Owing to these influencing factors, the reliability of RFA measurement remains low (75).

Nonetheless, RFA is the only tool that can supply clinically relevant information at various stages of implant treatment. At this time, RFA/ISQ has been advocated for the following clinical situations: to determine implant primary stability and osseointegration, to assess feasibility of immediate loading, and to monitor bone density and implant stability over time. RFA/ISQ could also theoretically be used to differentiate between implants with and without bone loss, although current evidence does not support this approach (76). ISQ value is fairly reliable in assessing an osseointegrated implant and the rigidity of the bone–implant interface (77). In cases in which integration is questionable, however, the ISQ value fluctuates (78). Hence, RFA/ISQ cannot provide definitive values for the success, failure, or long-term prognosis of an implant. More clinical studies are needed to provide a better understanding of how to use this tool in assessing implant stability and predicting implant failure before it occurs. More details and analyses carried out with the Osstell device are described in Section 6.3 below.

5.2.3. Quantitative ultrasound methods. Storani de Almeida et al. (79) carried out preliminary experiments with a screw inserted in an aluminum block and measured variations of the implant ultrasonic response. The technique was adapted to assess in vitro the amount of bone in contact with titanium prototype cylindrical implants using quantitative ultrasound (QUS) (80). Numerical simulations have also been performed (81), showing that implant ultrasonic response is sensitive to (*a*) the amount of bone in contact with the implant, (*b*) cortical bone thickness, and (*c*) surrounding bone material properties. The same approach was also applied to the case of a dental implant embedded in a bone substitute material and subject to fatigue solicitation (53), and findings showed that implant ultrasonic response is sensitive to fatigue time as well. More recently, QUS techniques were adapted to estimate dental implant primary stability in bovine bone (82). **Figure 3** shows the ultrasonic setup used for the two aforementioned in vitro studies (80, 53).

In order to investigate the effect of bone healing on the ultrasonic response of the bone–implant interface, coin-shaped implants described in more detail in Sections 4.3 and 5.4 have been

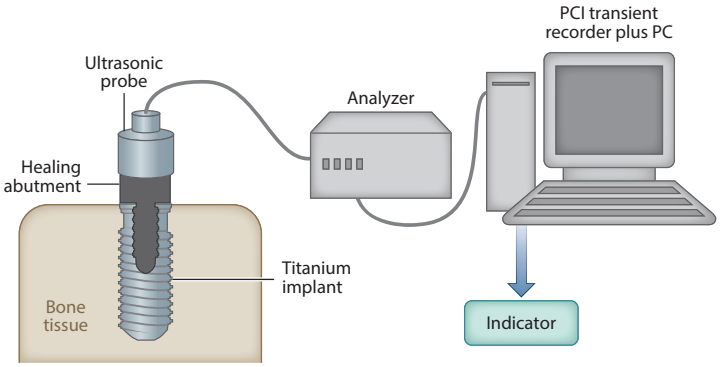


Figure 3

Schematic description of the ultrasonic experimental setup used for implant stability estimation (53, 80). The transducer is placed on the emerging part of the implant, and measurement is carried out using appropriate electronics (analyzer and signal processing unit). Abbreviation: PCI, peripheral component interconnect.

employed. Changes in the planar bone–implant interface's ultrasonic response can be measured in controlled conditions. One study measured the ultrasound response of the interface in vitro after 7 and 13 weeks of healing time (83) and found a significant difference between both healing durations. All these studies show the potential for QUS techniques to investigate bone quality around an implant. However, further work is necessary in order to improve the reproducibility of the measurements and to consider dental implants under in vivo conditions.

5.3. Torque Test

5.3.1. Insertion torque test. An high degree of insertion torque (IT) usually suggests that an implant is firmly inserted into bone. The IT test derives from cutting resistance analysis, which measures the work required to cut off a unit of bone during implant surgery; this energy appears to be significantly correlated with bone density (84). Generally, a torque gauge with a readout in N·cm is incorporated within the implant delivery tool (low-speed handpiece or torque wrench). The recording allows the surgeon to differentiate and quantify bone density during implant placement (85). The IT test provides an objective assessment of bone density, which correlates well with other findings (e.g., RFA/ISQ value and histologic bone volume) (86). A recent study showed that there was no implant failure when an implant was inserted with a torque higher than 35 N·cm. By contrast, 14% failed when the implant was inserted with 25–35-N·cm torque (87). One major limitation of the IT test is that it cannot establish bone quality (which is defined by the bone microstructure, density, and mechanical properties) at the implant site. The test also cannot provide data on the amount of bone loss around an implant. Furthermore, assessment of long-term implant stability cannot be performed because it is a one-time measurement at the time of implant surgery.

