Chapter 4

Fracture of Dental Materials

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Additional information is available at the end of the chapter

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1. Introduction

Finding a material capable of fulfilling all the requirements needed for replacing lost tooth structure is a true challenge for man. Many such restorative materials have been explored through the years, but the ideal substitute has not yet been identified. What we use today for different restorations are different metals, polymers and ceramics as well as combinations of these materials. Many of these materials work well even though they are not perfect. For example, by coating and glazing a metal crown shell with a ceramic, it is possible to make a strong and aesthetic appealing crown restoration. This type of crown restoration is called a porcelain-fused-to-metal restoration, and if such crowns are properly designed, they can also be soldered together into so called dental bridges. The potential problem with these crowns is that the ceramic coating may chip with time, which could require a complete remake of the entire restoration. Another popular restorative material consists of a mixture of ceramic particles and curable monomers forming a so called dental composite resin. These composites resins can be bonded to cavity walls and produce aesthetic appealing restorations. A potential problem with these restorations is that they shrink during curing and sometimes debond and fracture. In addition to porcelain-fused-to-metal crowns and composites, all-ceramic and metallic restorations as well as polymer based dentures are also commonly used. These constructions have their inherent limitations too.

The reason it is difficult to make an ideal dental material is because such a material has to be biocompatible, strong, aesthetic, corrosion resistant and reasonable easy to process, properties that are difficult to find in one single material. Besides, material as well as processing costs of such a material should be relatively low in order to make the use of the material wide among all social-economical groups. That demand makes the ideal material identification process even more challenging.

Today, dentistry to a great deal is driven by aesthetic demands, restricting the selectable materials mainly to tooth colored materials. Because of that demand, dentists are moving

away from traditional metallic restorations with high fracture toughness values, toward resin based composites and all-ceramic restorations with rather low fracture toughness values. Modern all-ceramic restorations consist of core structures made by fracture tough ceramics such as alumina and partly stabilized zirconia. However, the rather opaque appearance of these two ceramics often requires that they are veneered with less fracture tough but more aesthetic appealing ceramics. The use of more aesthetic appealing materials has not increased the longevity of dental restorations, but in some cases when composite resins are being used, the move toward bonded composites might have increased the way tooth structures can be preserved. The benefit of such usage is that it decreases the amount of tooth structure needed to be removed during preparation and can therefore increase the longevity of the tooth.

The intention with this chapter is to give an overview of some fundamental fracture mechanics aspect of aesthetic restorative materials such as dental ceramics and dental composite resins, as well as some fracture mechanics considerations related to the way ceramics and composites are bonded to the tooth via a cement/adhesive. However, before addressing these man-made materials, the two most important dental materials, the biologically developed materials, enamel and dentin, will be discussed. An insight into the fracture mechanics of these two substrates clearly shows how sophisticated Nature was when these two biologic materials evolved. An understanding of enamel and dentin shows quite clearly where the limitations and short-comings are with the man-made dental materials, and may help us in developing better restorative dental materials.

2. The tooth

Nature provided animals and humans with teeth to be used for digesting food, but also as tools for hunting and self-defense. To fulfill these functions, Nature developed enamel to become the hardest biological tissue. Tooth enamel ranks 5 on Mohs hardness scale, where steel is ranked 4.5 and thus slightly softer than enamel. Its Young's modulus is 83 GPa, which falls between aluminum (69 GPa) and bronze (96-120 GPa)[1]. The enamel can be described as the whitish looking shell covering the visible part of a tooth positioned in the alveolar socket (Figure 1).

Regarding enamel and dentin, the first hard tissue to form is dentin, produced by newly differentiated odontoblasts. The first formed dentin layer is called mantle dentin and is approximately 150 μm thick and contains loosely packed coarse collagen fibrils surrounded by precipitated hydroxyapatite crystals [2]. Tiny side-branching channels oriented parallel to the dentin-enamel-junction (DEJ) and connected to the protoplasmatic extension of the odontoblasts are parts of the mantel dentin. The mantle dentin matrix is slightly less (4 vol-%) mineralized then the rest of the finally formed dentin.

As the odontoblasts move away from the DEJ, each of them leaves a cell extension protruding from the odontoblasts to the DEJ with the side-branching channels of the mantle dentin. These cell extensions may remain in contact with the DEJ during the formation of dentin as well as during the lifetime of the tooth, and they form channels through the dentin as the odontoblasts move inwards toward the pulp. The secreted collagen fibers, which are mainly

oriented perpendicularly to the dentinal tubules, act as nucleisation centers for hydroxyapatite crystallites precipitating as the odontoblasts migrate inwards (Figure 2). The dentin formed by these collagen fibers represents the so called intertubular dentin (Figure 3). However, surrounding the odontoblastice processes are thin layers of collagen oriented parallel to the odontoblastic processes. These collagen layers are also mineralized and form the so called peritubular dentin, which is denser than the intertubular dentin located between the peritubular dentin tubules. An important difference between enamel and dentin is that dentin, in contrast to enamel, is a living tissue as long as the pulp is alive, while the enamel becomes a completely dead tissue as soon as the outer layer of the enamel has formed and the ameloblasts degraded.

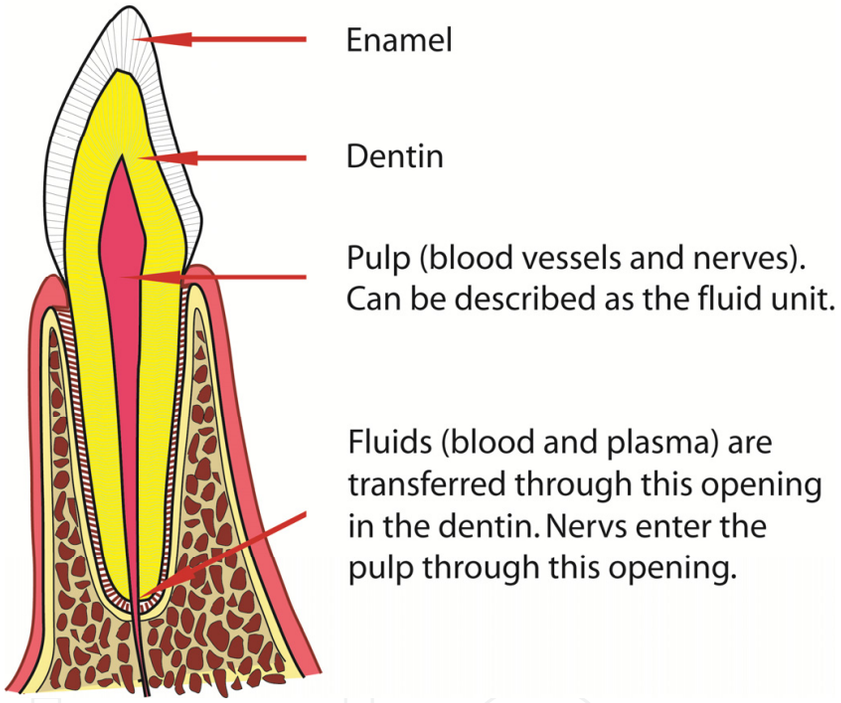

Figure 1. Drawing showing a tooth attached in its alveolar socket. The root of the tooth is attached to the alveolar socket via collagen fibers, the so called periodontal ligament. Blood vessels and nerves enter the pulp chamber via the apical opening.

