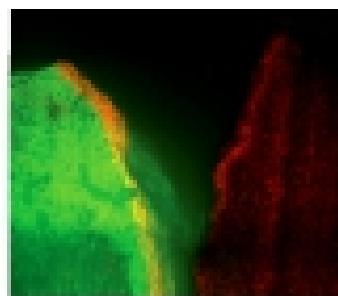
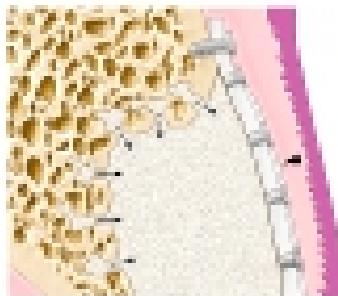
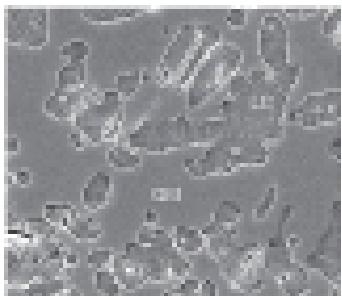




# Craig's RESTORATIVE DENTAL MATERIALS

FIFTEENTH EDITION



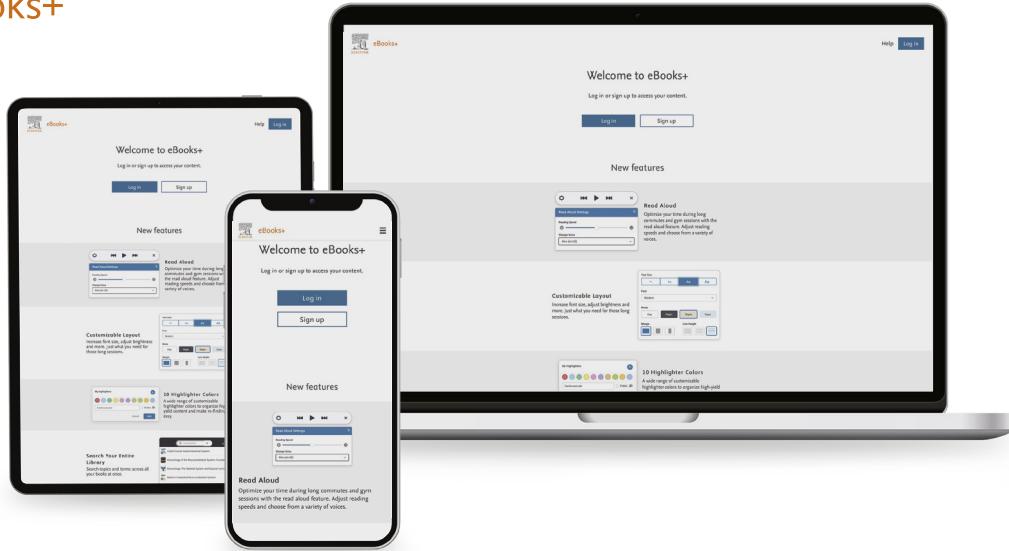
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# Craig's RESTORATIVE DENTAL MATERIALS

FIFTEENTH EDITION

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*To the many mentors, mentees, and colleagues  
with whom we have collaborated*

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# Preface

The 15th edition of this classic textbook has been extensively updated to include many recent developments in dental biomaterials science and new materials for clinical use. The sequencing of the materials has also been streamlined. The book continues to be designed for predoctoral dental students and also provides an excellent update of dental biomaterials science and clinical applications of restorative materials for students in graduate programs and residencies.

Dr. Carmem S. Pfeifer takes on the lead editor role from Dr. Ronald L. Sakaguchi, who has served in that capacity for the 13th and 14th editions and now serves as coeditor. Dr. Pfeifer is professor and division head of Biomaterial and Biomedical Sciences, Department of Oral Rehabilitation and Biosciences at the School of Dentistry, Oregon Health & Science University (OHSU), with a joint appointment in the Department of Biomedical Engineering, School of Medicine, in Portland, Oregon. She earned her DDS and PhD in Dental Materials Sciences from the University of São Paulo, Brazil.

Dr. Sakaguchi serves as Dean of the School of Dentistry at OHSU. He earned a BS in cybernetics from the University of California Los Angeles (UCLA), a DDS from Northwestern University, an MS in prosthodontics from the University of Minnesota, a PhD in biomaterials and biomechanics from Thames Polytechnic (London, England; now the University of Greenwich), and an MBA in entrepreneurship from Babson College.

Dr. Jack L. Ferracane returns as coeditor of the 15th edition. Dr. Ferracane serves as professor and chair of the Department of Oral Rehabilitation and Biosciences at the School of Dentistry (OHSU) in Portland, Oregon. He earned a BS in biology from the University of Illinois and an MS and a PhD in biological materials from Northwestern University.

We thank our many chapter authors for their effort and expertise in revising the text from the previous edition: Dr. Hong-seok An, Dr. Juliana Branco da Costa, and Dr. Justin Merritt from Oregon Health & Science University; Dr. Roberto R. Braga from the University of São Paulo; Dr. Jason Alan Griggs from the University of Mississippi; Dr. John C. Mitchell from Midwestern University; Dr. Danielle P. Wingrove from the University of Utah; and Dr. Yu Zhang from the University of Pennsylvania.

The organization of the 15th edition was updated so that each chapter focuses on a specific class of materials. The chapter on testing of dental materials and biomechanics is now provided as online content only. One new chapter dedicated to dental adhesives was created ([Chapter 7](#)), encompassing the new adhesive technologies. The chapter on dental composites ([Chapter 8](#)) has been extensively revised to include newer technology in photopolymerization. [Chapter 13](#), on technology, has also seen a major update, with a focus not only on CAD-CAM but also on additive manufacturing technologies applied to dentistry.

An enhanced ebook version, included with every new print purchase, is available for this textbook. Included is the majority of the procedural, or materials handling, content that was in the previous editions. In addition, newly produced interactive videos for the most common procedures in dentistry are available at <http://ebooks.health.elsevier.com/>. The Evolve Resources website, available for instructors at <http://evolve.elsevier.com/Pfeifer/restorative/>, includes an image collection and PowerPoint lecture slides to supplement the print version of the book.

**Carmem Pfeifer**

**Jack Ferracane**

**Ronald Sakaguchi**

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We are deeply grateful to Lauren Boyle and Kelly Skelton, Senior Content Strategists at Elsevier, for guidance in the initial planning and approval of the project; to Sudha Satwa Sen, Senior Developmental Editor at Elsevier, for the many suggestions and support throughout the design process and writing of the manuscript.

Thanks also to many others at Elsevier for their behind-the-scenes work and contributions to the book.

Lastly, we thank our colleagues in our respective institutions for the many informal chats and suggestions offered and our families who put up with us being at our computers late in the evenings and on many weekends. It truly does take a community to create a work like this textbook, and we thank you all.

**Carmem Pfeifer  
Jack Ferracane  
Ronald Sakaguchi**

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# Role and Significance of Restorative Dental Materials

RONALD SAKAGUCHI

## CHAPTER OUTLINE

**Scope of Materials Covered in Restorative Dentistry**

**A Systems Approach to Restorative Materials**

**Application of Various Sciences**

**Future Developments in Biomaterials**

Developments in materials science, tissue engineering, regenerative dentistry, imaging and subtractive computer-aided design/computer-aided manufacturing (CAD/CAM), and additive (3D printing) manufacturing have dramatically changed the way we look at the replacement of components of the human anatomy. The replacement of tooth structure lost to disease, wear, and injury continues to be a large part of general dental practice. Restorative dental materials are the foundation for the replacement of tooth structure.

Form and function are important considerations in the replacement of lost tooth structure. Although tooth form and appearance are aspects most easily recognized, the function of the teeth and supporting tissues contributes greatly to the quality of life. The links between oral and general health are well known. Proper function of the elements of the oral cavity, including the teeth and soft tissues, is needed for eating, speaking, swallowing, proper breathing, and smiling.

Restorative dental materials make the reconstruction of the dental hard tissues possible. In many areas, the development of dental materials has progressed more rapidly than for other anatomical prostheses. Because of their long-term success, patients often expect dental prostheses to outperform the natural materials and structures they replace. The application of materials science is unique in dentistry because of the complexity of the oral cavity, which includes bacteria, high forces, ever-changing pH, and a warm, fluid environment. The oral cavity is considered to be the harshest environment for a material in the body. In addition, when dental materials are placed directly into tooth cavities as restorative materials, there are very specific requirements for manipulation of the material. Knowledge of materials science and biomechanics is very important when choosing materials for specific dental applications and when designing the best solution for restoration of tooth structure and replacement of teeth.

A review of the history of dentistry may be found on the book's website at <http://evolve.elsevier.com/sakaguchi/restorative>.

## Scope of Materials Covered in Restorative Dentistry

Restorative dental materials include representatives from the broad classes of materials: metals, polymers, ceramics, and composites. Dental materials include such items as resin composites, ceramics, cement, glass ionomers, metals, gypsum materials, impression materials, denture base resins, casting investments, dental amalgams, and other materials used in restorative procedures. The requirements for material characteristics and performance range from high flexibility required by impression materials to high stiffness required in crowns and fixed dental prostheses. Materials for dental implants require integration with bone. Some materials are cast to achieve excellent adaptation to existing tooth structures, whereas others are machined under digital control to produce very reproducible dimensions and structured geometries. When describing these materials, physical and chemical characteristics are often used as criteria for comparison. To understand how a material works, we study its chemical structure, its physical and mechanical characteristics, and its manipulation to result in the best performance.

Most restorative materials are characterized by physical, chemical, and mechanical parameters that are derived from test data. Improvements in these characteristics might be shown in laboratory studies, but the real test is the material's performance in function in the mouth and the ability of the material to be manipulated properly by the dental team. In many cases, errors and variations in handling and manipulation can negate the improvements in physical, chemical, and mechanical properties. Therefore it is very important for the dental team to understand how to manipulate dental materials appropriately.

## A Systems Approach to Restorative Materials

The practice of clinical dentistry requires a complete understanding of the various clinical techniques and a solid foundation of knowledge of the biological, chemical, and physical principles of human anatomy and function. It is important to understand the "how" and "why" associated with the function of natural and synthetic dental materials. The best patient outcomes will be achieved when a systems approach is used to assess the chemical, physical, and mechanical aspects of dental materials and oral function, together with the physiological, structural, and other biological properties of the tissues that support the restorative and

rehabilitative constructs. This integrative approach, when combined with the best available scientific evidence, clinician experience, patient preferences, and patient modifiers, results in the best patient-centered care.

## Application of Various Sciences

In the chapters that follow, fundamental characteristics of materials are presented, along with numerous practical examples of how the basic principles relate to clinical applications. Test procedures and fabrication techniques are discussed briefly but not emphasized. Many of the details of manipulation are found on the book's website at <http://evolve.elsevier.com/sakaguchi/restorative>.

Knowledge and application of fundamental principles of materials and mechanics are essential for the design and optimal prognosis of restorations. For example, the prognosis of long-span fixed dental prostheses, or bridges, is dependent on the stiffness and fracture resistance of the materials. When considering aesthetics, the hardness of the material is an important property because it influences the ability to polish the material. Some materials release fluoride when exposed to saliva, which might be beneficial in high-caries-risk patients. When selecting a ceramic for CAD-CAM fabrication of an all-ceramic crown, the machining and wear characteristics of the ceramic are important. For implants, surface texture, coating, and geometry are critical considerations for bone and soft tissue adaptation. These are just a few examples of the many interactions between the clinical performance of dental materials and fundamental scientific principles.

## Future Developments in Biomaterials

The 2021 Global Burden of Disease Study reported that among the 371 diseases and injuries assessed, oral disorders were the most prevalent Level 3 diseases: dental caries in permanent teeth (#1), periodontal diseases (#8), dental caries in deciduous teeth (#14), and edentulism (#20) (GBD 2021 Diseases and Injuries Collaborators 2021). These oral diseases and conditions affect about 3.69 billion people worldwide.

In the United States, about 50% of adults aged 20 to 64 have lost at least one permanent tooth to an accident, periodontal disease, a failed root canal, or tooth decay. In adults aged 65 and older, 13% have lost all of their natural teeth. That number is twice as large for adults aged 75 and over (18%) than for adults aged 65 to 74 (9%) (CDC/NCHS, National Health and Nutrition Examination Survey). For children aged 5 to 19 years, 13% have untreated dental caries. For adults aged 20 to 44, that number is 26%.

The demand for restorative care continues to be high, although there are many inequities in the dental care available to and utilized by marginalized populations. For some populations, there has been a shift from removable prostheses to implant-supported, fixed prostheses. For single-tooth loss, implants enable restoration of single crowns rather than multiunit, longer-span restorations. Research into implant coatings, surface textures, graded properties, alternative materials, and new geometries will continue to grow. These advances will improve the viability of implants, with enhancements to bone and soft tissue health.