5.3.2. Reverse torque test. Unlike the IT test, the reverse torque (RT) test measures the critical torque threshold at which bone-to-implant contact (BIC) is destroyed (88). This test indirectly provides information on the degree of BIC of a given implant. The RT value has been correlated with histologic assessments in animal studies (89). Clinically integrated implants in humans had a RT value of 45 to 48 N·cm (90). One study reported that no implant could be removed during abutment connection when the RT was 20 N·cm (90). Unfortunately, the RT test is too destructive to be used as a diagnostic tool (91). In addition, the arbitrary 20-N·cm threshold used

to indicate successful osseointegration has yet to be supported by scientific data. Furthermore, many other factors can affect the RT value, and they may not be easily controlled in clinic. RT value, like IT value, can provide only information such as whether osseointegration was successful or failed, without quantifying degrees of osseointegration. As such, the RT test is used mainly in research.

5.4. Debonding the Bone–Implant Interface

The different torque tests described above remain of limited interest for understanding the basic biomechanical phenomena responsible for implant stability because they use real implants with complex geometry, which does not allow (*a*) carrying out biomechanical testing under controlled conditions or (*b*) retrieving quantitative information on the adhesion of the bone–implant interface. The implant design influences the test results (92) and leads to spatially complex, nonuniform, multiaxial stress fields at the interface (93).

Specific implant models with a planar surface have been conceived to minimize the effects of friction and to work under standardized conditions (94). These animal models involve the use of flat coin-shaped implants placed onto cortical bone of rabbit tibia without mechanical fixation to the bone. Such models use tensile tests that are performed by applying a force perpendicular to the bone–implant interface (50, 51). However, this approach may be of limited interest for analyzing the phenomena involved in the rupture between bone and an implant because tensile tests correspond to a flat-punch configuration, which is a mechanically unstable situation (95). The measured pull-out force depends strongly on initial boundary conditions and cannot be used to retrieve the adhesion energy (or the strain energy release rate), which is the only physically meaningful parameter from a mechanical point of view. For these reasons, investigators developed an experimental approach, based on a mode III cleavage mechanical device, that aimed to understand the behavior of a planar bone-implant interface submitted to torsional loading (96). In mode III, the shear stress acts parallel to the plane of the crack and parallel to the crack front. Coin-shaped titanium implants were inserted on the tibiae of a New Zealand white rabbit for 7 weeks. After sacrifice, mode III cleavage experiments were performed on bone samples. An analytical model allowed the assessment of different physical quantities related to the bone–implant interface, such as torsional stiffness ($\sim$20.5 N·m·rad^{-1}), bone shear modulus ($\sim$240 MPa), maximal torsional loading ($\sim$0.056 N·m), mode III fracture energy ($\sim$77.5 N·m^{-1}), and the associated stress intensity factor (0.27 MPa·m$^{1/2}$).

6. FACTORS AFFECTING OSSEOINTEGRATION QUALITY

6.1. The Bone–Implant Interface

6.1.1. Definition of the interface. The interface includes the bone–implant boundary and a volumetric region extending out into the bone. Strain fields due to implant loading will typically radiate outward into the bone for a reasonably large distance (e.g., out to about one implant radius) (97).

Structurally, the interface of a typical osseointegrated implant is not composed of 100% bone–implant contact; it also includes implant contact with marrow-type tissue (98). This interfacial structure means that implant–bone stress transfer is determined by a variety of factors including the quantity and spatial extent of bone around the implant, the degree to which bone is bonded to the implant surface, and the biomechanical properties of the bone and bone–implant boundary.