The formation of mantle dentin triggers the ameloblasts to start secreting enamel proteins on the newly formed mantle dentin. The first hydroxyapatite crystals that form on the mantle dentin are randomly packed in this first formed enamel and interdigitated with the crystallites of dentin. Eventually the dentin crystallites present in the mantel dentin act as nucleation sites for the first enamel crystallites.

After the first layer of structureless enamel has formed, the ameloblasts move away from the DEJ, which permits the formation of the so called Tomes' processes, which form at the ends the ameloblasts closest to the DEJ. When the Tomes' processes are established, the enamel rods start developing (Figure 4).

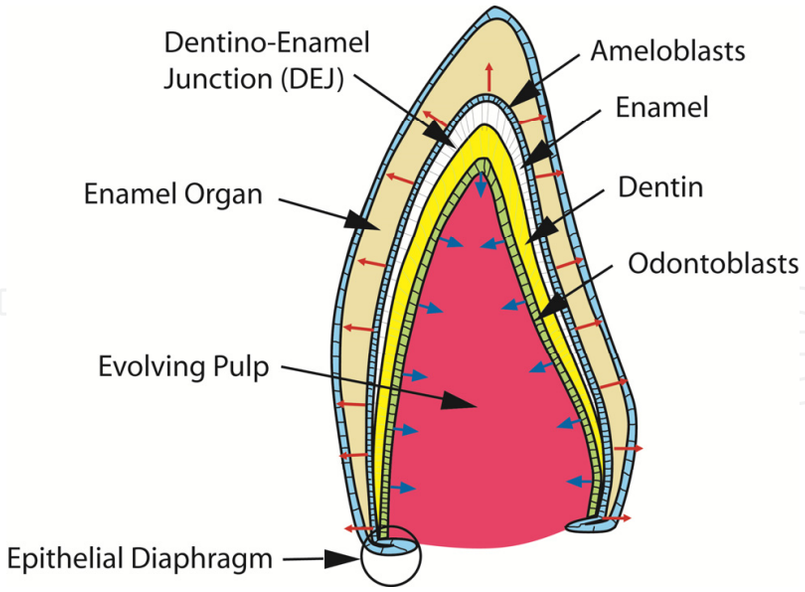

Figure 2. The hard tissues are formed by the odontoblasts (dentin) and ameloblasts (enamel). During the development of the tooth, epithelial cells have formed a bell shaped enamel organ. Inside that bell is connective tissue that shows active budding of capillaries. At a certain stage, the fibroblasts in contact with the epithelium bell become highly differentiated and develop into odontoblasts and form the first layer of dentin. That layer stimulates the epithelium cells in contact with the dentin at the DEJ to differentiate into ameloblasts and form enamel. As a consequence, the two cell types move in opposite direction as they form dentin (blue arrows) and enamel (red arrows). When the ameloblasts reach the outer cells of the enamel organ they start degrading and lose vitality. At the same time, the dentin has increased in thickness and the epithelial diaphragm with odontoblasts have grown downwards and developed the root and the pulp chamber (see Figure 1).

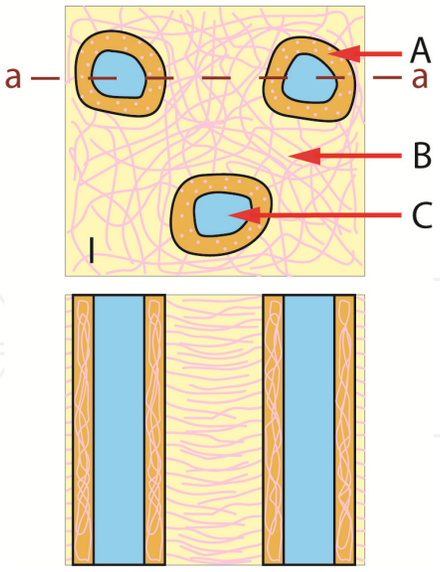

Figure 3. The top drawing represents a cross-section of dentin, perpendicular to the peritubular dentin (A). The lower drawing represents a plane parallel to the odontoblastic processes (C) and cut along a-a. In the peritubular dentin, collagen fibers represented by pink lines are present parallel to the odontoblastic processes. Hydroxyapatite precipitate along these fibers, and together they form the so called peritubular dentin (A). Collagen precipitates perpendicular to the odontoblastic processes too, and when hydroxyapatite precipitate in that matrix, the intertubular dentin (B) is formed.

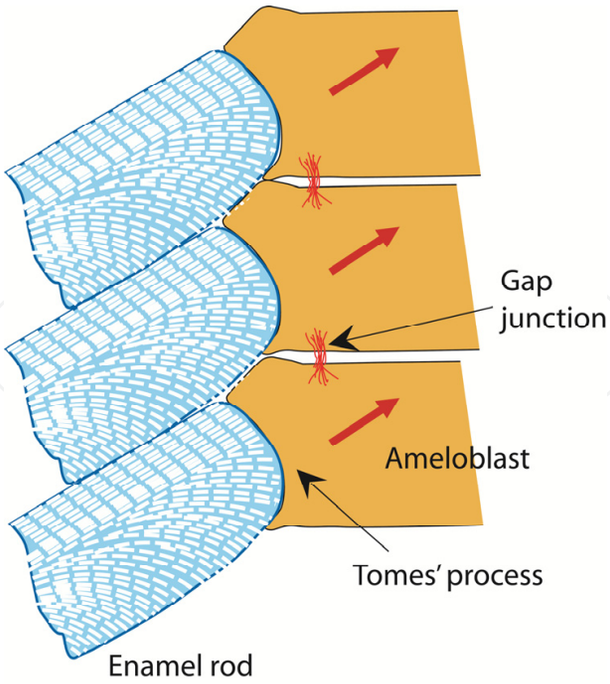

Figure 4. After the first layer of structureless enamel has formed on the mantel dentin, the ameloblast differentiate its end closest to the precipitated enamel into the so called Tomes' process. This unit can be described as a concave formation from which hydroxyapatite crystallites precipitate. The c-axis of these crystallites are perpendicular to the surface of Tomes's process, explaining the the well organized precipitation of the hydroxyapatite crystallites in each enamel rod.