Dental and orofacial aesthetics will continue to be a focus for some consumers, which will promote the development and sales of tooth-whitening systems, Botox, and aesthetic restorations. A more natural-looking appearance with character is preferred by many over a uniform, dazzling white dentition, which was the trend of the last decade. There will be more demand for materials that mimic natural dentition and provide the same depth of color and optical characteristics as natural teeth.

With the aging of the population, exposed root surfaces, dry mouth, and worn dentitions will be more common. These are challenging conditions to restore, and materials will need to function in an environment with reduced salivary flow and atypical salivary pH and chemistry. This population will be managing multiple chronic illnesses with many medications and will have limitations in maintaining adequate oral home care. Restorative materials will be challenged in this difficult environment.

Advances in tissue regeneration are accelerating. Our understanding of the oral microbiome is expanding. Biofabrication and 3D bioprinting methods are creating new and more natural structures and materials. This is a very exciting time for materials research, and clinicians will have much to look forward to in the near future as this body of research develops new materials for expanded clinical applications.

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# 2

# The Oral Environment

JUSTIN MERRITT

## CHAPTER OUTLINE

### Enamel

#### The Mineral

#### Dentin

Physical and Mechanical Properties

The Dentin–Enamel Junction

#### Oral Biofilms and Oral Health

Early Oral Biofilm Development on Enamel

Oral Biofilm Maturation

Oral Biofilm Development on Restorative and Implant Materials

Interactions of Oral Biofilms With Common Restorative Materials

Interactions of Oral Biofilms With Denture and Implant Materials

Caries Prevention

knowledge of the composition, structure, and properties of the adhesive tissue substrates.

### Enamel

[Fig. 2.1](#) shows a schematic diagram of a posterior tooth sectioned to reveal the enamel and dentin components. Enamel forms the hard outer shell of the crown and, as the most highly calcified tissue, is well suited to resisting wear due to mastication.

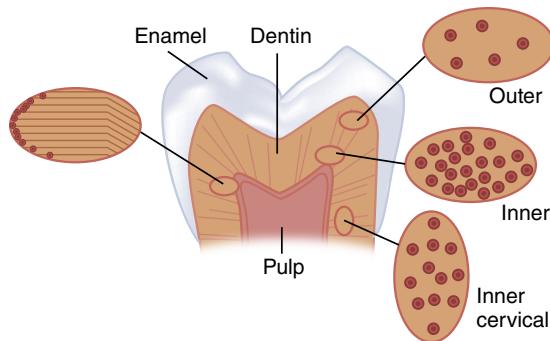
Enamel is formed by ameloblasts starting at the DEJ and proceeding outward to the tooth surface. The ameloblasts exchange signals with odontoblasts located on the other side of the DEJ at the start of the enamel and dentin formation, and the odontoblasts move inward from the DEJ as the ameloblasts move outward to form the enamel of the crown. Most of the enamel organic matrix, which is composed of amelogenins and enamelin, is resorbed during tooth maturation to leave a calcified tissue that is largely composed of minerals and a sparse organic matrix. The structural arrangement of enamel forms keyhole-shaped structures known as *enamel prisms* or *rods* that are about 5 µm across, as seen in [Fig. 2.2](#).

The overall composition of enamel is about 96% mineral by weight, with 1% lipid and protein and the remainder being water. The organic portion and water probably play important roles in tooth function and pathology, and it is often more useful to describe the composition on a volume basis. On that basis, we see that organic components make up about 3% and water 12% of the structure. The mineral is formed and grows into very long crystals of hexagonal shape about 40 nm across; these crystals have yet to be synthetically duplicated. There is some evidence that the crystals may span the entire enamel thickness, but this is difficult to prove because most preparation procedures lead to fracture of the individual crystallites. It appears that they are at least thousands of nanometers long. If this is true, then enamel crystals provide an extraordinary “aspect” ratio (length-to-width ratio) for a nanoscale material, and they are very different from the much smaller dentin crystals. The crystals are packed into enamel prisms or rods that are about 5 µm across, as shown in [Fig. 2.2](#). These prisms are revealed easily by acid etching, extending in a closely packed array from the DEJ to the enamel surface and lying roughly perpendicular to the DEJ, except in cuspal areas where the rods twist and cross, known as *decussation*, which may increase fracture resistance. About 100 crystals of the mineral are needed to span the diameter of a prism, and the long axes of the crystals tend to align themselves along the prism axes, as seen in [Fig. 2.2](#).

The crystals near the periphery of each prism deviate somewhat from the long axis toward the interface between prisms. The deviation in the tail of the prism is even greater. The individual crystals

The tooth contains three specialized calcified tissues: enamel, dentin, and cementum ([Fig. 2.1](#)). Enamel is unique in that it is the most highly calcified tissue in the body and contains the least organic content of any of these tissues. Enamel provides the hard outer covering of the crown that allows efficient mastication. Dentin and cementum, like bone, are vital, hydrated, biological composite structures formed mainly of a collagen type I matrix reinforced with the calcium phosphate mineral called *apatite*. Dentin forms the bulk of the tooth and is joined to the enamel at the dentin–enamel junction (DEJ). The dentin of the tooth root is covered by cementum, which provides a connection of the tooth to the alveolar bone via the periodontal ligament. Although the structure of these tissues is often described in dental texts, their properties are often discussed only superficially. However, these properties are important with regard to the interrelationships of the factors that contribute to the performance necessary for the optimum function of these tissues.

In restorative dentistry, we are interested in providing preventive treatments that will maintain tissue integrity and replace damaged tissues with materials that ideally mimic the natural appearance and performance of those tissues when necessary. Thus knowledge of the structure and properties of these tissues is desirable both as a yardstick to measure the properties and performance of restorative materials and as a guide to the development of materials that will mimic their structure and function. In addition, many applications, such as dental bonding, require us to attach synthetic materials to the calcified tissues, and these procedures rely on detailed

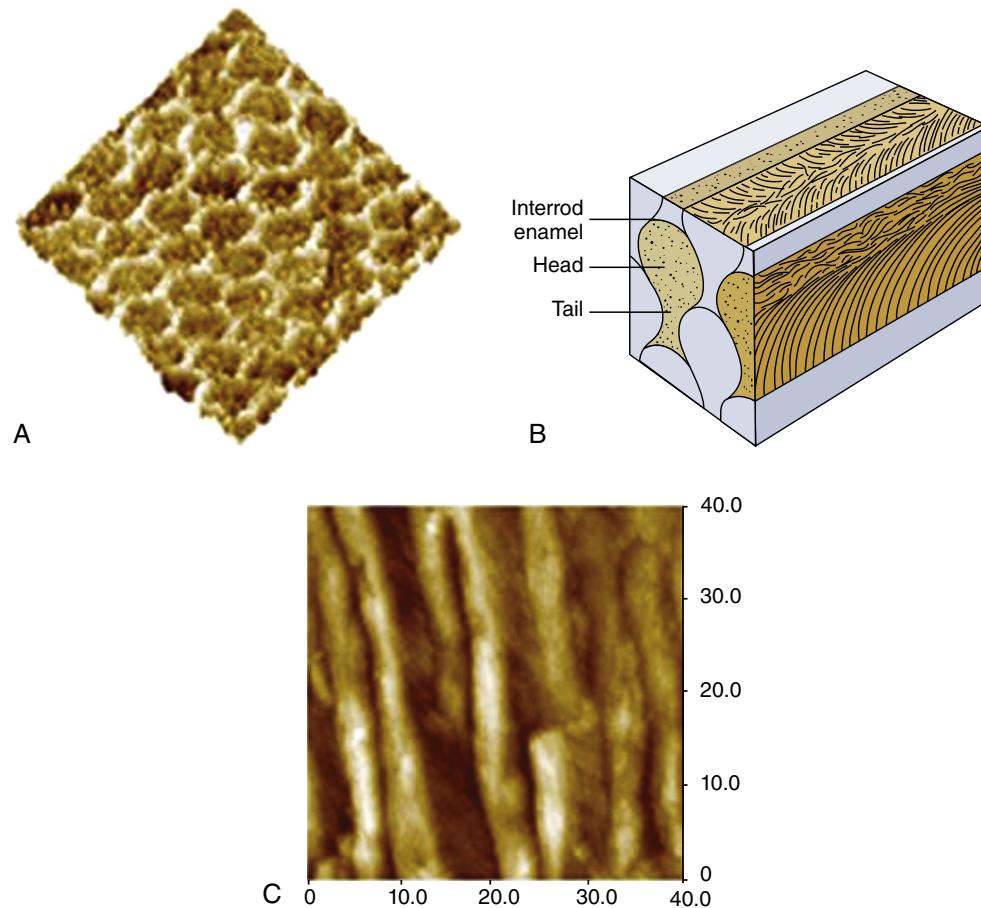


• **Fig. 2.1** Schematic diagram of a tooth cut longitudinally to expose the enamel, dentin, and the pulp chamber. On the *right* side are illustrations of dentin tubules as viewed from the top, which show the variation in the tubule number with location. On the *left* is an illustration of the change in direction of the primary dentin tubules as secondary dentin is formed. (From Marshall SJ, Balooch M, Breunig T, et al. Human dentin and the dentin-resin adhesive interface. *Acta Mater.* 1998;46:2529–2539).

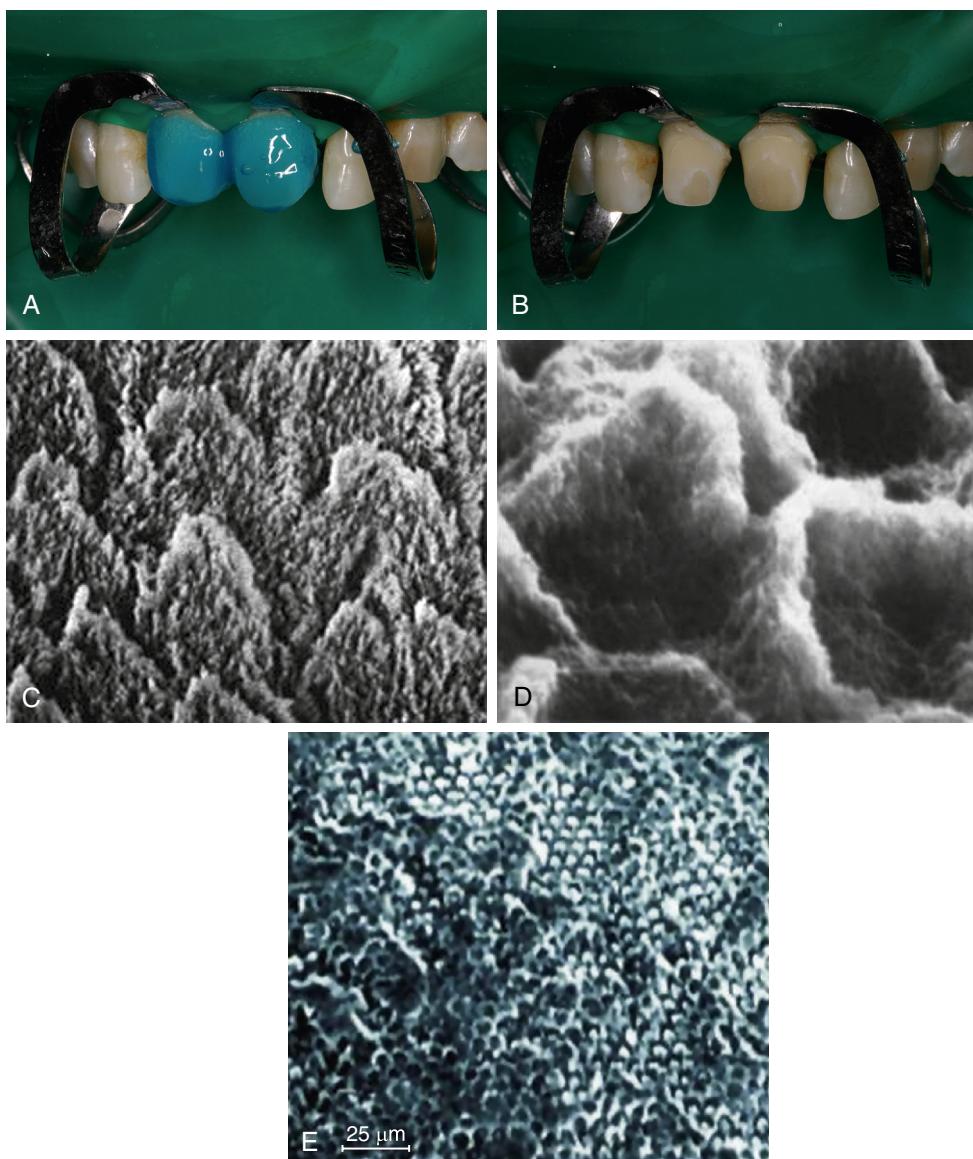
within a prism are also coated with a thin layer of lipid and/or protein that plays important roles in mineralization, although much remains to be learned about the details. Recent work suggests that this protein coat may lead to increased toughness of the enamel. The interfaces between prisms, or interrod enamel, contain the

main organic components of the structure and act as passageways for water and ionic movement. These areas, also known as *prism sheaths*, are of vital importance in etching processes associated with bonding and other demineralization processes, such as caries.