6.1.2. Stresses and strains at the interface. The interface's main purpose is biomechanical: It supports the implant. The implant is loaded by in vivo biting forces and moments (99), which, in turn, create stresses and strains in interfacial bone, the bone–implant boundary, and the implant. Although detailed stress-strain conditions depend on many factors (see Section 6.1.1.), if interfacial stresses and strains exceed mechanical and biological limits of tissue or implant, failure will occur. This makes it necessary to analyze how the biomechanical properties of the interface affect its stress transfer and integrity. One approach is to calculate stress-strain conditions and compare them with the stress-strain limits of the materials involved—for example, the ultimate and fatigue strengths of bone and the implant. Investigators frequently take this approach (e.g., 100), but it can be problematic owing to lack of accurate data about in vivo bone properties (see Section 4.3).

6.1.3. Micromotion (relative motion) and implant stability: mechanics. Consider the mechanics of micromotion at the bone–implant boundary in the special case of no bone–implant bonding.[2] Under the same axial load, a nonbonded implant moves more than a bonded implant, as seen in the simple example shown in **Figure 2**: The axial stiffness (axial load/axial displacement) is 11.9 N/μm for the bonded case versus 10.5 N/μm for the nonbonded case. Notably, this difference exists even though the interfacial bone has the same properties in each simulation (in this finite element simulation, for bone, Young's elastic modulus $E = 10$ GPa and Poisson's ratio $\nu = 0.33$; for titanium, $E = 105$ GPa and $\nu = 0.33$). Stiffness also depends on the mechanical properties of the interfacial bone. Models like those shown in **Figure 2** show that for the same implant bonded to bone and loaded, a decrease in the bone's modulus from 10 to 1 GPa decreases the axial stiffness from 11.9 to 9.91 N/μm. Also, via in vivo measurements of axial stiffness of an implant undergoing micromotion in healing bone in mouse tibiae, it was possible to estimate an increase in the modulus of the healing interfacial tissue over 7 days, as shown in **Figure 4** (97).

6.1.4. Micromotion (relative motion) and implant stability: biology. With respect to the biological consequences of micromotion, Brånemark & Skalak's (33), in warning against having "progressive relative motion between the implant and surrounding living bone" (see Section 4.1), asserted that an increase in relative motion in vivo signals a decrease in implant stability as well as interfacial integrity. The underlying mechanism is complicated because micromotion is believed to trigger additional deterioration in interfacial biology and integrity, which, in turn, creates more micromotion—thus creating a vicious cycle.

For example, one mode of interfacial failure involves implant loosening (e.g., the implant has increased mobility or decreased stiffness) even though the implant was initially tight in the bone as a result of healing. Mechanistically, such failure is not completely understood, but one hypothesis is that it starts with fatigue microdamage (MDx) to interfacial bone due to locally high strains and stresses secondary to repetitive implant loading. The MDx then stimulates increased bone remodeling, which is bone's natural response to fatigue MDx (101, 102). In a worst-case scenario in which the rate of MDx production is large and spatially extensive, remodeling starts over a large region of bone. This creates a problem because the initial resorptive phase of remodeling (remodeling proceeds via activation-resorption-formation) transiently causes a net loss of interfacial bone over an appreciably sized region. And if implant loading continues on this decreased

[2] Earlier we noted that shear and tensile bond strengths at a typical bone–implant boundary are not large. Reported values are in the range of a few MPa—which is small in comparison with, for example, tensile bond strengths of the boundary between a dental adhesive and a biomaterial (~tens of MPa) and the tensile strength of bone itself (~100 MPa). Also, stress analyses show that the tensile stresses that develop at the bone–implant boundary can reach 1 MPa and thus overcome a putative 1-MPa tensile bond strength at the boundary (17).