The secretion from the peripheral site of the Tomes' process results in the formation of what is referred to as the enamel matrix wall. These walls enclose pits into which the Tomes' processes fit. These sites are then filled with matrix proteins acting as nucleating agents for the hydroxylapatite crystallites. The crystallites that precipitate in these two matrices (the matrix wall and the central pit) have different orientation. It is important to emphasize that the final wall and pit enamel have the same composition. The only difference is the orientation of the crystallites in these two enamel types.

A cross-section of the enamel rods reveals that the individual rods have a key-hole shaped structure (Figure 5).

As the ameloblasts move toward their final destiny, they produce enamel rods that are somewhat wavy and interwoven (Figure 6). Independent on these waves, the enamel rods form angles that are roughly perpendicular to the outer as well as inner surfaces of the enamel shell (Figure 7). The hard enamel can be described as a hard shield protecting the underlying dental tissue of the visible part of the tooth.

Enamel consists mainly of hydroxyapatite crystallites, which are oriented in very well organized larger bundles of crystallites. These larger bundles are referred to as enamel rods. Each enamel rod is made by enamel forming cells, the so called ameloblasts. The diameters of the rods range from 4-8 μm. During enamel formation, the ameloblasts secret different proteins (amelogenins and enamelins), which act as nucleating agents for the hydroxyl

apatite crystallites. During enamel formation, the ameloblasts move from the dentin-enamel junction (DEJ) to the surface of the final enamel crown. When the enamel shell has reached its final shape, the ameloblasts degenerate and die, explaining why mature enamel is a non-vital tissue made up by ~85 vol-% hydroxyapatite and 15 vol-% proteins and water [2].

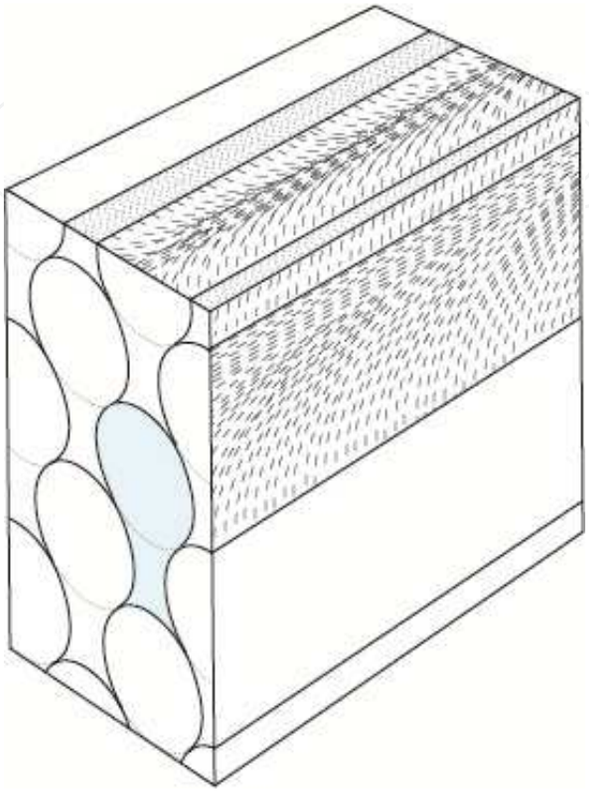

Figure 5. Cross section of enamel rods shows the key-hole structutre (blue). The longitudinal orientation of the hydroxyapatite crystallites can be explained by considering how Tomes' process controls the crystallite orientation (Figure 4). Figure redrawn after [3].

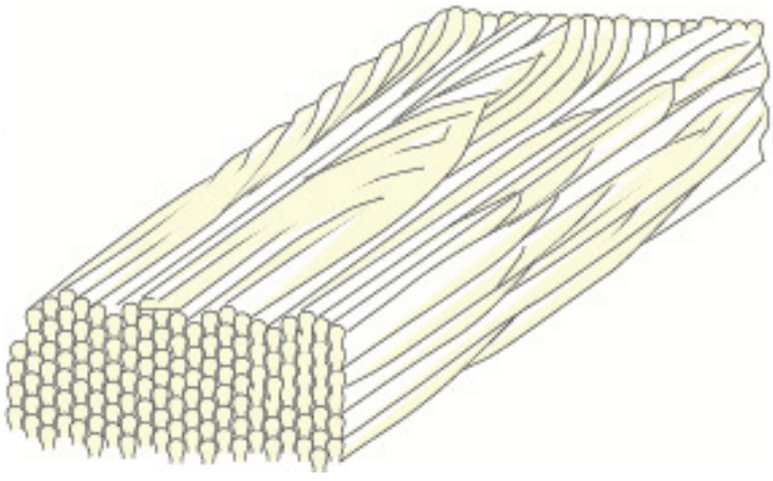

Figure 6. The keyhole shaped rods become more and more interwoven as the rods approaches the DEJ. Redrawn from [4]. The intervowen structure shown in the drawing is also characteristic for the cusp tips, where that type of enamel is called "gnarled enamel".