Etching of enamel with acids such as phosphoric acid, commonly used in enamel bonding, eliminates smear layers associated with cavity preparation, dissolves persisting layers of prismless enamel in deciduous teeth, and differentially dissolves enamel crystals in each prism. The pattern of etched enamel is categorized as type 1 (preferential prism core etching; Fig. 2.2A), type 2 (preferential prism periphery etching; Fig. 2.3C), and type 3 (mixed or uniform). Sometimes these patterns appear side by side on the same tooth surface (Fig. 2.3E). No differences in micro-mechanical bond strength of the different etching patterns have been established. In a standard cavity preparation for a composite, the orientation of the enamel surfaces being etched could be perpendicular to enamel prisms (perimeter of the cavity outline), oblique, cross section of the prisms (beveled occlusal or proximal margins), and axial walls of the prisms (cavity preparation walls). During the early stages of etching, when only a small amount of enamel crystal dissolution occurs, it may be difficult or impossible to detect the extent of the process. However, as the etching pattern begins to develop, the surface etched with phosphoric acid develops a frosty appearance (Fig. 2.3B), which has been used as the



• **Fig. 2.2** Enamel microstructure. (A) Atomic force microscopy images showing prism cross sections. (B) Schematic diagram of keyhole-shaped enamel prisms or rods about 5  $\mu\text{m}$  in diameter. (C) Atomic force microscopy images showing the axes of the prisms. Crystallite orientation deviates in the interrod and tail area and the organic content increases in the interrod area. (Modified from Habelitz S, Marshall SJ, Marshall GW, et al. Mechanical properties of human dental enamel on the nanometer scale. *Arch Oral Biol.* 2001;46:173–183.)



**Fig. 2.3** Etching enamel. (A) Gel etchant dispensed on the enamel portion of the preparation. (B) Frosty appearance after etching, rinsing, and drying. (C) Magnified view of etch pattern with preferential prism periphery etch (type 1). (D) Bonding agent revealed after dissolving enamel. (E) Mixed etch patterns showing type 1 (light prisms with dark periphery) and type 2 (dark cores with light periphery) etching on the same surface. (C and D, After Marshall GW, Olson LM, Lee CV. SEM Investigation of the variability of enamel surfaces after simulated clinical acid etching for pit and fissure sealants. *J Dent Res.* 1975;54:1222–1231; E, After Marshall GW, Olson LM, Lee CV. SEM investigation of the variability of enamel surfaces after simulated clinical acid etching for pit and fissure sealants. *J Dent Res.* 1975;54:1222–1231; E, from Marshall GW, Marshall SJ, Bayne SC. Restorative dental materials: scanning electron microscopy and x-ray microanalysis. *Scanning Microsc.* 1988;2:2007–2028.)

traditional clinical indicator for sufficient etching. This roughened surface provides the substrate for infiltration of bonding agents that can be polymerized after penetration of the etched enamel structure so that they form micromechanical bonds to the enamel when polymerized. With weaker acids in most self-etching bonding agents, this frosty appearance typically cannot be detected.

There are two other important structural variations of enamel. Near the DEJ, the enamel prism structure is not as well developed in the very first enamel formed, so that the enamel very close to the DEJ may appear aprismatic or without the prism-like structure. Similarly, at the completion of the enamel surface, the ameloblasts

degenerate and leave a featureless layer called *prismless enamel* on the outer surface of the crown. This layer is more often observed in deciduous teeth and is often worn off in permanent teeth. However, if present, this causes some difficulty in creating an effective etching pattern and may require roughening of the surface or additional etching treatments. The outer surface of the enamel is of great clinical significance because it is the surface subjected to daily wear and undergoes repeated cycles of demineralization and remineralization. As a result of these cycles, the composition of the enamel crystals may change, for example, as a result of exposure to fluoride. Thus the properties of the enamel might be expected



• **Fig. 6.14** Typical molar with stained fissures and no diagnosable caries. (A) Before sealing and (B) after sealing with a natural-colored sealant material. (From Hatrick CD, Eakle WS, Bird WF. *Dental Materials: Clinical Applications for Dental Assistants and Dental Hygienists*. 2nd ed. Saunders; 2011.)

Sealant materials have a variety of features that must be selected carefully by the health care provider. Most current materials are light-cured rather than self-cured because of the ease of use and controlled rate of cure. Tooth-colored or clear resins are available that are natural looking on the tooth surface, but they are also available in opaque or tinted materials to make the recall examination process easier (Fig. 6.15). Color-reversible, photosensitive sealants include photosensitive pigments that are normally colorless but change to green or pink when exposed to the dental curing light to help determine whether the sealant adequately covers the pit and fissures. The color change lasts for about 5 to 10 minutes after exposure but can be repeated at recall visits with another exposure to a dental curing light.

An increasing number of resin sealants are marketed with the claim that they release fluoride. The release is highest in the first

24 hours after placement and then tapers to a low maintenance level, which to date has not been proven to provide any significant improvement in clinical protection against caries.

### **Application of Resin-Based Pit and Fissure Sealant**

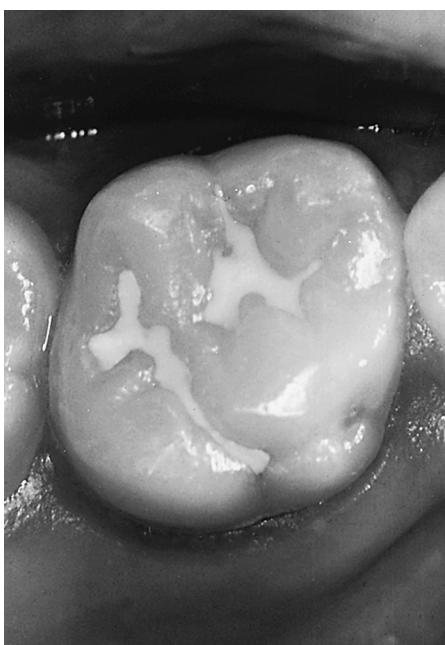
The handling characteristics of a sealant depend on the composition of the material and the surface to which it is applied. Optimal preparation of the surface will lead to close adaptation of the sealant to the tooth enamel, a strong seal against the ingress of oral fluids and debris, and long-term material retention.

The penetration of any of the sealants to the bottom of the pits and fissures and filling them without voids is important. Air or debris can be trapped in the bottom of the fissure that prevents it from being completely filled, as shown in Fig. 6.16. Control of the viscosity of the sealant is important to obtain optimum results, and sealants that are too thick and viscous will not penetrate the pits and fissures, or the etched enamel itself as well. Penetration of sealant into etched enamel, forming tags to a depth of 25 to 50 µm, is shown in Fig. 6.17.

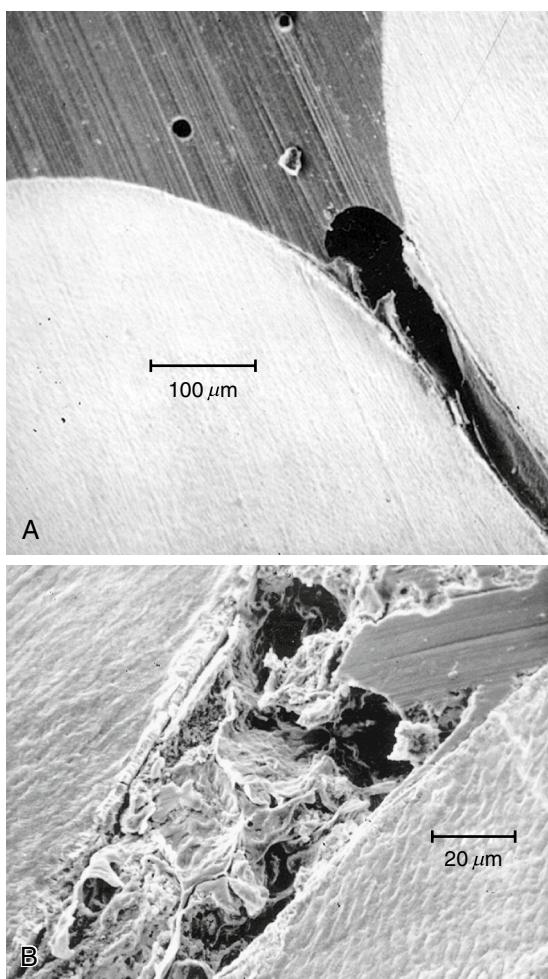
One technique that maximizes the benefits of conservative, adhesive dentistry is preventive resin restoration (PRR). PRR is indicated when some areas of the fissure system of the tooth are associated with carious dentin and others are not. Once the lesion is removed, no further preparation is performed. If the resultant preparation is restricted to narrow and shallow opening of the fissure, a resin sealant or flowable composite is placed. If additional tooth structure of the tooth is removed, a posterior dentin composite is placed in that area, and the remaining fissures and the surface of the composite restoration are sealed with resin sealant or flowable composite.

### **Glass Ionomers Sealants**

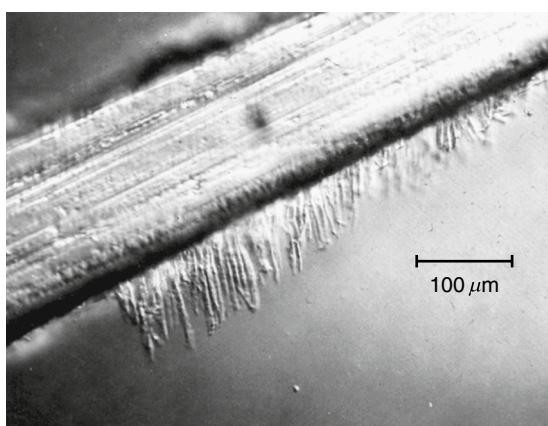
Because of their demonstrated ability to release fluoride and provide some caries protection on tooth surfaces at risk, glass ionomers have been suggested and tested for their ability to function as a fissure sealant. Glass ionomers are generally more viscous than their resin-based counterparts, which somewhat reduces their penetration to the depth of a fissure and mechanical retention to the enamel surface compared with resins. They are also more



• **Fig. 6.15** Maxillary molar tooth with opaque sealant that has been in place for 5 years.



• Fig. 6.16 Section showing a fissure incompletely filled with sealant as a result of (A) air and (B) debris. (From Gwinnett AJ. The bonding of sealants to enamel. *J Am Soc Prevent Dent.* 1973;3:21–29.)



• Fig. 6.17 Penetration of sealant into etched enamel. These tags are responsible for the bonding to enamel. (From Gwinnett AJ. The bonding of sealants to enamel. *J Am Soc Prevent Dent.* 1973;3:21–29.)

brittle and less resistant to occlusal wear. However, clinical studies using various formulations of glass ionomers have shown that, in spite of the significantly lower retention rates than resin sealants, glass ionomer-based sealants have a much greater fluoride deposition in the enamel surfaces. Thus there is a greater potential for latent caries protection after sealant loss. In summary, although glass ionomer retention appears to be poorer than for resins, clinical outcomes in terms of caries recurrence are similar if not better than with resins.

In areas where high-risk children do not have access to definitive treatment, a conservative caries management technique can seal remaining caries in a fluoride-rich environment and establish some degree of remineralization. Atraumatic restorative treatments (ARTs) involve opening a lesion, removing soft surface decay with hand instruments, and filling or sealing the surface with highly filled glass ionomer with a fast setting time. Two-year survival rates of single- and multiple-surface ART restorations in primary teeth have been reported as 93% and 62%, respectively. Five-year survival rate of single-surface ART restorations in permanent teeth was reported as 80%.

### Flowable Composites as Sealants

Low-viscosity composites referred to as *flowable composites* are marketed for a wide variety of applications, such as PRR, cavity liners, restoration repairs, and cervical restorations. The properties of flowable composites are described in Chapter 8.

Flowable composites are usually packaged in syringes or in capsules (Fig. 6.18) for direct application to the pit or fissure. As with lower-viscosity resin sealants, trapping of air in the sealant must be avoided. Because they have higher filler content than most resin sealants, flowable composites should have better wear resistance. Flowable composites appear to provide good retention and caries resistance after 24 months.