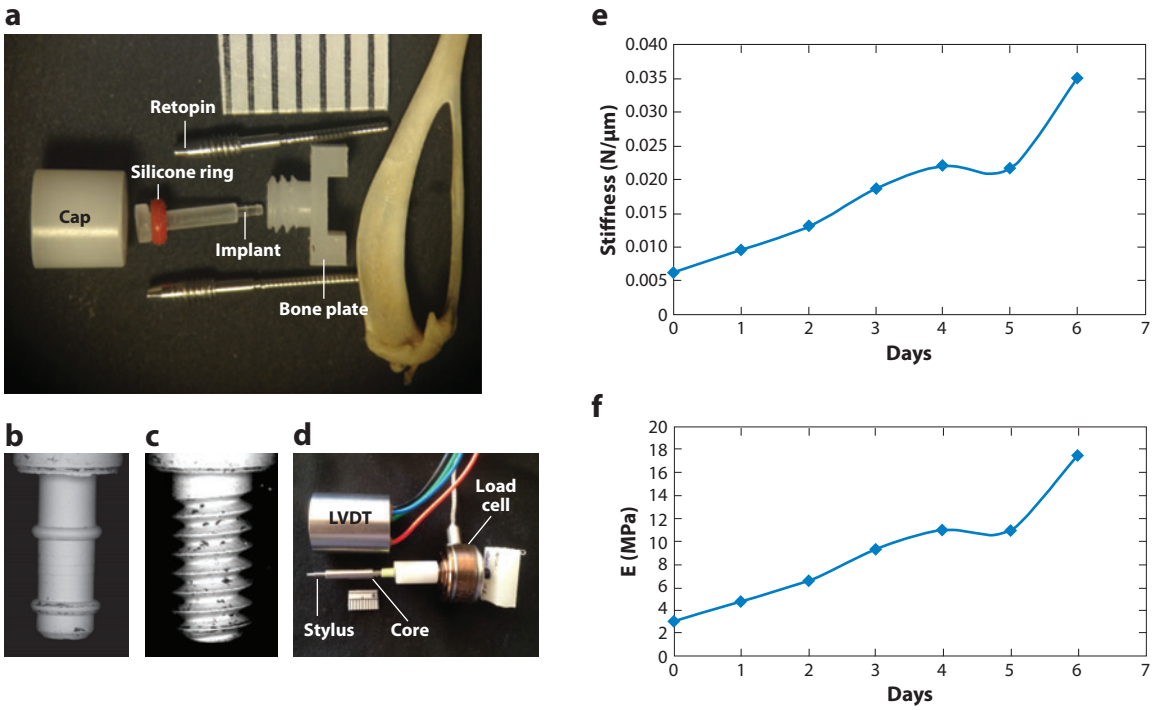


Figure 4

Excerpts from Reference 97. (*a*) Parts of the micromotion system used in a mouse tibiae model to control axial implant displacements at the implant site. (*b*,*c*) Bottom portions of the pin- and screw-shaped implants, each of which has a 0.5-mm outside diameter. (*d*) Miniature displacement and load transducers used to move implants. (*e*) Variation of the interface's axial stiffness (force on implant/implant displacement in interface) over the 7-day healing time and (*f*) values of the Young's elastic modulus of the healing interface as estimated using a FEM based on the axial stiffness values. Abbreviations: *E*, Young's elastic modulus; FEM, finite element model; LVDT, linear variable displacement transducer.

volume of bone, interfacial strains will increase, thus causing more MDx. In turn, more MDx triggers another round of bone remodeling/resorption, more loosening, and so on in a vicious cycle, until complete failure occurs.

A second type of micromotion-related failure can occur in immediate loading. It has long been known (103) that excessive micromotion during the early healing phase can block bone regeneration around any implant. In such a case, the histological result is often a thick nonmineralized fibrous tissue surrounding the implant. Clinical problems with this interfacial fibrous tissue include inflammation due to percolation of intraoral bacteria into the interface and a less-stiff interface that can lead to unequal load sharing among implants—for example, in a typical clinical case where four to six oral implants are used to support a full-arch prosthesis (104).

The mechanism underlying fibrous tissue development in such cases of excessive micromotion is unclear. Evidence indicates that fibrous tissue formation is largely independent of implant biomaterial but strongly dependent on biomechanics (105). For example, one hypothesis is that local interfacial strain arising from micromotion is a key factor affecting cellular response—that is, micromotion measured at the clinical level is important, but it is the strain state at the level of molecules, cells, etc., that regulates the biological response (106). This hypothesis resembles Perren's concepts about strain's role in long bone fracture healing (107, 108).