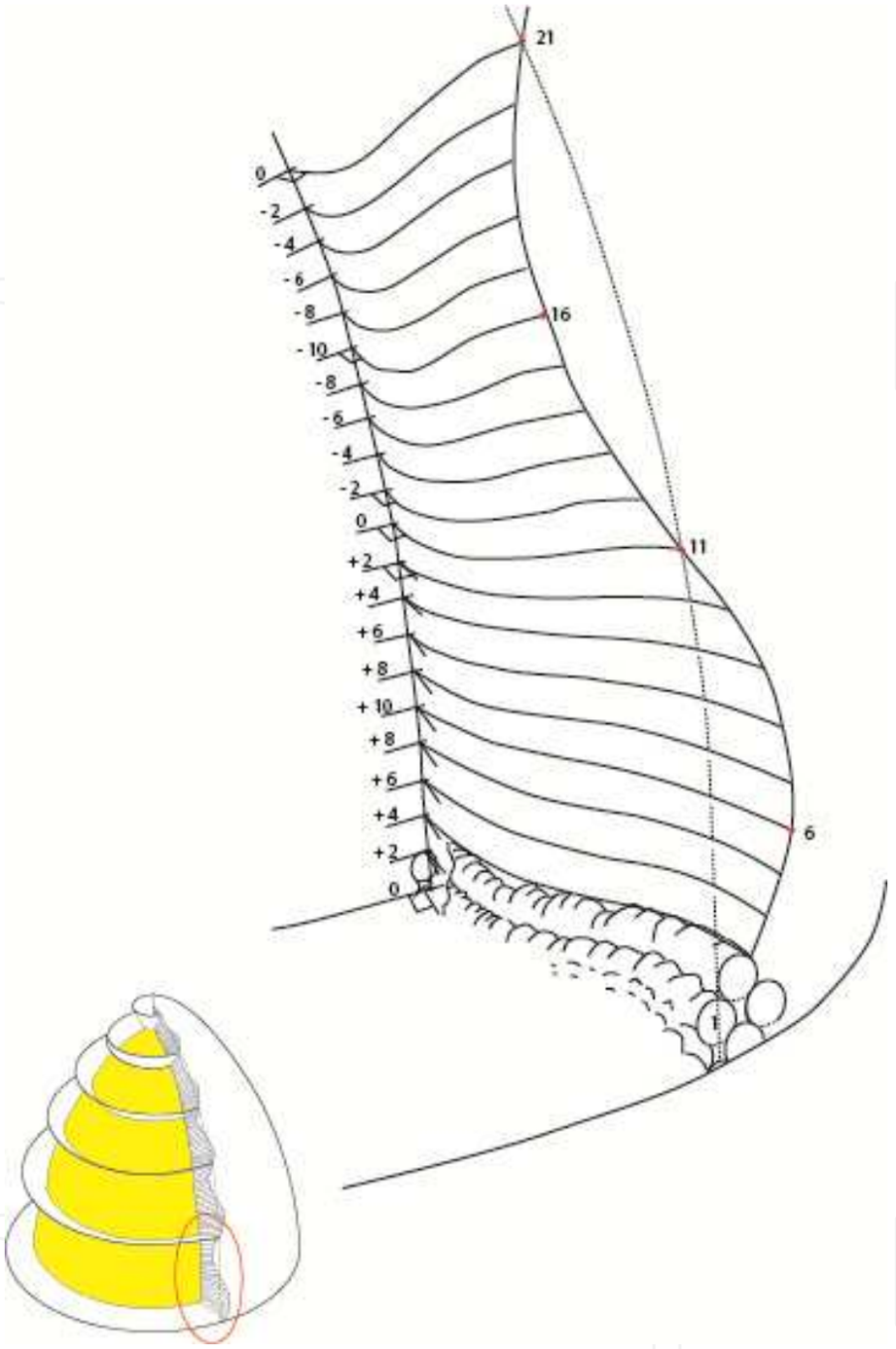

Figure 7. The bottom left drawing shows the orientation of the rods along the long axis of the tooth. As seen from both drawings, there is a continous shift in orientation resulting in the S-shaped orientation. If the line joining points 1, 11 and 21 along the S-shaped curve represents a plane forming 90 degrees to the enamel surface, it is seen from the drawing that there is a difference in rod orientation that can be described as 90 ± 10 degrees. Redrawn from [2].

The pulp chamber is a cavity inside the dentin formed by the surrounding dentin. The pulp chamber contains soft tissue, blood vessels and nerves and is lined by the odontoblasts. As the tooth grew older, the odontoblasts continue to produce dentin, causing the size of the pulp chamber to decrease with age.

As seen from the properties presented in Table 1, enamel has lower fracture toughness than dentin, but significantly higher hardness and modulus of elasticity. These properties suggest that enamel is a highly brittle material that should easily chip away from the dentin. Fortunately that is not the case. The reason can be related to a firm enamel-dentin attachment as well the sophisticated anisotropic composite structure of both enamel and dentin. If cracks propagate through the enamel, they often stop before they reach the enamel-dental interface, and if they continue propagating they usually stop when they reach the enamel-dentin interface. That explains why fractured teeth are not as common as one otherwise would expect by considering force and fatigue levels teeth have to withstand.

Hard tissue	Modulus of elasticity (GPa)	Fracture toughness (MPa $m^{1/2}$)	Hardness (GPa)
Enamel	78 ± 1 to 98 ± 4	0.44 ± 0.04 to 1.55 ± 0.29	2.83 ± 0.10 to 3.74 ± 0.48
Dentin	18 ± 1 to 22 ± 1	3.08 ± 0.33	0.53 ± 0.01 to 0.63 ± 0.03

Table 1. Highest and lowest reported values in Xu et al.'s study[5], except for the fracture toughness value of dentin which is from El Mowafy and Watts study[6]. Identified variations relate to the anisotropic nature of enamel and dentin as well as variations among teeth.

2.1. Fracture mechanical aspects of enamel

As discussed earlier, the tooth can be described as a rather complicated composite structure developed to serve the user. Nature adapted the principle that teeth must be hard and rigid in order to generate sufficiently high local stress levels. These stresses are capable of penetrating tissues during hunting and fighting, but also capable of crushing hard food. At the same time, enamel has also been designed to limit the inherent brittle nature of hydroxyapatite by dispersing propagating cracks and thereby resist some brittle failures.

By orienting the rods on the cusp tips along the axis of the tooth, a parallel model composite is formed in that region (Figure 8). At the same time, by orienting the rods more or less perpendicular to the long axis of the tooth in the remaining parts of the crown, a series model composite is formed in that part of the crown. These models are valid under the assumption the load is in an axial direction. Since the parallel model results in a stiffer combination than a series model material, the tooth has been designed so that rigidity is optimized in the chewing/biting direction and flexing in a direction perpendicular to that direction.

As the stiffness of the parallel model exceeds the stiffness of the series model, we can understand how such a design assists an animal attacking another animal. During such an attack, the canines of the attacking animal may penetrate the tissue of the attacked animal, but that bite may not necessarily result in an instant kill. During the biting action, the attacking animal benefits from the stiffness of the canines (parallel model behavior of the tip of the canines make the tooth stiff like a steel arrow). However, if the attacked animal was not killed instantaneously, most likely it will try to get loose from the attacking animal's jaws. During that attempt the risk of fracturing the canines of the attacking animal increases.

However, thanks to the rod and tubule orientations in the cervical and mid crown regions, the material characteristics of the enamel in these bendable regions are represented by the series model, thereby allowing the tooth to flex somewhat and absorb mechanical energy rather than fracture.

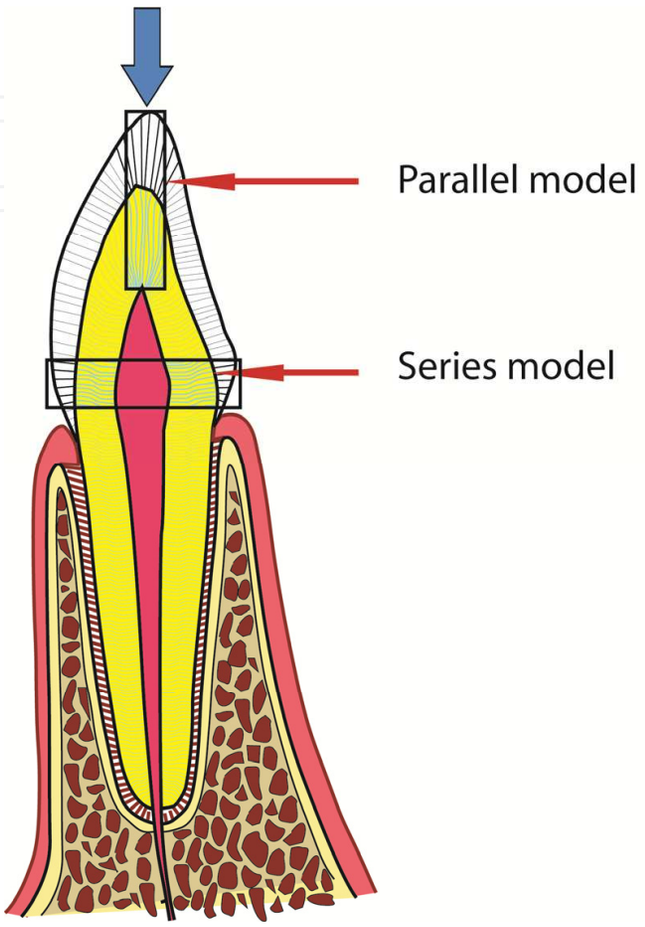

Figure 8. A tooth loaded in axial direction (blue arrow) responds in two ways. On the cusp tips, the rods and the tubules are oriented parallel with the load and resulting in a material which modulus can be predicted from a parallel model prediction. In the cervical region of the tooth, the modulus can be predicted from a series model.