When flowable composites are used as preventive restorations, their low viscosity is a benefit in extending the restoration into adjacent fissures as a sealant. A 24-month study showed the retention and caries incidence of a flowable composite to be equivalent to that of a fluoride-containing fissure sealant.

### Clinical Outcomes Using Different Types of Sealants

Many clinical studies using Bis-GMA-based resins have been documented. In earlier studies on effectiveness of treatment with



• Fig. 6.18 Flowable composite resin in syringe delivery system and capsule. (Courtesy of Solventum, San Diego, California, United States.)



• Fig. 8.10 A bulk-fill composite. (Filtek, Courtesy 3M Company, St. Paul, MN.)

Although the idea of bulk fill is not new, and several materials have been available over the years, recently, there has been a surge of interest in these materials. A host of new products, including both flowable and high-viscosity pastes, have become available from various manufacturers. There are several approaches that the manufacturers use in an attempt to achieve greater depth of cure through the bulk of the material. The first one is to use either lower filler concentration or fillers with a refractive index that more closely matches that of the organic matrix, both aiming to facilitate light penetration. Because of the translucency limitation, these materials are available only in a limited range of shades. Another strategy is to use initiators that are capable of more efficiently reacting even as light intensity decreases with depth, one such example being a product incorporating a new germanium initiator to enable greater cure depths. These composites are recommended for use primarily in class 1 and 2 cavity preparations, although their use is not contraindicated for other classes as long as there is adequate shade match with the surrounding tooth (see Table 8.1).

A number of approaches have been taken to control polymerization shrinkage stress as well as provide good marginal adaptation to the axial walls and floor of the cavity due to bulk filling. In one approach, a rheology modifier that responds to sonic energy has been incorporated into the composite paste. This allows for a decrease in viscosity of the initially thick paste to provide good adaptation to the cavity surfaces. In other approaches, materials have been formulated to include modulators of the polymerization reaction and thus relieve shrinkage stress. These include the incorporation of a light-sensitive monomer that cleaves when exposed to light, the addition of an addition-fragmentation monomer that provides network rearrangement during polymerization and leads to stress relaxation, and the addition of special fillers with low elastic modulus to absorb stresses during polymerization.

The flowable bulk-fill composites appear to have better curing efficiency than the high-viscosity ones. Typical physical properties are listed in Table 8.3. In general, mechanical and physical properties cured in larger depths appear to be within normal range of incrementally placed traditional composites. In addition, important properties include greater depth of cure compared to

traditional composites. In several cases, the measured volumetric shrinkage may be similar to traditional composites, but the shrinkage stress is lower. There is increasing clinical performance data on the recent bulk-fill composites and those corroborate the results reported from *in vitro* studies. The quality of marginal adaptation seems to be comparable to incrementally placed composites, although a few studies have shown equivocal results for bulk-fill materials. A few long-term clinical studies have become available, and several systematic reviews and meta-analysis studies seem to indicate that these materials have similar performance to their conventional counterparts. However, the ultimate clinical effectiveness of these newer bulk-fill composite materials will become apparent as more high-quality evidence clinical studies are published.

## Flowable Composites

These light-activated, low-viscosity composites are recommended for cervical lesions, restorations in deciduous teeth, and other small, low- or nonstress-bearing restorations (see Table 8.1). In much of the older dental literature, these are also referred to as flowable composites. Typically, they are delivered through a needle tip attached to the head of the syringe. In some instances, they are also used as pit and fissure sealants. These composites contain dimethacrylate resin and inorganic fillers with a particle size of 0.4 to 3.0  $\mu\text{m}$  and filler loading of 42% to 53% by volume (see Table 8.2). These low-viscosity composite materials typically show shear thinning (pseudoplasticity). When expressed through a syringe tip, there is a reversible structural breakdown due to disruption of the hydrogen bonds between the polymer and filler. After the material has been placed, the hydrogen bonds reform very quickly so that the material does not slump.

The newest generation of flowable composites contains nano-filler particles at a volume loading somewhat lower than universal or multipurpose composites. Recently, self-adhesive flowable composites (commonly referred to as self-adhesive flowable) have become available. Dual-cured flowable composites, in conjunction with bonding agents, have been used in the treatment of internal resorption. This technique seals the dentinal tubules and strengthens the remaining tooth structure.

Typical properties of these composites are listed in Table 8.3. Flowable composites have a low modulus of elasticity, which may make them useful in cervical abfraction areas. Because of their lower filler content, they exhibit higher polymerization shrinkage and lower wear resistance than universal composites. The viscosity of these composites allows them to be dispensed by a syringe with a needle tip for easy handling. Gentle heating of higher-viscosity composites can improve their flow and enable them to be placed as flowable composites.

## Laboratory Composites

Laboratory composites are used to fabricate dental prosthetic devices extraorally and then affixed to the tooth preparation with the aid of adhesives. Although the newest generations of direct composites have excellent properties, their use can still be a challenge for large cavities with high C-factors due to concerns of polymerization stress leading to microleakage, postoperative sensitivity, poor interproximal margins, and recurrent caries. In such cases, a prefabricated and indirectly placed laboratory composite is clinically preferred. The basic composition of the laboratory composites is similar to the new generation of direct composites prior



• **Fig. 8.11** A reconstructed composite resin core prepared for a cast metal crown. (Courtesy Dr. Charles Mark Malloy, Portland, OR.)

to the polymerization step. Crowns, inlays, and veneers bonded to metal copings can be prepared with composites processed in the laboratory (see *Table 8.1*) using various combinations of light, heat, pressure, and vacuum to increase the degree of polymerization, density, mechanical properties, and wear resistance.

Typical properties of laboratory composites are listed in *Table 8.3*. Although their ultimate mechanical properties are not as high as ceramic restorations, due to their lower modulus and stress absorption capacity, these polymeric composites are often the materials of choice for certain clinical situations where a considerable amount of masticatory force is transmitted to the restoration (e.g., coronal restoration on a dental implant). For increased strength and rigidity, laboratory composites can be combined with fiber reinforcement. Restorations are usually bonded with resin cements.

### Core Build-Up Composites

If sufficient tooth structure remains to retain and support a full-coverage restoration, but extensive regions of dentin have been lost to disease, the core of the tooth can be restored before final preparation and impression. Composites are commonly used in this application. Core composites are available as self-cured, light-cured, and dual-cured products. Core composites are usually tinted (blue, white, or opaque) to provide a contrasting color with the tooth structure. Some products release fluoride. An example of a composite core build-up is shown in *Fig. 8.11*. Typical properties of core composites are listed in *Table 8.3*.

Composite cores have the following advantages as compared with amalgam: they can be bonded to dentin, can be finished immediately, are easy to contour, and can have a more natural color under ceramic restorations. Composite cores are bonded to

remaining enamel and dentin using bonding agents. A bonding agent recommended by the manufacturer of the core material should be used because some self-cured composite core materials are incompatible with some light-cured bonding agents. Retention of the final restoration should not rely on the composite structure alone because adhesion of the composite core to remaining dentin alone is insufficient to resist rotation and dislodgement of the crown.

### Provisional Composites

Provisional restorations maintain the position of the prepared tooth, seal and insulate the preparation and protect the margins, establish proper vertical dimension, aid in diagnosis and treatment planning, and help evaluate candidates for aesthetic replacements. Provisional inlays, crowns, and fixed partial dentures are usually fabricated from acrylic resins or composites. Provisional restorations fabricated from composites are generally harder, stiffer, and more color stable than those made from acrylics.

### Light-Curing Units

The most common light sources used in dentistry to photoactivate composites are quartz-tungsten-halogen (QTH) and blue light-emitting diode (LED). Definitions of terms that describe light sources used to activate dental resins are listed in *Table 8.6*.

### Quartz-Tungsten-Halogen Light-Curing Units

Quartz-tungsten-halogen light-curing units have a broad emission spectrum and can be used to activate polymerization of composites. The wavelength ranges covered are usually between 400 and 500 nm, with peak wavelength emission at 450 to 490 nm (there are some variations among the different commercial units). Typically, the irradiance ranges from 600 to 1000 mW/cm<sup>2</sup>, but higher-intensity QTH units are available. Some units can be controlled to provide two or three different intensities (step cure) or at a continuously increasing (ramp cure) intensity. A typical 2-mm-thick resin composite restoration requires a radiant exposure of 16 J/cm<sup>2</sup> (800 mW/cm<sup>2</sup> × 20 seconds = 16 J/cm<sup>2</sup>) for proper polymerization.

A QTH light source consists of a broad-spectrum light bulb (typically 75 W), several filters, a reflector, a fan, a power supply, and a light guide. The broad-spectrum output of the QTH bulb is clipped by a blue bandpass filter that only allows a narrow band of wavelengths centered around 470 to 480 nm (blue wavelengths) to be transferred to the light guide. A UV filter blocks the passage of UV wavelengths. A dichroic reflector focuses the light on the end of the light guide. The reflector also enables infrared wavelengths

**TABLE 8.6** Definitions of Terms Used to Describe Light Sources for Polymerization of Dental Resins

Term	Unit	Definition
Spectral emission	nm	Effective bandwidth of wavelengths emitted by light source
Spectral requirement	nm	Bandwidth of wavelengths required to activate photoinitiator(s) of dental resin
Flux	mW	Number of photons per second emitted by light source
Irradiance or radiant exitance	mW/cm <sup>2</sup>	Number of photons per second emitted by light source per unit area of curing tip
Energy	J <sup>a</sup>	Flux × time
Energy density	J/cm <sup>2</sup>	Radiant exitance × time

<sup>a</sup>Joule (J) = 1000 mW × s

**TABLE 8.7****Factors Causing Decrease in Intensity of Light from Quartz-Tungsten-Halogen Light-Curing Units and Maintenance Hints**

Factors	Maintenance Hints
Dust or deterioration of reflector	Clean or replace reflector
Burn-out of bulb filament	Replace bulb
Darkening/frosting of bulb	Replace bulb
Age of components	Monitor intensity, replace unit
Chipping of light tip	Replace light tip
Resin deposit on light tip	Clean or replace light tip
Change in line voltage	Get built-in voltage regulator
Lack of uniformity across light tip	Overlap curing on larger surface
Increased distance of tip from material to be cured	Keep light tip close to material

to dissipate as heat through the back of the housing. Because substantial heat is generated by the 75 W bulb, a fan is necessary to cool the bulb and assembly. Bulb life ranges from 50 to 75 hours. Some lamps produce considerable heat at the curing tip, which may produce pulpal irritation. Maintenance of QTH lights must be provided on a regular basis, as summarized in Table 8.7.

### Blue Light-Emitting Diodes

Solid-state LEDs use junctions of doped semiconductors (p-n junctions) based on gallium nitride to emit blue light. The spectral output of blue LEDs is much narrower, and there are several types on the market. All commercially available LED curing units (Fig. 8.12) have at least one chip with spectral output between 450 and 490 nm, so these units are effective for curing materials with camphorquinone photoinitiators. The first few models on the market only had this range of output and are known now as “mono-wave LEDs.” More modern LED units have multiple chips (“poly-wave LEDs”), some of which cover the 450–490 nm wavelength range and some of which cover the 390–420 nm wavelength range. The need for poly-wave LED units arose from the introduction of highly aesthetic materials. Some composite shades are very light to match teeth that have undergone bleaching, so

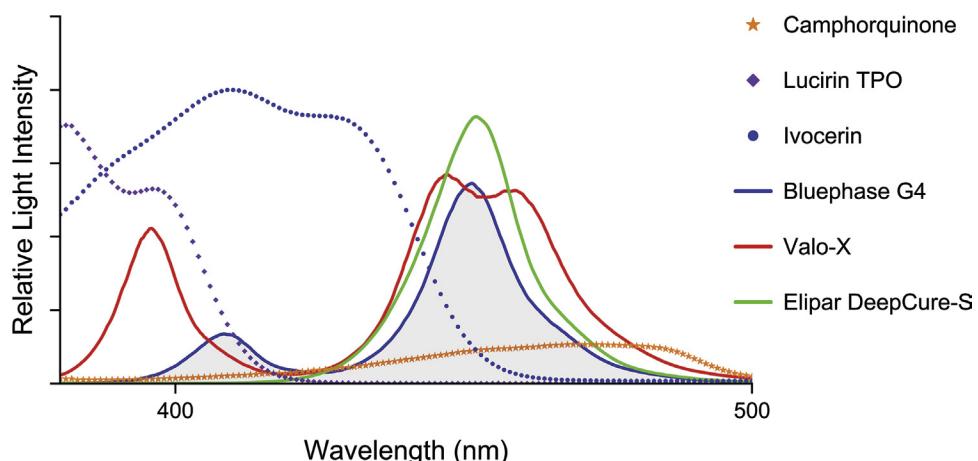


• **Fig. 8.12** LED source for photoinitiation. (Espe Elipar S10, Courtesy 3M Company, St. Paul, MN.)