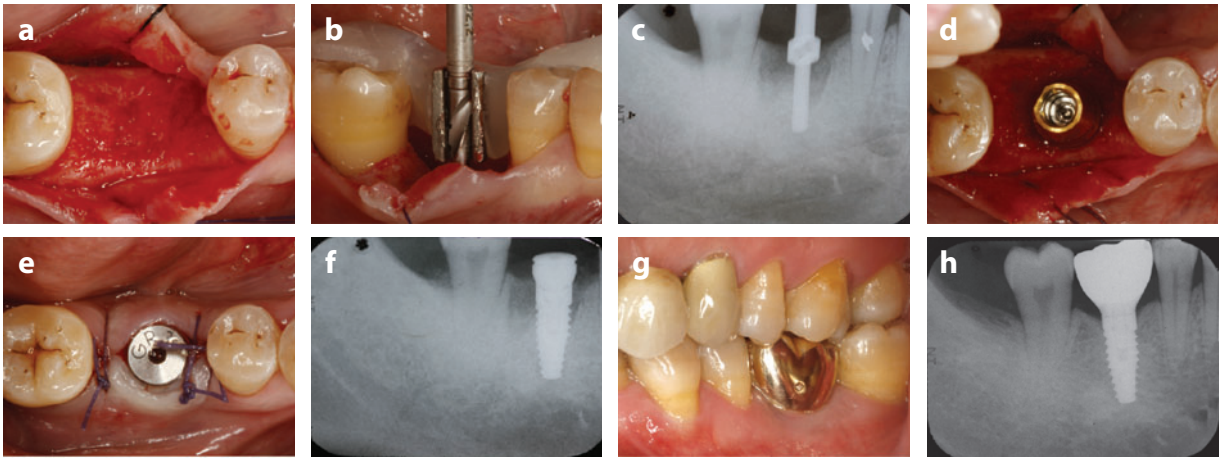


Figure 5

Case demonstration of how implant surgery is performed. (*a*) The gum flap. (*b*) After use of a round bur, a small (2.2-mm diameter) twist drill is guided into ideal implant positions by a prefabricated surgical guide (acrylic with half metal sleeve). (*c*) An X-ray is taken to check implant position with the guide pin in place. (*d*) After sequential twist drills are used to enlarge the osteotomy site, the dental implant (with fixture mount) is then placed. A countersink bur is used to enlarge the cortical aspect of the bone in order to accommodate the tapered shape of the implant. (*e*) The flap is then sutured with a healing cap in place. (*f*) A final X-ray shows the implant's placement. (*g*) The final crown with occlusion at 4 years post surgery. (*h*) A four-year follow-up radiograph shows excellent bone healing without any bone loss.

As for what constitutes excessive micromotion, Brånemark & Skalak (33) originally proposed a threshold of about 10–20 μm, but this range is probably too small. Recent work (97, 109) suggests that the threshold for deleterious micromotion is more likely in the range of about 50–250 μm. However, as noted above, it is not the clinically observed micromotion per se that is decisive; rather, it is the micromotion-related strain state at the bone–implant interface that is most relevant in triggering cellular responses. According to the recent work described above, local interfacial strain state depends on micromotion plus implant-related factors, including initial fit of the implant in the bone site, implant loading, and implant geometry/surface texture.

6.2. Surgical Procedure

6.2.1. Implant surgery. **Figure 5** illustrates how implant surgery is performed. Despite variations in implant systems, a series of instruments is used sequentially when placing dental implants. The drilling sequence starts with a round burr to mark the implant position and perforate the cortical bone. A small-diameter (~2 mm) twist drill is then used to prepare the implant osteotomy site and to guide the implant to the required depth and ideal position. A dental radiograph is taken to check the implant's position. Next, twist drills of increasing diameters are used to enlarge the osteotomy site.

Traditionally, the two-stage implant protocol is carried out to ensure that osseointegration and healing can occur in an undisturbed environment. In this approach, implants are submerged beneath the flap during the healing period (3–4 months in the mandible and 5–6 months in the maxilla, although these durations may be subjected to changes according to the clinician). One-stage implant placement can be performed when good primary implant stability is achieved and no advanced bone grafting is performed. The initial 4–8 weeks after implant placement are

then crucial to osseointegration, and caution must be taken to avoid any disturbances to the implant fixtures (110, pp. 1120–30).

6.2.2. Implant five-dimensional positioning. A successful implant therapy relies on a comprehensive examination, thorough treatment plan, and accurate implant placement. Placing implants at the ideal position is critical to avoid possible complications, injuries of anatomic structures, marginal bone loss, and compromised appearances (111). Implants should be placed in a prosthetically driven position, so that axial occlusal forces are exerted on the implant-supported restoration and fixture, thereby minimizing biomechanical complications and peri-implant marginal bone loss (112).