Some basic science studies have been conducted to study the fracture behavior of enamel and dentin through the years. In one such study [7] the investigators studied a mandibular molar tooth restored with different Class II amalgam preparations. By use of finite element analysis, the stress distribution induced along the internal edges as a result of occlusal loading was calculated, and by use of Paris law the cyclic crack growth rate of sub-surface flaws located along the dentinal internal edges was determined. Based on the assumptions used in their calculations, they claimed that flaws located within the dentin along the buccal and lingual internal edges can reduce the fatigue life of restored teeth significantly. Sub-surface cracks as short as 25 μm were capable of promoting tooth fracture well within 25 years from the time of restoration placement. Furthermore, cracks longer than 100 μm reduced the fatigue life of the tooth to less than 5 years. Consequently, sub-surface cracks introduced

during cavity preparation with conventional dental burs may serve as a principal source for premature restoration failure.

As the hardest and one of the most durable load bearing tissues of the body, enamel has attracted considerable interest from both material scientists and clinical practitioners due to its excellent mechanical properties. In a recent article [8] possible mechanisms responsible for the excellent mechanical properties of enamel were explored and summarized. What these authors emphasized was the hierarchical structure and the nanomechanical properties of the minor protein macromolecular components. The experimental and numerical results supported the made assumptions. For example, enamel showed to have lower elastic modulus, higher energy absorption ability and greater indentation creep behavior than sintered hydroxyapatite material. These findings suggest that the structural and compositional characteristics of the minor protein component significantly regulate the mechanical properties of enamel in order to better match its functional needs.

The fascinating aspect of enamel is that its structure seems to have evolved and adapted to the need of the user of the teeth. For example, in some recent publications [9, 10], these issues have been discussed. Lucas et al. [9] proposed a model based on how fracture and deformation concepts of teeth may be adapted to the mechanical demands of diet, while Constantino et al [10] used that model by examining existing data on the food mechanical properties and enamel morphology of great apes (Pan, Pongo, and Gorilla). They paid particular attention to whether the consumption of fallback foods plays a key role in influencing great ape enamel morphology. Their results suggest that so is the case, and that their findings may explain the evolution of the dentition of extinct hominins.

Along these lines, Lee et al.[11] did a comparative study of human and great ape molar tooth enamel. They used nano-indentation techniques to map profiles of elastic modulus and hardness across sections from the enamel–dentin junction to the outer tooth surface. The measured data profiles overlapped between species, suggesting a degree of commonality in material properties. Using established deformation and fracture relations, critical loads to produce function-threatening damage in the enamel of each species were calculated for characteristic tooth sizes and enamel thicknesses. The results suggest that differences in load-bearing capacity of molar teeth in primates are less a function of underlying material properties than of morphology.

From the above studies, it is quite clear that Nature has adapted the structure of enamel to resist fractures. In a study by Bajaj [4] the crack growth resistance behavior and fracture toughness of human tooth enamel was determined. The results were quantified using incremental crack growth measures and conventional fracture mechanics. The results revealed that enamel undergoes an increase in crack growth resistance (i.e. rising R-curve) with crack extension from the outer to the inner enamel, and that the rise in toughness is a function of distance from the dentin enamel junction (DEJ). The outer enamel exhibited the lowest apparent toughness (0.67± 0.12 MPa $m^{0.5}$), and the inner enamel exhibited a rise in the growth toughness from 1.13 MPa $m^{0.5}$/mm to 3.93 MPa $m^{0.5}$/mm. The maximum crack growth resistance at fracture (i.e. fracture toughness (K_C)) ranged from 1.79 to 2.37 MPa $m^{0.5}$.

Crack growth in the inner enamel was accompanied by a host of mechanisms operating from the micro- to the nano-scale. Decussation in the inner enamel promoted crack deflection and twist, resulting in a reduction of the local stress intensity at the crack tip (Figures 6 and 7). In addition, extrinsic mechanisms such as bridging by unbroken ligaments of the tissue and the organic matrix promoted crack closure. Micro-cracking due to loosening of prisms was also identified as an active source of energy dissipation. The unique microstructure of enamel in the decussated region promotes crack growth toughness that is approximately three times that of dentin and over ten times that of bone.

In addition to the micro- and nano-structure of enamel, the tooth anatomy by itself is such that it has adapted to force conditions present in the oral cavity. Anderson et al. [12] modeled what they believed drove the initial evolution of the cingulum. Recent work on physical modeling of fracture mechanics has shown that structures which approximate mammalian dentition (hard enamel shell surrounding a softer/tougher dentine interior) undergo specific fracture patterns dependent on the material properties of the food items [9, 13]. Soft materials result in fractures occurring at the base of the stiff shell away from the contact point due to heightened tensile strains. These tensile strains occur around the margin in the region where cingula develop. In Anderson et al.'s [12] study, they tested whether the presence of a cingulum structure would reduce the tensile strains seen in enamel using basic finite element models of bilayered cones. Finite element models of generic cone shaped ''teeth'' were created both with and without cingula of various shapes and sizes. Various forces were applied to the models to examine the relative magnitudes and directions of average maximum principal strain in the enamel. The addition of a cingulum greatly reduces tensile strains in the enamel caused by ''soft-food'' forces. The relative shape and size of the cingulum has a strong effect on strain magnitudes as well. Scaling issues between shapes are explored and show that the effectiveness of a given cingulum to reducing tensile strains is dependent on how the cingulum is created. Partial cingula, which only surround a portion of the tooth, are shown to be especially effective at reducing strain caused by asymmetrical loads, and shed new light on the potential early function and evolution of mammalian dentitions.

2.2. Fracture mechanical aspects of dentin

Dentin is not as brittle as enamel. However, considering that enamel rests on dentin, and that cracks may propagate through the enamel, it is important to understand the fracture mechanics of dentin.