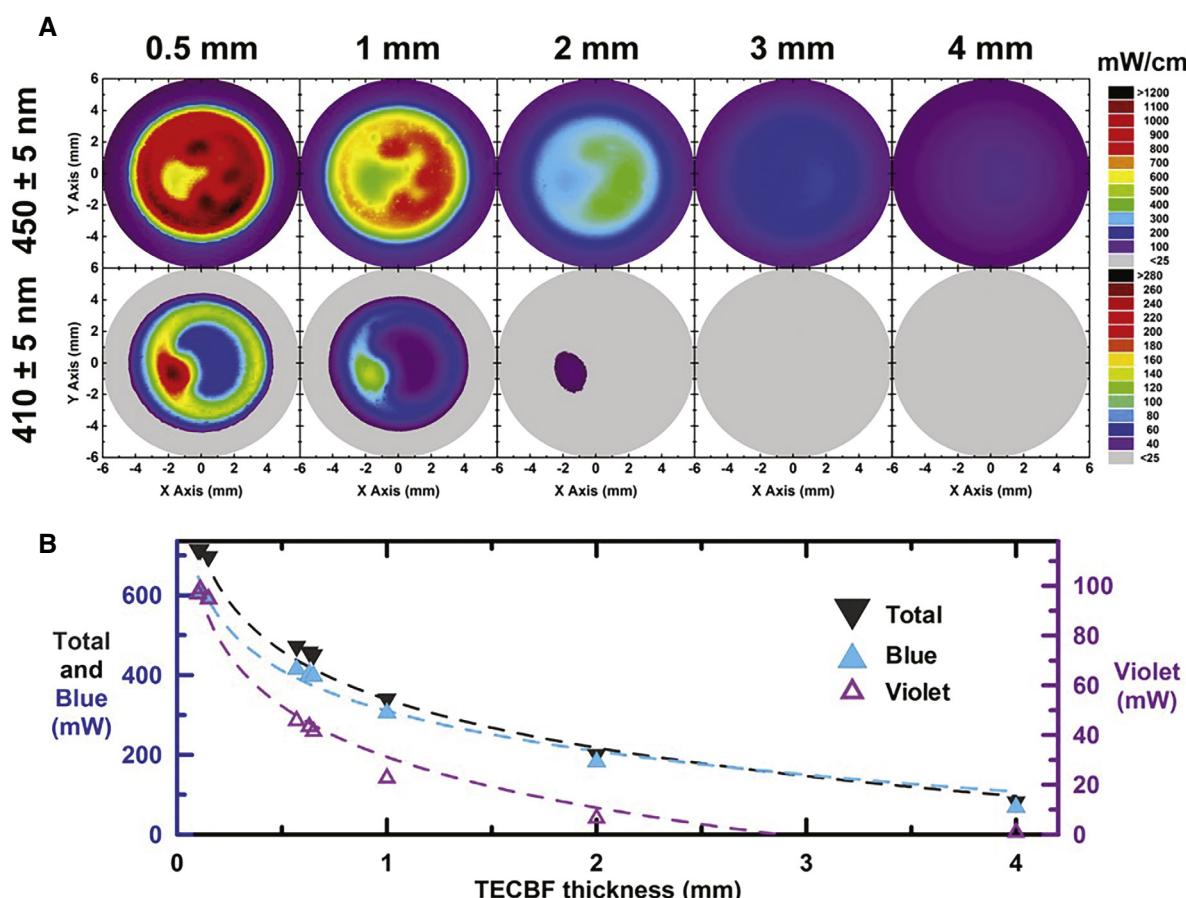
there is a need to replace part of the deep yellow camphorquinone with colorless alternative photoinitiators, such as PPD or BAPO. It is then paramount to match the spectral output of the light-curing unit and the absorbance spectrum of the photoinitiator (Fig. 8.13) to ensure proper cure of the composite. LED units do not require a filter, have a long life span, and do not produce as much heat as QTH devices. However, heat becomes a concern even with LED sources when very high irradiances are produced by the light.

It is also important to note that for poly-wave LED light-curing units, as the light penetrates through the composite, the loss of intensity is not equal for each of the different wavelengths (Fig. 8.14). The intensity of the near-UV wavelength is lower, to begin with, and it decreases more rapidly. This is especially important if thicker increments were to be used, as is the case in bulk-fill composites.

Unlike the halogen lights, the heat generated from LED curing lights is much less, which means it does not require a fan to cool it; however, the heat at the tip of the light can be high. Thus these are lighter in weight and smaller than the halogen units. Because LEDs inherently have low consumption of power, they can use rechargeable batteries, making them portable. The newer generation of LED lights has dramatically decreased in size and is lighter, more ergonomic, and easier to use. They are also high powered and emit light intensity of 1000 to 1400 mW/cm<sup>2</sup>. This feature allows more efficient and deeper curing depths. Some models are available with a “turbo” function, and can reach up to 3400 mW/cm<sup>2</sup>, which theoretically would allow for very short exposure times. However, these should be used with extreme caution because the potential for heat generation is amplified, and damage to pulpal and gingival tissues might occur. There is also



• **Fig. 8.13** Overlaying spectral output and absorbance spectra for common commercial light-curing units and initiators, respectively. Note that the single-wave LED curing units do not provide adequate spectral output for a near-UV initiator (Lucirin TPO). (Used with permission of Dr. Richard Price.)



• **Fig. 8.14** Light intensity decreases through the bulk of a composite material for a poly-wave LED curing unit. Loss of energy is much more noticeable for the near-UV wavelength, which starts out lower to begin with. (From Harlow JE, Rueggeberg FA, Labrie D, Sullivan B, Price RB. Transmission of violet and blue light through conventional (layered) and bulk cured resin-based composites. *J Dent.* 2016;53:44–50. <https://doi.org/10.1016/j.jdent.2016.06.007>.)

the concern with the extremely short exposure times because reciprocity does not always hold true. For example, the properties of a material cured at 800  $\text{mW/cm}^2$  for 20 s might be different than those of a material cured at 3200  $\text{mW/cm}^2$  for 5 s, even if both of those regimens deliver a total energy of 16  $\text{J/cm}^2$ . In general, extremely short exposure times at high intensity, other than the heat generation concern already mentioned, may also lead to inferior mechanical properties.

## Blue Light Hazards

Irrespective of the type of light-curing unit, because of the high intensity, the operator should not look directly at the tip or the reflected light from the teeth. A number of devices are marketed, including special goggles and shields, to filter the visible-light beam so the operator can directly observe the curing procedure and protect the patient and staff. The hazard from blue light exposure has been well-documented in the past several years, and there is risk of serious damage to the retina of dental care personnel.

## Prosthetic Applications of Polymers

Acrylic polymers have a wide variety of applications in restorative dentistry as denture bases, artificial teeth, denture repair materials,

impression trays, provisional restorations, and maxillofacial appliances for skeletal defects.

The vast majority of dentures made today are fabricated from heat-cured poly(methyl methacrylate) and rubber-reinforced poly(methyl methacrylate). Fractures of dentures still occur but are usually associated with carelessness or unreasonable use by the patient. Considering functional stresses, the oral environment, and expected service life, denture base materials perform remarkably well.

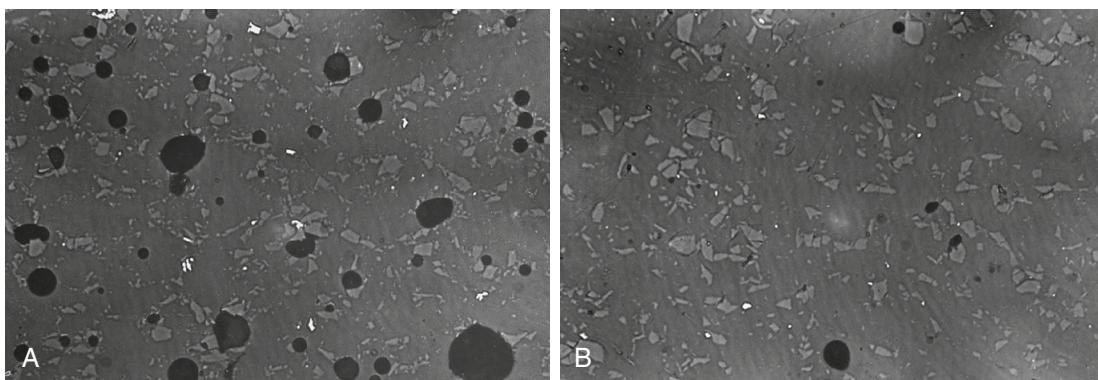
## Physical Form and Composition

Denture base plastics are commonly supplied in a powder-liquid or a gel form. The powder-liquid type may contain the materials listed in **Box 8.1**.

### Powder

Most commercial materials contain poly(methyl methacrylate), modified with small amounts of ethyl, butyl, or other alkyl methacrylates to produce a polymer somewhat more resistant to fracture by impact. The powder also contains an initiator such as benzoyl peroxide or diisobutylazonitrile to initiate the polymerization of the monomer liquid after being added to the powder.

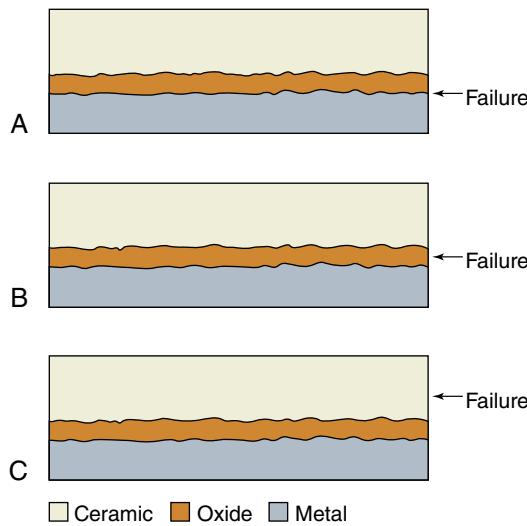
The peroxide initiator may be added to the polymer or be present as a residual from the polymerization reaction and is present in amounts from 0.5% to 1.5%.



• **Fig. 10.8** Air-fired and vacuum-fired porcelain. (A) Optical micrograph of air-fired porcelain, showing porosity. (B) Optical micrograph of vacuum-fired porcelain showing minimal porosity. (Courtesy J.O. Semelman, York, PA, 1959, Dentsply International.)



• **Fig. 10.9** View of metal-ceramic fixed dental prostheses. (Courtesy Dr. Charles Mark Malloy, Portland, OR.)



• **Fig. 10.10** Diagram showing three observed types of bond failure in metal-ceramic systems. (A) Metal-metal oxide (adhesive); (B) metal oxide-metal oxide (cohesive); and (C) ceramic-ceramic (cohesive). Note: The dimensions of the layers are not to scale.

with metal oxides on the surface of metal (Fig. 10.10) and by mechanical interlocking made possible by roughening of the metal coping.

5. CTEs of the porcelain and metal must be compatible so that the veneering porcelain never undergoes tensile stresses, which

would lead to cracking. Metal-ceramic systems are therefore designed so that the CTE of the metal is slightly higher than that of the porcelain, thus placing the veneering porcelain in compression (where it is stronger) following cooling (see Fig. 9.9). This is assuming that linear coefficients of thermal expansion of both porcelain and metal are identical to linear coefficients of thermal contraction.

6. Adequate stiffness and strength of the metal framework are especially important for FDPs and posterior crowns. High stiffness of the metal reduces tensile stresses in the porcelain by limiting deflection amplitude and deformation (strain). High strength is essential in the interproximal connector areas of FDPs.
7. High resistance to deformation at high temperatures is essential. Metal copings are relatively thin (0.4 to 0.5 mm); no distortion should occur during firing of the porcelain or the fit of the restorations would be compromised.
8. Adequate design of the restoration is critical. The preparation should provide for adequate thickness of the metal coping, as well as enough space for an adequate thickness of the porcelain to yield an aesthetic restoration. During the preparation of the metal framework, prior to porcelain application, it is important that all sharp angles be eliminated and rounded to later avoid stress concentration in the porcelain. If full porcelain coverage is not used (e.g., a metal occlusal surface), the position of the metal-ceramic junction should be located at least 1.5 mm from all centric occlusal contacts.

### Metal-Ceramic Bonding

The bond strength between porcelain and metal is an important requirement for good long-term performance of metal-ceramic restorations. In general, the bond is a result of chemisorption by diffusion between the surface oxide layer on the alloy and the porcelain. For metal alloys that do not oxidize easily, this oxide layer is formed during a special firing cycle prior to opaque porcelain application. For metal alloys that do oxidize easily, the oxide layer is formed during wetting of the alloy by the porcelain and subsequent firing cycle. The most common mechanical failure for metal-ceramic restorations is debonding of the porcelain from the metal. Many factors control metal-ceramic adhesion: the formation of strong chemical bonds, mechanical interlocking between the two materials, and thermal residual stresses. In addition, as noted earlier, the porcelain must be wet and fuse to the surface

to form a uniform interface with no voids. These factors are also important for ceramic coatings on metallic implants.