Generally, in the upper anterior esthetic region, an implant should be placed such that 2 mm of buccal bone is preserved in order to minimize peri-implant mucosal recession (113). In the posterior region, having 1 mm of buccal bone thickness is acceptable and the implant fixture is directed at the central groove of the adjacent posterior teeth. Between a single implant and its adjacent teeth, a minimum of 1.5 mm is needed in esthetic regions, whereas at least 3 mm is needed in other (e.g., posterior) areas (114). A recent study reported a significant relationship between the recession of distal papillae and implant-to-tooth distance. It showed that the presence of full papillae significantly dropped to 39% when the implant-to-tooth distance was less than 2.5 mm (111). For multiple implants, a minimum 3-mm distance between implants is beneficial to reduce bone loss (115). The concept of platform switching has been proposed to overcome inadequate mesiodistal spacing (116), but more studies are needed to validate this approach.

6.2.3. Factors influencing/contributing to early implant bone loss. Besides implant position, other influencing factors that may compromise the outcome of implant therapy are implant angulation (117) and soft-tissue thickness ($\geq$2 mm is adequate) (118). In addition, early implant bone loss has been linked to a number of factors and is most likely caused by surgical trauma, biological width reformation, microgap/junction effects, infection, premature occlusal overload, and crestal stress/module (37). Precautionary steps should be taken to avoid this early implant bone loss.

6.3. Factors Related to the Implant

6.3.1. Surface roughness, implant biomaterial, and surface chemistry/biochemical functionalization. For a freshly installed implant in an immediate loading scenario, stability depends on several variables besides surface-related factors, including implant initial fit, bone type, insertion torque, etc. (see Section 2.1). There appear to be no studies on whether surface factors affect implant stability in an immediate loading scenario. However, in delayed loading, Sul et al. (119) reported a positive effect of surface chemistry on ISQ values but no relationship between RFA signal and roughness [the latter finding is in agreement with others (120)]. They also suggested that increased mechanical interlocking might explain the increase in stability, as roughness affects bone-implant contact and stability months later. Similar findings come from other tests of stability in relation to roughness (e.g., 121, 122).

6.3.2. Implant geometry. Liu & Brunski (123) showed that tapered Mk IV implants had higher axial and lateral stiffness than nontapered Brånemark implants when tested in vitro in bovine bone. Others reported analogous results (124–126). Rompen et al. (127) found that although the tapered Mk IV engaged more bone than the Mk III in the cortical site, it had about the same RFA as the Mk III. Ostman et al. (128) reported "higher ISQ. . .for wide-platform versus regular or narrow platform implants" and "lower ISQ values. . .for implants in softer bone." Akca et al.

(129) reported that "bone micromorphology has a prevailing effect over implant design on. . .initial implant stability" and that insertion torque is more sensitive than ISQ in revealing biomechanical properties at the interface.

6.3.3. Misfit of the implant in its site. Different studies reveal that implant stability is affected by the size and shape of the prepared site versus those of the implant. Beer et al. (130) reported that an "adapted preparation technique" (no pretapping, minimal or no countersinking, reduced drill dimension, etc.) may improve success rates in soft bone sites—an idea also researched by others (e.g., 131, 132). However, a large misfit between implant and bone may not help stability in the long run even if there is improved stability at insertion (133) because the increased misfit causes increased compressive stress in bone (92, 134), which could increase interfacial bone microdamage (MDx) and related bone remodeling (135, 136). Nevertheless, Khayat et al. (137) installed tapered implants in undersized sites in human jaws using insertion torque greater than or equal to 70 N·cm (with some torques as high as 176 N·cm) and did not observe problems with osseointegration (in terms of marginal bone loss) 1 year later.

6.4. Factors Related to the Patient

Many factors have to be considered to ensure the long-term success of dental implants. The three major causes of implant failure are impaired host response, disruption of a weak bone–implant interface after abutment connection, and postsurgical infection (138).

6.4.1. Age. Age per se is not a contraindication for implant placement (139), but increasing age results in reduced blood supply and oxygen tension, which adversely affects osseointegration. Implant placement in adolescents, those between the ages of 10 to 19, is contraindicated (140) because it often leads to occlusal disharmony (141).