Human dentin is known to be susceptible to failure under repetitive cyclic fatigue loading. Nalla et al. [14] addressed the paucity of fatigue data through a systematic investigation of the effects of prolonged cyclical loading on human dentin. They performed the evaluations in an environment of ambient temperature and where the dentin was kept in a Hank's balanced salt solution. The results they got were discussed in the context of possible mechanisms of fatigue damage and failure. The stiffness loss data collected were used to deduce crack velocities and the thresholds for such cracking. They concluded that the

presence of small (on the order of 250 μm) incipient flaws in human dentin will not radically affect their useful life as Arola et al.[7] claimed.

Kruzic et al. [15] investigated the fracture toughness properties of dentin in terms of resistance-curve (R-curve) behavior, i.e., fracture resistance increase with crack extension. Of particular interest was the identification of relevant toughening mechanisms involved in the crack growth. Their study was conducted on elephant dentin, and they compared hydrated and dehydrated dentin. Crack bridging by uncracked ligaments, observed directly by microscopy and X-ray tomography, was identified as a major toughening mechanism. Further experimental evidence were provided by compliance-based experiments. In addition, with hydration, dentin was observed to display significant crack blunting leading to a higher overall fracture resistance than in the dehydrated material. In this paper they show how uncracked bridges remain behind the propagating crack, giving the dentin some fracture toughness.

Bajaj et al. [16] used striations resulting from fatigue crack growth in the dentin of human teeth to identify difference between young and old dentin. They used compact tension (CT) specimens obtained from the coronal dentin of molars from young ($17 \leq$ age ≤ 37 years) and senior (age ≥ 50 years) individuals, and exposed the dentin to cyclic Mode I loads. Striations evident on the fracture surfaces were examined using a scanning electron microscope and contact profilometer. Fatigue crack growth striations that developed in vivo were also examined on fracture surfaces of restored molars. The average spacing in the dentin of seniors (130 ± 23 μm) was significantly larger ($p < 0.001$) than that in young dentin (88 ± 13 μm). Fatigue striations in the restored teeth exhibited features that were consistent with those that developed in vitro and a spacing ranging from 59 to 95 μm. Unlike metals, the striations in dentin developed after a period of cyclic loading that ranged from 1×10^3 to 1×10^5 cycles. The study showed that the cracks tend to propagate perpendicular towards the orientation of the tubules, and climb along a plane tangential to the peritubular cuffs and then continue perpendicularly to the tubules.

Yan et al. [17] showed that rather than using a linear-elastic fracture mechanics (LEFM)(K_C) that ignores plastic deformation and tend to underestimate the fracture toughness, a plastic fracture mechanics (EPFM)(K_{JC}) approach was used. The presence of collagen (approximately 30% by volume) was assumed to enhance the toughening mechanisms in dentin. By comparing the values of the fracture toughness values estimated using either LEFM or EPFM, they found that the K_C and K_{JC} values of plane parallel as well as antiplane parallel specimens were different. The fracture toughness estimated based on K_{JC} was significantly greater than that estimated based on K_C (32.5% on average; $p<0.001$). In addition, K_{JC} of antiplane parallel specimens was significantly greater than that of in-plane parallel specimens. Consequently, in order to critically evaluate the fracture toughness of human dentin, EPFM should be employed rather than LEFM.

3. Man made dental materials

By considering the sophistication of the biological materials enamel and dentin, it is easy to understand why it is such a challenge to identify a man made material that can compete

with the biological hard tissues. In addition to their mechanical properties, such a material should be biocompatible, aesthetic, corrosion resistant, easy to process and reasonable inexpensive, making such an identification extremely challenging. Of these properties, strength values within a group of materials are often used by manufacturers in their marketing and by the dentist when it comes to selecting a product. Unfortunately, strength by itself may not be the best parameter to choose. The reason is that strength is a conditional rather than an inherent material property [18]. Strength data alone should therefore not be used to extrapolate and predict the performance of a structure. Instead, they should be used together with the microstructure of the material, processing history, testing methodology, testing environment and failure mechanism. Structural failures are determined by additional failure probability variables in concert with strength that describe stress distributions, flaw size distributions, which can contribute to either single or multiple failure modes. Lifetime predictions require additional information about the time dependence of slow crack growth. Basic fracture mechanics principles and Weibull failure modeling are important to consider.

To make dental treatment even more challenging, just consider how dentists cut teeth and use different materials. During the cutting process, flaws of different sizes are most likely induced in the remaining tooth structure. Flaws and different defects are also most likely induced during handling and insertion of different materials. The impact of such flaws can be devastating for any material, particularly for brittle ceramic materials. To show how different surface treatments can affect the strength properties, Table 2 has been included to show how different surface treatments of glass can affect its strength [19]. A severely sandblasted glass lose as much as 67% of its original strength, while a drawn silica fiber tested in vacuum is 400 times stronger than the glass, a difference that can be related to the presence of water molecules in air.

Glass treatment	Strength (MPa)
Glass rods “as received” from factory	45
Severely sand blasted	14
Acid etched and lacquered	1725
Drawn silica fibers tested in vacuum	12000 – 16000

Table 2. Effect of surface treatments on the strength of glass

By use of Griffith’s equation[20], one can show how the stress level is affected by flaw size and surface energy and explain the results presented in Table 2. That equation further shows that any processing step affecting the size, orientation or distribution of flaws will affect the measured strength of materials, particularly brittle materials. It also shows how environmental conditions may affect surface energy and thereby also the strength.

3.1. Fracture mechanics aspects of ceramics

Clinical experience suggests that all-ceramic crowns may not be as durable as their porcelain-fused-to-metal counterparts, particularly when placed on molar teeth. The reason

relates to the brittleness of ceramics, making them prone for chipping and fracturing [21-27]. In the 1980s and 1990s, crowns were fabricated as enamel-like monoliths from micaceous glass-ceramics (Dicor, Dentsply/Caulk, Milford, DE) and high leucite porcelains (IPS Empress, Ivoclar, Schaan, Lichtenstein), but these ceramics showed unacceptably high failure rates and were soon replaced by improved ceramics [28, 29]. Subsequent crown design has focused on retention of porcelain as an aesthetic veneer fired to much stronger alumina-based ceramics, either glassinfiltrated (InCeram, Vita Zahnfabrik, Bad S.ackingen, Germany) or pure and dense (Procera, Nobel Biocare, Goteborg, Sweden) alumina, as supporting cores. Although alumina-based crowns continued to replace metal-based crowns, failure rates remained an issue [30]. During the past 15 years, ultra-strong core ceramics, e.g. yttria-stabilized zirconia (Y-TZP) and alumina-matrix composites (AMC)[31] have gained in popularity but have yet to be documented regarding their clinical long-term success.