From a practical standpoint, the surface roughness at the metal-ceramic interface has a large effect on the quality of the metal-ceramic bond. Airborne particle abrasion is routinely used on metal frameworks for metal-ceramic restorations to produce a clean surface with controlled roughness. During the firing cycle, the porcelain softens, its viscosity decreases, and the porcelain first wets the metal surface before the interlocking between porcelain and metal is created. The increased area of the rough metal surface also permits the formation of a greater density of chemical bonds. The contact angle between the porcelain and metal is a measure of the wetting and, to some extent, the quality of the bond that forms. Low contact angles are indicative of good wetting. The contact angle of porcelain on a gold (Au) alloy is about 60 degrees. A scanning electron micrograph of the oxidized surface of a gold (Au)-platinum (Pt)-palladium (Pd) 98% noble alloy is shown in Fig. 9.8. However, rough surfaces can reduce adhesion if the porcelain does not wet the surface and voids are present at the interface.

The formation of an oxide layer at the surface of the metal has been shown to be the key to an adequate metal-ceramic bond. Noble metal alloys, which are resistant to oxidizing, usually have other more easily oxidized elements added, such as indium (In) and tin (Sn), to form an oxide layer and improve the bond. The oxide layer is formed during a special firing cycle prior to porcelain application. Some noble alloys containing silver have been shown to lead to porcelain discoloration or greening, explained by ionic diffusion of silver in the porcelain. Base-metal alloys contain elements, such as nickel (Ni) and chromium (Cr), that oxidize easily, and care must be taken to avoid the formation of too thick an oxide layer. Manufacturers specify firing conditions for the formation of an optimal oxide layer and often indicate the color of the oxide. Oxides rich in nickel (NiO) tend to be dark gray, whereas those rich in chromium ( $\text{Cr}_2\text{O}_3$ ) are greenish. If firing recommendations are not followed, these oxides may dissolve in the porcelain during firing, leading to discoloration visible in areas where the porcelain is thinnest; for example, near the gingival margin of the restoration. Some alloys form oxide layers rich in  $\text{Cr}_2\text{O}_3$ , which do not bond or adhere well to the alloy. These alloys typically require the application of a bonding agent to the alloy surface to modify the type of oxide formed. In some cases, manufacturers recommend an oxidation firing under reduced pressure to limit the thickness of the oxide layer. An oxidation firing in air may lead to a thicker oxide layer.

High thermal residual stresses between the metal and porcelain can lead to failure. If the metal and ceramic have largely different thermal expansion coefficients, the two materials will contract at different rates during cooling and large thermal residual stresses will form along the metal-ceramic interface. If these stresses are very high (whether tensile or compressive), the porcelain will crack and/or delaminate from the metal. Even if these stresses do not cause immediate failure, they can still weaken the bond, and lead to delayed failure. To avoid these problems, porcelains and alloys are formulated to have adequately matched thermal expansion coefficients. Most porcelains have coefficients of thermal expansion between  $13.0$  and  $14.0 \times 10^{-6}/^\circ\text{C}$ , and metals between  $13.5$  and  $14.5 \times 10^{-6}/^\circ\text{C}$ . The difference of  $0.5 \times 10^{-6}/^\circ\text{C}$  in thermal contraction between metal and porcelain causes the metal to contract slightly more than the ceramic during cooling. This condition places the porcelain under slight residual compression, which makes it less sensitive to the tensile stresses induced by mechanical loading.

A metal-ceramic bond may fail in any of three possible locations (see Fig. 10.10). Knowing the location of failure provides considerable information on the quality of the bond. The highest bond strength leads to failure within the porcelain when tested (see Fig. 10.10C); this is observed with some alloys that were properly prepared with excellent wetting by the porcelain and is also called a *cohesive failure*. Testing these high-strength specimens using the push-through shear test shows that the bond strength is approximately the same as the shear strength of the porcelain. Another possible cohesive failure is within the oxide layer (see Fig. 10.10B). Failures occurring at the interface between metal and oxide layer (see Fig. 10.10A) are called *adhesive failures* and are commonly observed with metal alloys that are resistant to forming surface oxides, such as pure gold or platinum, and exhibit poor bonding. Base-metal alloys commonly exhibit failures within the oxide layer if an excessively thick oxide layer is present.

## Ceramics for Metal-Ceramic Restorations

Ceramics for metal-ceramic restorations must fulfill five requirements: (1) they must simulate the appearance of natural teeth, (2) they must fuse at relatively low temperatures, (3) they must have thermal expansion coefficients compatible with alloys used for metal frameworks, (4) they must age well in the oral environment, and (5) they must have low abrasiveness. Porcelains are carefully formulated to achieve these requirements. These ceramics are composed of a crystalline phase (leucite) dispersed in a glassy (amorphous) matrix. Their chemical composition includes silica ( $\text{SiO}_2$ ), alumina ( $\text{Al}_2\text{O}_3$ ), sodium oxide ( $\text{Na}_2\text{O}$ ), and potassium oxide ( $\text{K}_2\text{O}$ ) (Table 10.4). Opacifiers ( $\text{TiO}_2$ ,  $\text{ZrO}_2$ ,  $\text{SnO}_2$ ), various heat-stable coloring oxides, and small amounts of fluorescing oxides ( $\text{CeO}_2$ ) are added to match the appearance of the dentin-enamel complex structure. The presence of a large amount of glassy phase in dental porcelains (80–90 vol%) permits a translucency similar to that of enamel. Coloring oxides and opacifiers allow fine tuning of the final appearance and shade control. Porcelain is supplied as a fine powder of controlled granulometry.

### Dental Porcelain

#### Composition

The quality of any ceramic depends on the choice of components, correct proportioning of each component, and control of the firing procedure. High-purity components are used in the manufacture of dental porcelains because of the stringent requirements of optical properties and chemical inertness, combined with adequate strength, toughness, and thermal expansion.

In its mineral state, feldspar, the main raw component of dental porcelains for metal-ceramic restorations, is crystalline and opaque. Chemically, it is designated as potassium aluminosilicate, with a composition of  $\text{KAlSi}_3\text{O}_8$  or  $\text{K}_2\text{O} \cdot \text{Al}_2\text{O}_3 \cdot 6\text{SiO}_2$ . Feldspar melts incongruently at about  $1150^\circ\text{C}$ , forming various silicate minerals, possibly leucite ( $\text{KAlSi}_2\text{O}_6$  or  $\text{K}_2\text{O} \cdot \text{Al}_2\text{O}_3 \cdot 4\text{SiO}_2$ ), and molten glass.

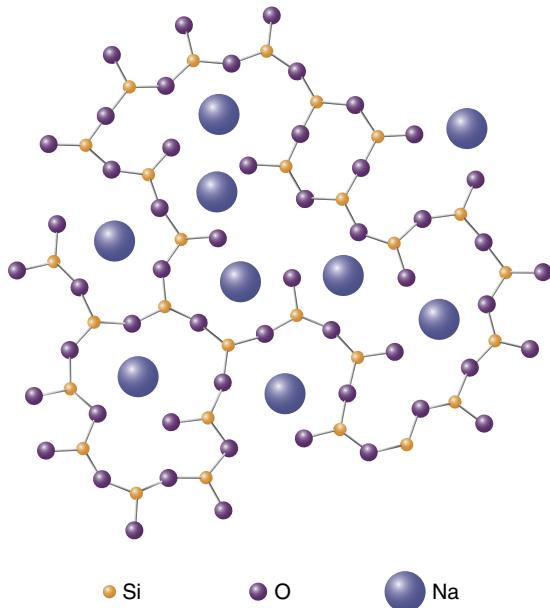
#### Manufacture

Many dental porcelain manufacturers buy feldspar as powder already screened and cleaned from impurities to their specifications. Other raw materials used in the manufacture of dental porcelains are various types of silica ( $\text{SiO}_2$ ) in the form of fine powder, alumina ( $\text{Al}_2\text{O}_3$ ), as well as alkali and alkaline earth carbonates as fluxes. During the manufacturing process, the ground components are carefully mixed together and heated to about  $1200^\circ\text{C}$  in

**TABLE 10.4** Composition of Dental Ceramics for Fusing to High-Temperature Alloys

Compound	Bident Opaque BG 2 (%)	Ceramco Opaque 60 (%)	V.M.K. Opaque 131 (%)	Bident Dentin BD 27 (%)	Ceramco Dentin T 69 (%)
$\text{SiO}_2$	52.0	55.0	52.4	56.9	62.2
$\text{Al}_2\text{O}_3$	13.55	11.65	15.15	11.80	13.40
$\text{CaO}$	—	—	—	0.61	0.98
$\text{K}_2\text{O}$	11.05	9.6	9.9	10.0	11.3
$\text{Na}_2\text{O}$	5.28	4.75	6.58	5.42	5.37
$\text{TiO}_2$	3.01	—	2.59	0.61	—
$\text{ZrO}_2$	3.22	0.16	5.16	1.46	0.34
$\text{SnO}_2$	6.4	15.0	4.9	—	0.5
$\text{Rb}_2\text{O}$	0.09	0.04	0.08	0.10	0.06
$\text{BaO}$	1.09	—	—	3.52	—
$\text{ZnO}$	—	0.26	—	—	—
$\text{UO}_3$	—	—	—	—	—
$\text{B}_2\text{O}_3$ , $\text{CO}_2$ , and $\text{H}_2\text{O}$	4.31	3.54	3.24	9.58	5.85

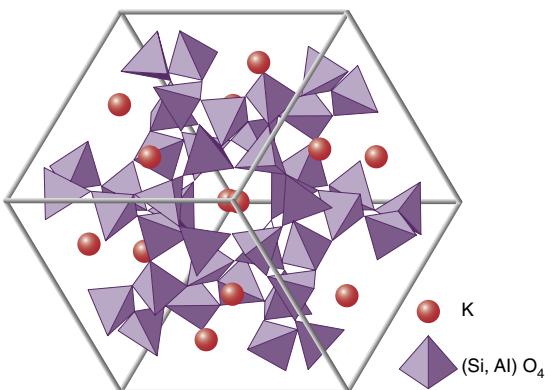
From Nally JN, Meyer JM. Experimental study on the nature of the ceramic-metallic bonding. *SSO Schweiz Monatsschr Zahnheilkd.* 1970;80(3):250–278. [Article in French].



• **Fig. 10.11** Two-dimensional structure of sodium silicate glass. Na, Sodium; O, oxygen; Si, silicon. (Modified from Warren BE, Biscoe J. Fourier analysis of x-ray patterns of soda-silica glass. *J Am Ceram Soc.* 1938;21(7):259–265.)

large crucibles. As mentioned earlier, feldspar melts incongruently at about 1150°C to form a glassy phase with an amorphous structure, as illustrated in Fig. 10.11, and crystalline phases, possibly consisting of leucite, a potassium aluminosilicate ( $\text{KAlSi}_2\text{O}_6$ ).

The mix is then cooled very rapidly (quenched) in water causes the mass to shatter into small fragments. The product obtained, called a *frit*, is ball milled to achieve proper particle size distribution. Coloring pigments in small quantities are added at this stage to obtain the delicate shades necessary to mimic natural teeth. The metallic pigments include titanium oxide for yellow-brown shades, manganese oxide for lavender, iron oxide for brown, cobalt



• **Fig. 10.12** Three-dimensional structure of leucite ( $\text{KAlSi}_2\text{O}_6$ ). Al, Aluminum; K, potassium; O, oxygen; Si, silicon.

oxide for blue, copper or chromium oxides for green, and nickel oxide for brown. In the past, uranium oxide was used to provide fluorescence; however, because of the small amount of radioactivity, lanthanide oxides (such as cerium oxide) have been substituted for this purpose. Tin, titanium, and zirconium oxides are used as opacifiers.

After the manufacturing process is completed, feldspathic dental porcelain consists of a glassy (or amorphous) phase and leucite ( $\text{KAlSi}_2\text{O}_6$ ) as a crystalline phase. The glassy phase formed during the manufacturing process has properties typical of glass, such as low toughness and strength, and high translucency. The crystalline structure of leucite is tetragonal at room temperature (Fig. 10.12). Leucite undergoes a reversible crystallographic phase transformation at 625°C, a temperature above which its structure becomes cubic. This transformation is accompanied by a thermal expansion resulting in a 1.2 vol% increase of the unit cell. This explains the high thermal expansion coefficient associated with tetragonal leucite (greater than  $20 \times 10^{-6}/^\circ\text{C}$ ). As a result, the amount of leucite present (10–20 vol%) controls the thermal expansion coefficient of the porcelain so that it is adequately matched to that of dental alloys.