6.4.2. Medical/behavioral conditions and related treatments.

6.4.2.1. Osteoporosis and its related medications. Osteoporosis is the most common metabolic bone disease that can potentially affect bone healing around implants. Interestingly, in a systematic review of current literature (142), one group concluded that there are no data to contraindicate the use of dental implants in osteoporotic patients; however, adjustments of the surgical technique and a longer healing period may be considered in order to achieve osseointegration.

6.4.2.2. Systemic diseases. Implant surgery should be avoided in patients with systemic conditions, such as those related to recent myocardial infarction and cerebrovascular injury, valvular prosthesis surgery, immunosuppression, bleeding issues, active treatment of malignancy, drug abuse, psychiatric illness, and IV bisphosphonate use, in order to minimize postsurgery infection, implant failure, or even patient death (143). Diabetes mellitus patients are known to suffer from delayed or impaired wound healing, which may jeopardize implant stability; therefore, caution should be taken in patients with uncontrolled diabetes (HbA1c $\geq$ 7%).

Antiresorptive medications, such as bisphosphonates, have been shown to cause osteonecrosis of the jaw (ONJ) because bisphosphonates inhibit bone osteoclast precursors, promoting osteoclast apoptosis and osteoblast proliferation (144). The cumulative incidence for antiresorptive agent–induced osteonecrosis of the jaw (ARONJ) ranges from 0.8 to 12% and is significantly higher in patients using IV medications (145).

Similar concerns arise when managing patients who are undergoing radiation therapy (146) because it causes vascular changes that deplete nourishment to the bone, resulting in an osteoporotic-like condition. Moy et al. (147) reported the relative risk ratio for implant failure in radiation patients was 2.73 over 11 years.

6.4.2.3. Medical behavioral factors. Habits such as smoking and substance abuse (drugs and alcohol) can also lead to impaired wound healing thus increasing the risk of infection and peri-implantitis. There are conflicting reports about whether smoking affects implant osseointegration: Early studies often showed positive association (148), whereas recent studies have failed to identify a link (149). This is most likely attributed to different implant surfaces used, as smoking has a much bigger effect on pure titanium implants than on rough surface-coated implants. It may be unwise to place implants in patients who smoke. In a 5-year study, Weyant found that abuse of alcohol was a risk factor for implant healing and eventual failure (150). However, more studies in this area are needed.

7. CONCLUSION

Dental implants have become a common treatment modality for replacing missing teeth. This article explains how the bone and dental implant interact and provides an overview of current methods used to detect implant stability. There have been attempts to correlate the results of various stability tests with more fundamental biomechanical aspects of the supporting interface, such as elastic modulus of bone, percentage bone–implant contact (BIC), spatial location of BIC, density of bone, degree of bone–implant bonding, etc. On one hand, it is worthwhile to seek such correlations because they help define how the bone supports the implant. On the other hand, such data are of limited use in determining when—and with how much force—we can load an implant. To find this answer, more information is needed than is currently provided by existing stability tests. That is, the load-bearing capacity of an interface is ultimately governed by strength properties of the interfacial bone (e.g., yield strength, compressive or fatigue strength, etc.), and noninvasive stability tests, at best, provide information about the elastic properties of bone at the interface. In the future, perhaps some type of appropriate stability testing will contribute to a noninvasive method for predicting the ultimate load-bearing capacity of an interface.

DISCLOSURE STATEMENT

The authors are not aware of any affiliations, memberships, funding, or financial holdings that might be perceived as affecting the objectivity of this review.

ACKNOWLEDGMENTS

G.H. acknowledges the support of the French National Research Agency (ANR) through the EMERGENCE program (project WaveImplant No. ANR-11-EMMA-039). G.H. also acknowledges the help of Vincent Mathieu, Romain Vayron, and Adrien Michel. H.L.W. acknowledges help from Dr. Jia-Hui Fu (assistant professor at National University of Singapore) and Fernando Suarez (graduate student at University of Michigan, School of Dentistry). J.B. acknowledges support of NIH-NIBIB grant R01EB000504 and many useful discussions with coworkers Drs. Jill A. Helms (Stanford University) and Antonio Nanci (University of Montreal).

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