Clinically, bulk fractures are the reported cause of all ceramic crown failure whether the crown is a monolith or a layered structure. According to a fractographic evaluation by Thompson et al.[32], in which they evaluated fractured and recovered Dicor and Cerestore crowns, they found that failures generally did not ensue from damage at the occlusal surface. Instead, for Dicor the cracks emerged from the internal surface, while in the case of Cerestore, the initiation occurred at the porcelain/core interface inside the core materials. In other studies it has been shown that radial cracks are initially contained within the inner core layer, but subsequently propagate to the core boundaries, ultimately causing irretrievable failure. This failure mechanism raises an interesting question: If the ceramic core materials are so strong, why do the cracks not originate in the weak outer porcelain? In the case of porcelain-fused-to-metal, porcelain failures do seem to occur preferentially in the porcelain, although there is some indication that such failures may be preceded by plasticity in the ductile metal [33]. That in turn raises the question: What are the important material parameters that govern these failure modes in crown structures, and how may they be optimized? Maybe McLean's [33] suggestion from 1983 that layered all-ceramic crowns should perform well if the core fracture strength exceeded the yield strength of base metal alloys (about 400–500MPa for gold).

Before diving deeper into the fracture strength of the core, let us accept that there are several factors which can be associated with crack initiation and propagation in dental ceramic restorations. These factors include: (a) shape of the restoration; (b) micro-structural inhomogeneities; (c) size and distribution of surface flaws; (d) residual stresses and stress gradients, induced by polishing and/or thermal processing; (e) the environment in contact with the restoration; (f) ceramic/cement interfacial features; (g) thickness and thickness variation of the restoration; (h) elastic module of restoration components; and (i) magnitude and orientation of applied loads. The possible interactions among these variables complicate the interpretation of failure analysis observations, explaining why fracture behavior of all-ceramic crowns is rather tricky problem to understand.

Even though McLean's[34] suggestion that the core fracture strength exceeded the yield strength of base metal alloys (about 400–500MPa for gold) might be tempting to adopt to, it

is very important to realize that ceramics, in contrast to metals, are brittle materials, and that strength is more of a "conditional" than an inherent material property, and strength data alone cannot be directly extrapolated to predict structural performance [18]. Strength data, particularly of brittle materials, are meaningful when placed into context via knowledge of material microstructure, processing history, testing methodology, testing environment and failure mechanism(s). Lifetime predictions require additional information about the time dependence of slow crack-growth. Basic fracture mechanics principles and Weibull failure modeling are key factors to consider as well as the role of interfacial stresses. Thus, in order to understand the actual clinical failure mode it is absolutely necessary to consider all the variables listed in the previous paragraph until results from in vitro strength testing can be considered to have any clinical value.

Natural teeth as well as most modern ceramic restorations can be described as layered structures. In the case of teeth the layers are enamel and dentin, while in the case of all ceramics a core ceramic and a porcelain coating. There are also unlayered ceramics in use, but since they are resting on a cement layer and dentin, even they can be described as layered structures. In a study by Jung et al. [35], they determined whether coating thickness and coating/substrate mismatch are key factors in the determination of contact induced damage in clinically relevant bilayer composites. They studied crack patterns in two bilayer systems conceived to simulate crown and tooth structures, at opposite extremes of elastic/plastic mismatch. In one case they looked at porcelain on glass-infiltrated alumina ("soft/hard"), and glass-ceramic on resin composite ("hard/soft"). Hertzian contacts were used to investigate the evolution of fracture damage in the coating layers, as functions of contact load and coating thickness. The crack patterns differed radically in the two bilayer systems: In the porcelain coatings, cone cracks initiate at the coating top surface; in the glass-ceramic coatings, cone cracks again initiate at the top surface, but additionally, upward-extending transverse cracks initiate at the internal coating/substrate interface, where the latter were dominant. This study revealed that the substrate has a profound influence on the damage evolution to ultimate failure in bilayer systems. It was also found that the cracks were highly stabilized in both systems, with wide ranges between the loads to initiate first cracking and to cause final failure, implying damage-tolerant structures. Finite element modeling was used to evaluate the tensile stresses responsible for the different crack types.

In a follow up study, Jung et al.[36] assumed that the lifetimes of dental restorations are limited by the accumulation of contact damage introduced during chewing, and that the strengths of dental ceramics are significantly lower after multi-cycle loading than after single-cycle loading. To test that hypothesis, they looked at indentation damage and associated strength degradation from multi-cycle contacts using spherical indenters in water. They evaluated four dental ceramics: "aesthetic" ceramics porcelain and micaceous glass-ceramic (MGC), and "structural" ceramics--glass-infiltrated alumina and yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) They found that at large numbers of contact cycles, all materials showed an abrupt transition in damage mode, consisting of strongly enhanced damage inside the contact area and attendant initiation of radial cracks outside. This transition in damage mode is not observed in comparative static loading tests,

attesting to a strong mechanical component in the fatigue mechanism. Radial cracks, once formed, lead to rapid degradation in strength properties, signaling the end of the useful lifetime of the material. Strength degradation from multi-cycle contacts were examined in the test materials, after indentation at loads from 200 to 3000 N up to 10^6 cycles. Degradation occurs in the porcelain and MGC after ~ 10^4 cycles at loads as low as 200 N; comparable degradation in the alumina and Y-TZP requires loads higher than 500 N, well above the clinically significant range.

In another study from the same year, Drummond et al. [37] evaluated the flexure strength under static and cyclic loading and determined the fracture toughness under static loading of six restorative ceramic materials. Their intent was primary to compare four leucite ($K_2O•Al_2O_3•4SiO_2$) strengthened feldspathic (pressable) porcelains to a low fusing feldspathic porcelain and an experimental lithium disilicate containing ceramic. All materials were tested as a control in air and distilled water (without aging) and after three months aging in air or distilled water to determine flexure strength and fracture toughness. A staircase approach was used to determine the cyclic flexure strength. The mean flexure strength for the controls in air and water (without aging or cyclic loading) ranged from 67 to 99 MPa, except the experimental ceramic that was twice as strong with mean flexure strength of 191–205 MPa. For the mean fracture toughness, the range was 1.1–1.9 MPa $m^{0.5}$ with the experimental ceramic being 2.7 MPa $m^{0.5}$. The effect of testing in water and aging for three months caused a moderate reduction in the mean flexure strength (6–17%), and a moderate to severe reduction in the mean fracture toughness (5–39%). The largest decrease (15–60%) in mean flexure strength was observed when the samples were subjected to cyclic loading. The conclusion they draw from the study was that the lithium disilicate containing ceramic had significantly higher flexure strength and fracture toughness when compared to the four pressable leucite strengthened ceramics and the low fusing conventional porcelain. All of the leucite containing pressable ceramics did provide an increase in mean flexure strength (17–19%) and mean fracture toughness (3–64%) over the conventional feldspathic porcelain. Further, the influence of testing environment and loading conditions implies that these ceramic materials in the oral cavity might be susceptible to cyclic fatigue, resulting in a significant decrease in the survival time of all-ceramic restorations.