• Fig. 10.13 Scanning electron micrograph showing the microstructure of feldspathic porcelain for metal-ceramic restorations. GM, Glassy matrix; LC, leucite crystal.

The microstructure of conventional feldspathic porcelain is shown in Fig. 10.13; the glassy phase has been lightly acid-etched to reveal the leucite crystals. Typical compositions for opaque and dentin porcelain powders are given in Table 10.4.

Feldspathic porcelains have other qualities that make them well suited for metal-ceramic restorations. They fuse at lower temperatures than do many other ceramic materials, lessening the potential for distortion of the metal coping. This is made possible by the presence of alkali oxides ( $\text{Na}_2\text{O}$  and  $\text{K}_2\text{O}$ ) in the glassy matrix; these oxides are responsible for the creation of nonbridging oxygens in the glass network, thereby lowering the fusing temperatures to the range  $930^\circ\text{C}$  to  $980^\circ\text{C}$ . Porcelains having an even lower fusing temperature ( $760^\circ\text{C}$ – $780^\circ\text{C}$ ) and high CTE ( $15.8 \times 10^{-6}/^\circ\text{C}$ ) are also available. These porcelains are designed to be compatible with bonding to yellow high-Au alloys, which have coefficients of thermal expansion between 16.1 and  $16.8 \times 10^{-6}/^\circ\text{C}$ . They can, however, be abrasive to opposing teeth because of their hardness; this becomes a significant problem if the porcelain surface is roughened by occlusal adjustments.

### Effect of Design on Metal-Ceramic Restorations

Because ceramics are weak in tension and can withstand very little strain before fracturing, the metal framework must be rigid to minimize deformation of the porcelain. However, copings should be as thin as possible to allow space for the porcelain to mask the metal framework without overcontouring the porcelain. This consideration is especially true for alloys that appear gray. This might lead to the conclusion that nickel-chromium (Ni-Cr) or cobalt-chromium (Co-Cr) alloys would be superior to the noble alloys because their moduli of elasticity (stiffness) are 1.5 to 2 times greater and the thickness of the coping could be halved. However, loading the restoration places it in bending, and the bending deflection is a function of only the first power of the modulus, whereas it is a function of the cube of the thickness. It can be shown that for a typical dental metal-ceramic restoration, the thickness of a base-metal coping can be reduced only about 7% because of the higher modulus of elasticity. Thus the advantage of the higher modulus for the base-metal alloys is minimal.

The labial margin of metal-ceramic prostheses is a critical area regarding design because there is little porcelain thickness at the margin to mask the appearance of the metal coping and to resist

fracture. Recommended margin designs include a 90-degree shoulder, a 120-degree shoulder, or a shoulder bevel. Provided that the shoulder depth is at least 1.2 mm, these designs should all provide sufficient porcelain thickness to minimize the risk of porcelain fracture.

When using partial porcelain coverage, such as when a metal occlusal surface is desired, the position of the metal-ceramic joint is critical. Because of the large difference in modulus of elasticity between porcelain and metal, stresses occur at the interface when the restoration is loaded. These stresses should be minimized by placing the metal-ceramic junction at least 1.5 mm from centric occlusal contacts.

The geometry of the interproximal connector area between abutment crown and pontic is critical in the design of a metal-ceramic FDP. The inciso-cervical thickness of the connector should be large enough to prevent deformation or fracture because deflection is decreased as the cube of the thickness; greater thickness will minimize deflection of the framework, which may lead to debonding or fracture of the porcelain. It should be noted that an FDP is not a uniform beam; maximum deflection on loading will occur at the thinnest cross section, which is the interproximal connector area. However, connector thickness cannot impinge on gingival tissues or restrict access to oral hygiene procedures.

### Failure and Repair of Metal-Ceramic Restorations

Metal-ceramic restorations remain a viable material combination selected for crown and bridge applications and have a 10-year success rate of about 95%. The majority of retreatments are due to biological failures, such as tooth fracture, periodontal disease, and secondary caries. Prosthesis fracture and esthetic failures account for only 20% of retreatment cases for single-unit restorations. For metal-ceramic FDPs, prosthesis fracture is the most common reason for retreatment, with long-span FDPs (five or more units) having approximately twice the incidence of failure compared with short-span FDPs.

When metal-ceramic prosthesis fails, it is often due to adhesive failure between porcelain and metal or cohesive failure within the ceramic near the metal-ceramic interface. Ideally, the prosthesis should be retrieved, metal surfaces should be cleaned, and a new oxide layer should be formed on the exposed area of metal prior to porcelain application and firing. However, this cannot be achieved intraorally, and removal of the prosthesis is both unpleasant for the patient and time consuming. Thus a variety of techniques have been developed for porcelain repair using resin composites. All of these techniques present the challenge of bonding chemically dissimilar materials. When porcelain fragments are available and no functional loading is exerted on the fracture site, silane coupling agents can be used to achieve good adhesion between the composite and porcelain; however, comparable bonding agents are not as available for metal alloys and this type of repair is considered only temporary. Systems are available for coating the metal surface with silica particles through airborne particle abrasion. The particles are embedded in the metal surface upon impact, then a silane coupling agent can be applied. Alternatively, base metal alloys can be coated with tin followed by the application of an acidic primer. Both methods achieve adequate bond strength and may delay the eventual need for remaking the prosthesis.

For a discussion of porcelain application and properties of porcelain denture teeth, please go to the website <http://evolve.elsevier.com/sakaguchi/restorative>.

# 14

# Dental and Orofacial Implants

JASON ALAN GRIGGS

## CHAPTER OUTLINE

### Classification

Endosseous Implant

### Osseointegration and Biointegration

### Factors Affecting the Endosteal Implant

Geometry

Magnitude of the Force

Duration of the Force

Type of Force

Implant Diameter

Implant Length

### Surfaces and Biocompatibility

Ion Release

Surfaces

Surface Alterations

Surface Coatings

### Implant Materials and Processing

### Challenges and the Future

The practice of restorative dentistry seeks to replace the form and function of missing tooth structures. It was therefore expected that dentistry would follow orthopedic medicine in the use of implants to anchor prosthetic devices, and that has happened. Global dental implant sales are expected to reach over \$9.62 billion by 2030 at a compounded annual growth rate of 9.8%. In 2022 North America exhibited the largest market share, followed by Europe and the Asia-Pacific region. The Asia-Pacific region is expected to see the highest growth rate between 2023 and 2030.

Dental implants are fixtures that serve as replacements for the root of a missing natural tooth. Implants may be placed in the mandible or maxilla. When properly designed and placed, dental implants integrate with bone over time and serve as an anchor for dental prostheses. Dental implants are used to replace a single missing tooth or many teeth or to support a complete removable denture.

Worldwide, modern single-tooth implants have a success rate of nearly 95% survival at 15 years. Implants are permanent devices that are surgically anchored in the oral cavity and often provide significant advantages over other fixed or removable prosthodontic options. Implants are often more conservative than traditional fixed partial dentures because they save tooth structure

by eliminating the need for reduction of adjacent abutment teeth and they support the maintenance of healthy bone in the region. Implants also increase the stability of overdentures, resulting in bite forces similar to that of patients who retain their natural dentition.

## Classification

Historically, dental implants have been classified according to their design. This design was in turn based on the way in which they are surgically implanted. The three types of implants commonly used for the past 40 years are the subperiosteal implant, the transosteal implant, and the endosseous implant (Table 14.1). More recently, zygomatic implants have been used in case of a severely atrophic or missing maxilla.

## Endosseous Implant

Endosseous implants are by far the most common type of implant placed today. Implants are placed directly into the mandible or maxilla (Fig. 14.1). A pilot hole is drilled into the alveolar or basal bone beneath (in cases in which the alveolar bone has been partially or completely resorbed), and the implant body is inserted into this site. The top of the implant is positioned so that it either protrudes slightly through the cortical plate or is flush with the surface of the bone. Typically, a superstructure containing a prosthetic tooth or teeth connects to the implant body through an abutment that is screwed into the body directly through the mucosa.

## Osseointegration and Biointegration

A major issue for implant design is the development of a device made from materials that are physically and biologically compatible with alveolar bone. Ideally, bone should integrate with the device and remodel the bone structure around it, rather than responding to it as a foreign substance by encapsulating it with fibrous tissue. Under optimum circumstances, bone differentiation occurs directly adjacent to the material (osseointegration). Ideally this osseointegration provides a stable bone-implant connection that can support a dental prosthesis and transfer applied loads without concentrating stresses at the critical interface between bone and the implant.

*Osseointegration* is formally defined as the close approximation of bone to an implant material (Fig. 14.2). To achieve osseointegration, the bone must be viable, the space between the bone and implant must be less than 10 nm and contain no fibrous tissue, and the bone-implant interface must be able to survive loading by

**TABLE 14.1** **Implant Design Classification Scheme**

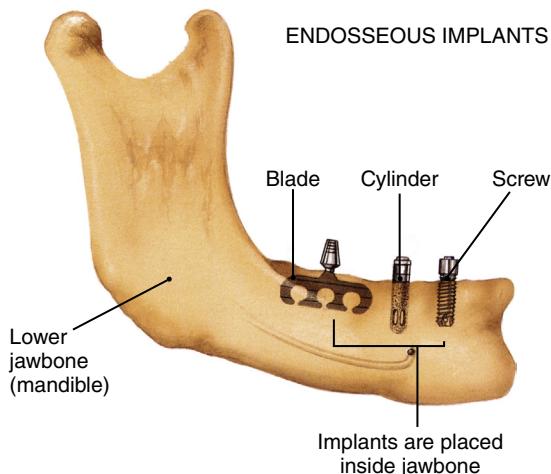
Implant Design	Contact With Bone	Composition	Location Used
Subperiosteal	Directly on bone surface under the gingival tissues; no bone penetration	Co-Cr-Mo (Vitallium)	Maxilla and mandible
Transosteal	Completely through the bone, penetrating the cortical wall twice	Titanium or Ti alloy	Mandible only
Endosteal	Within the bone, penetrating the cortical wall once	Titanium or Ti alloy	Maxilla and mandible

a dental prosthesis by remaining intact. In current practice, osseointegration is an absolute requirement for the successful implant-supported dental prosthesis. To achieve osseointegration between an implant and bone, a number of factors must be correct. The

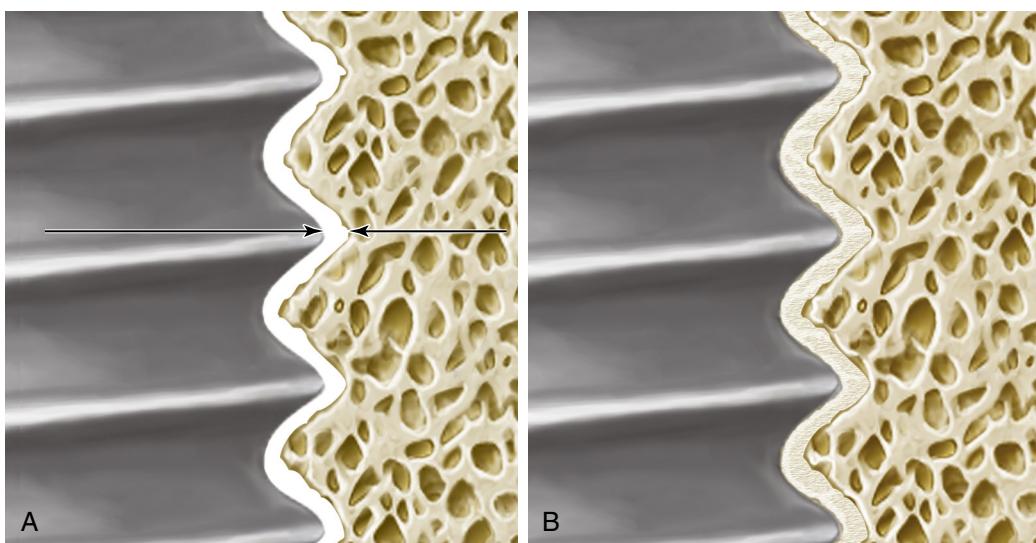
bone must be prepared in a way that does not cause necrosis or inflammation. The implant must be allowed to heal for a time without a load unless initial stability of the implant (as measured using vibration resonance in a magnetic field) is sufficient to prevent micromotion. Finally, the proper material must be implanted, because not all materials will promote osseointegration.

In recent years, various surface configurations have been proposed as means of improving the cohesiveness of the implant-tissue interface, maximizing load transfer, minimizing relative motion between the implant and tissue, minimizing fibrous integration and loosening, and lengthening the service life of the construct. Because of the necessity of developing a stable interface before loading, effort has been placed on developing materials and methods to accelerate tissue apposition to the implant surface. Surface-roughened implants and ceramic coatings have been implemented into clinical practice. Other techniques include electrical stimulation, bone grafting, and the use of growth factors and other tissue engineering approaches described in [Chapter 16](#).

The application of bioactive ceramics as implant materials was traditionally limited to their use as bone bonding and augmentation materials. There has been interest in coating titanium alloys with bioactive materials to promote an implant-bone connection. Bioactive ceramic materials are more than just biocompatible. The use of the term *bioactive* implies that they have the ability to elicit a favorable tissue response when implanted *in vivo*. These ceramics form a direct chemical bond with natural tissues and are most often designed with high solubility to bioresorb or biodegrade,



• **Fig. 14.1** Endosteal implant design. Shown here are three different endosteous implant designs. Notice that all of the designs are implanted directly within the bone. Although the blade design has fallen out of use, the press-fit (cylinder) and screw-shaped versions continue to be the most widely placed implant designs in use today.



• **Fig. 14.2** Osseointegration and biointegration. (A) In osseointegration, the implant material (*left*) and the bone (*right*) closely approximate one another. This approximation must be closer than 10 nm (arrows). In the intervening space, there can be no fibrous tissue. (B) In biointegration, the implant and bone are fused and continuous with one another. Osseointegration commonly occurs with titanium alloys, whereas biointegration occurs with ceramics and ceramic-coated metallic implants.

in order to be replaced over time by natural tissues. Commonly implanted dental ceramics include calcium phosphates with various calcium-to-phosphorus ratios (e.g., hydroxyapatite [HA] and tricalcium phosphate), bioactive glasses (mixtures of  $\text{SiO}_2$ ,  $\text{CaO}$ ,  $\text{P}_2\text{O}_5$ , and sometimes  $\text{Na}_2\text{O}$ , and  $\text{MgO}$ ), and glass ceramics.

Important examples of bioactive glasses and glass-ceramics include Bioglass (a glass containing a mixture of silica, phosphate, calcia, and soda); Ceravital (which has a lower alkali oxide concentration compared to Bioglass); Biogran (which has a different physical conformation compared to Bioglass); and A-W glass ceramic (a glass-ceramic-containing crystalline oxyapatite and fluorapatite [ $\text{Ca}_{10}\{\text{PO}_4\}_6\{\text{O},\text{F}_2\}$ ] and  $\beta$ -wollastonite [ $\text{SiO}_2\text{CaO}$ ] in a  $\text{MgO}\text{-CaO}\text{-SiO}_2$  glassy matrix). In addition, many other glass and glass-ceramic compositions, based on recently developed sol-gel synthesis methods, are being developed. Calcium phosphate ceramics vary in composition, depending on processing-induced physical and chemical changes. Among this group are the apatite ceramics. Of particular interest is HA, which is the synthetic version of the inorganic phase found in teeth and is similar to the inorganic phase found in bone (hydroxycarbonate apatite). For this reason, it is the bioactive ceramic material that has been most extensively investigated.

The impetus for using synthetic HA as a biomaterial stems from the perceived advantage of using a material similar to the mineral phase in natural tissues to replace them. Because of this similarity, better tissue bonding is expected. Additional perceived advantages of HA and other bioactive ceramics include low thermal conductivity, elastic properties similar to those of bone, control of in vivo degradation rates through control of material properties, and the possibility of the ceramic functioning as a barrier to metallic corrosion products when it is coated onto a metal substrate.

However, temperature-induced phase transformations while processing HA provoke considerable changes in its in vitro dissolution rate and the altered structure changes the biological reaction to the material, which becomes more or less bioactive, respectively, with vitrification and crystallization. Given the multitude of chemical compositions and structures resulting from processing bioactive ceramics and the resultant fact that pure HA is rarely used, the broader term *calcium phosphate ceramics* has been proposed in lieu of the more specific term HA. Each individual calcium phosphate ceramic is then defined by its own unique set of chemical and physical properties.

Although calcium phosphate ceramics are too brittle and too stiff to serve as stand-alone dental implants for prosthetic tooth replacement, there has been continuing interest in using a thin (50 to 75  $\mu\text{m}$ ) layer of ceramic materials to coat the surface of metallic implants. This provides the beneficial osseointegration characteristics of the ceramic combined with the high strength of the metallic alloy. Most manufacturers provide implants coated with calcium phosphate ceramic for use in sites where poor bone quality exists. A major limitation in using this concept in all clinical situations, however, has revolved around the inability to predict and maintain the bond strength of the coating to the metal. When the bioactive ceramic material resorbs in vivo, an unpredictable change occurs in the implant–bone interface, and implant micromotion and loosening may occur. This makes the long-term stability of these implants uncertain.

If successful, the ceramic coating becomes completely fused with the surrounding bone. In this case, the interface is called *biointegration* rather than osseointegration, and there is no intervening space between the bone and the implant (see Fig. 14.2). A number of ceramic coatings have been used in this manner.

Typically, these coatings have been applied to the surface of an implant via a plasma-spray deposition process. This results in a complex mixture of HA, tricalcium phosphate, and tetracalcium phosphate in the coating, rather than a recapitulation of the starting powder mixture. Physical properties of importance to the functionality of calcium phosphate ceramics include powder particle size, particle shape, pore size, pore shape, pore-size distribution, specific surface area, phases present, crystal structure, crystal size, grain size, density, coating thickness, hardness, and surface roughness.

The long-term integrity of the ceramic coating *in vivo* is not known, but evidence indicates that these coatings will resorb over time. In addition, results of *ex vivo* pushout tests indicate that the ceramic–metal bond fails before the ceramic–tissue bond and is the weak link in the system. Thus the weak ceramic–metal bond and the integrity of that interface over a lengthy service life of functional loading suggests reason for concern.

## Factors Affecting the Endosteal Implant

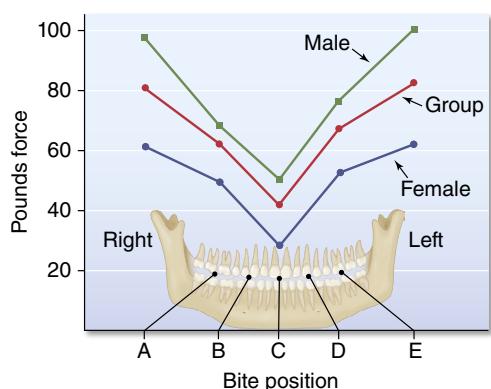
### Geometry

Two primary objectives influence a patient's decision to pursue dental implant treatment: aesthetics and function. To fulfill these objectives over an extended period, a dental implant must be capable of withstanding the occlusal forces generated in the oral environment and in turn transfer this load to the supporting tissues. Not only must loads be transferred, but they should also be of an appropriate direction and magnitude so tissue viability is maintained. In this respect, the implant principally acts to minimize and distribute the biomechanical forces. The forces are characterized by their magnitude, duration, and type. The ability to transfer force largely depends on attaining interfacial fixation. The interface between the implant and bone must stabilize in as short a time as possible postoperatively, and once stable, must remain so throughout its service life. Designing an “optimal” implant that meets all the foregoing objectives requires the integration of material, physical, chemical, mechanical, biological, and economic factors.

### Magnitude of the Force

The amount of load applied during normal chewing varies greatly, depending on the specific patient, the location in the arch, and the state of their dentition. Bite force values reported in the literature range from about 40 to 1250 N. The magnitude of force is greatest in the molar region because this area acts like the hinge of a lever (Fig. 14.3). The incisor region, in comparison, experiences about 10% of the magnitude seen in the posterior segment. This difference in load borne by the teeth and supporting bone dictates differences in mechanical requirements between anterior and posterior implants. Since stress depends not only on the applied load but also on the area over which this load is distributed, the loss of some teeth by a patient will greatly increase the stresses applied to the remaining teeth and implants in partially edentulous patients.

A prime requirement for any dental implant is adequate supporting bone height, width, and density. It is well established that bone grows in response to strain, and the presence of an excessive magnitude of stress applied to the bone will result in resorption or loss of bone. However, in the absence of a critical minimum level of strain for normal bone maintenance, the bone will also resorb. Therefore if the patient has been edentulous for a prolonged time,



• Fig. 14.3 Mean adult bite force at different positions in the jaw. (From Rugh SD, Solberg WK. The measurement of human oral forces. *Behav Res Methods Instrum.* 1972;4(3):125.)

the underlying bone will have resorbed and become less dense. It is common to place implants preferentially in the anterior mandible because this region has the greatest trabecular bone density when compared with the premolar or molar regions in both dentate and edentulous patients. When planning implant treatment, careful consideration must be given to the load distribution.

A great majority of the materials considered to be biocompatible are not suitable for use as implants, because their ultimate strength is not high enough to withstand the forces to which they are subjected during normal function. However, in order to survive and continue to function effectively, it is not only the ultimate strength, but also the modulus of elasticity (or stiffness) of the material that must be considered. Unless the bone experiences at least 1000 microstrains on a routine basis, it will begin to resorb. Most ceramic materials are extremely stiff; for example, polycrystalline aluminum oxide has a modulus of elasticity of  $\approx 372$  GPa. This stiffness is too high to transfer an adequate amount of an applied force to the bone. Instead, the stiffer implant material will carry a disproportionate amount of the load, causing stress shielding of the bone. By contrast, titanium has a modulus of elasticity of  $\approx 100$  GPa; still too high to be ideal, but much closer to that of bone ( $\approx 20$  GPa). It will permit normal physiological loading of the bone.

## Duration of the Force

When considering repetitive loading such as that occurring during mastication, it is more appropriate to consider the endurance limit of a material rather than its ultimate strength. The endurance limit is the highest amount of stress to which a material may be repeatedly subjected without failing. This limit is typically only about half of the ultimate strength of the material.

The tooth root-form implant is designed to be loaded parallel to the long axis and is vulnerable to fatigue failure from cyclic bending loads. These bending loads often result from premature contact, bruxism, inappropriate occlusal schemes, and the use of angled abutments. Off-axis loading should be avoided in design of the implant superstructure.

## Type of Force

An implant experiences three types of loads in function: tensile forces, compressive forces, and shear forces. As discussed, a well-designed implant transfers and distributes these forces to the supporting bone. Bone is composed of both inorganic and organic

constituents, and the inorganic components render it strongest when loaded in compression. Bone is about 30% weaker when placed in tension and nearly 70% weaker when subjected to shear forces. Therefore occlusion is a crucial consideration in designing the implant loading.

Smooth-sided cylindrical implant designs place the interface between the implant and the bone in nearly pure shear, the weakest possible loading scenario. These implant designs rely either on microscopic texturing of the implant body to offer some mechanical interlocking and provide retention, or on a coating. If the coating on these fails or resorbs, bone loss usually results from the lack of load transfer.

By contrast, screw-shaped implants have threads to engage the bone in compression and transfer the applied load. The thread designs have been extensively researched to provide a minimum of shear forces and maximal compression to the bone. This allows for the most favorable bone response. A number of recent thread designs have been introduced that use rounded thread tips (to reduce shear forces at the tip), changes in the thread angle (to maximize compression), two or more thread profiles on the same implant (which will cut at different locations in the osteotomy, thereby increasing the contact area), and/or a reduced thread height accompanied by an increased thread pitch (spreading the implant loading over a greater contact area while simultaneously increasing the strength of the implant body).

The concept of osseointegration around cylindrical or screw-shaped implants represents a situation of bone ongrowth. An alternative method of implant fixation is based on bone tissue ingrowth into roughened or three-dimensional porous surface layers. Such designs incorporating sintered beads or a sintered wire mesh are typical of orthopedic implants. Recently, implants with porous metal bodies have been developed and marketed. These new designs incorporate macroscale porosity or porous cellular structures, resembling cancellous bone. This results in a modulus of elasticity close to that of bone, thereby reducing the resulting stress shielding. Such retention systems have been shown to have higher bone/metal shear strength than other types of fixation. Increased interfacial shear strength results in a better stress transfer from the implant to the surrounding bone, a more uniform stress distribution between the implant and bone, and lower stresses in the implant. In principle, the result of a stronger interfacial bond is a decreased propensity for implant loosening. The theoretical progression of macroscale surface effects from the lowest implant/tissue shear strength to the highest is as follows: smooth, textured, screw threaded, plasma sprayed, and porous coated, porous body design.

## Implant Diameter

An increase in implant length or diameter increases the total surface area of the implant. As a consequence, the area for distribution of the occlusal forces is increased and the stress on the bone is decreased. The bending fracture resistance (and hence rigidity) of the implant increases greatly as the implant width increases and is related to the implant radius raised to the fourth power. This dramatic increase can be deleterious if the diameter chosen causes stress shielding by reducing bone strain to subphysiologic levels.

## Implant Length

As with increases in the width, increases in implant length also increase the surface area and reduce the bone stress. However, careful consideration of the bone quality is advised. In the highly dense