The studies conducted by Jung et al. [35, 36] were followed up by Rhee et al. [38] who approached the onset of competing fracture modes in ceramic coatings on compliant substrates from Hertzian-like contacts. They paid special attention to a deleterious mode of radial cracking that initiates at the lower coating surface beneath the contact, in addition to traditional cone cracking and quasiplasticity in the near contact area. The critical load relations were expressed in terms of well-documented material parameters (elastic modulus, toughness, hardness, and strength) and geometrical parameters (coating thickness and sphere radius). Data from selected glass, Al_2O_3 and ZrO_2 coating materials on polycarbonate substrates were used to demonstrate the validity of the relations. The formulation provides a basis for designing ceramic coatings with optimum damage resistance.

Deng et al. [39] used spherical indenters on flat ceramic coating layers bonded to compliant substrates. They identified critical loads needed to produce various damage modes, cone

cracking, and quasi-plasticity at the top surfaces and radial cracking at the lower (inner) surfaces are measured as a function of ceramic-layer thickness. The characteristic features of these were;

i. Cone cracks initiate from the top surface outside the contact circle, where the Hertzian tensile stress level reaches its maximum [40, 41]. The crack first grows downward as a shallow, stable surface ring, resisted by the material toughness T (K_{IC}), before popping into full cone geometry at load
   $P_C = A(T^2/E)r$
   with $A = 8.6x10^3$ from fits to data from monolithic ceramics with known toughness [42]
ii. Quasiplasticity initiates when the maximum shear stress in the Hertzian near field exceeds Y/2, with yield stress Y ~ H/3 determined by the material hardness H (load/projected area, Vickers indentation)[43].
   The critical load is $P_Y = DH(H/E)^2r^2$
   with D = 0.85 from fits to data for monolithic ceramics with known hardness [42].
iii. Radial cracks initiate spontaneously from a starting flaw at the lower ceramic surface when the maximum tensile stress in this surface equals the bulk flexure strength σ_F, at load
   $P_R = B\sigma_F d^2/\log(E_C/E_S)$
   with d being the ceramic layer thickness and B = 2.0 from data fits to well-characterized ceramic-based bilayer systems [38].

Thus, given basic material parameters, one can in principal make priori predictions of the critical loads for any given bilayer system. Note that P_C and P_Y are independent of layer thickness d, whereas P_R is independent of sphere radius r. These relations, within the limits of certain underlying assumptions, have been verified for model ceramic/substrate bilayer systems [38, 44]. They claimed that these damage modes, especially radial cracking, were directly relevant to the failure of all-ceramic dental crowns. The critical load data were analyzed with the use of explicit fracture-mechanics relations, expressible in terms of routinely measurable material parameters (elastic modulus, strength, toughness, hardness) and essential geometrical variables (layer thickness, contact radius).

Lawn et al. [45] conducted tests on model flat-layer specimens fabricated from various dental ceramic combinations bonded to dentin-like polymer substrates in bilayer (ceramic/polymer) and trilayer (ceramic/ceramic/polymer) configurations. The specimens were loaded at their top surfaces with spherical indenters, simulating occlusal function. The onset of fracture was observed in situ using a video camera system mounted beneath the transparent polymer substrate. Critical loads to induce fracture and deformation at the ceramic top and bottom surfaces were measured as functions of layer thickness and contact duration. Radial cracking at the ceramic undersurface occurred at relatively low loads, especially in thinner layers. Fracture mechanics relations were used to confirm the experimental data trends, and to provide explicit dependencies of critical loads in terms of key variables (material—elastic modulus, hardness, strength and toughness; geometric—layer thicknesses and contact radius). Tougher, harder and (especially) stronger materials show superior damage

resistance. Critical loads depend strongly (quadratically) on crown net thickness. The analytic relations provided a seemingly sound basis for the materials design of next-generation dental crowns.

3.2. Fracture mechanics aspects of dental composites

Dental composite resins consist of ceramic filler particles, usually within a size range of 1-5 μm and mixed with nano-sized (20-40 nm) particles. These inorganic filler particles are silane coated and mixed with a curable monomer to form a viscous paste that can be inserted into a prepared cavity, whereupon it can be shaped and cured. During curing, the silane coated particles bond chemically with the polymer matrix. The filler fraction in dental composites rarely exceeds 60-65 vol-% because of problems with having higher volumes of randomized packed filler particles. Depending on filler size and filler size distribution, it is possible to make different types of dental composites. Since the total filler surface area per gram filler increases as the filler size decreases, finer particles tie up more resin, causing the viscosity of the material to increase fastest with filler fraction of smallest particles. Because of that phenomenon, composites with the finest filler particles tend to contain the lowest filler volume. The modulus of elasticity of a dental composite can roughly be estimated by determine the theoretical modulus of both the series as well as parallel models, and assume that the modulus of the composite for a certain filler fraction falls somewhere between these boundaries.

When the first modern dental composites were introduced during the 60s, it soon became clear that their wear resistance when used on load bearing surfaces was not high enough to be able to resist wear on occlusal surfaces. As a consequence, research performed during the 60s to the 80s focused on finding a solution to the wear problem as well as developing an understanding of the wear mechanism of these materials. During that era, it became clear that some of the key factors associated with composite wear were the quality of the filler matrix bond as well as the filler particle size and distribution. At a symposium supported by 3M in 1984 [46], research findings revealed that the best posterior composites at that time had reached a wear resistance of the commonly used amalgams.

During the research involving wear of composites, researchers had identified that cracks sometimes developed in regions in contact with an opposing cusp. The wear in those regions were often described as two-body wear, while the more general and less dramatic wear occurring on other surfaces were described as a three-body wear caused by abrasive particles sliding over the composite surface during chewing. When it came to the so-called two-body wear, it seemed reasonable to assume that during cusp sliding, micro-cracks could be induced. Another possible wear mechanism induced in the contact region could also be fatigue wear, triggered by a Hertzian failure [47]. In both these cases, microscopic flaws would develop, and these flaws would then contribute to an accelerated wear in these regions. In 1988, Roulet [48] claimed that fractures within the body of restorations and at the margins were a major problem regarding the failure of posterior composites.

However, during the 70s and 80s, the focus on dental composites were related to what clinicians perceived as being the major reasons for failures, which included wear, recurrent caries and discolorations. The notion that flaws were involved in the wear process led Truong and Tyas [49] to determine stable crack growth in dental composites. They did so by use of a double-torsion technique to establish the relationship between the stress intensity factor (SIF) K_I and the crack velocity (v) for commercial and experimental composites. They tested dry, water-saturated and ethanol/water (3:1 v/v) saturated specimens. At a given crack velocity, the difference between the K_I of a dry specimen and that of a water-saturated specimen was attributed solely to the change of Young's modulus caused by the plasticizing effect of water. However, microcracking occurring during immersion in an ethanol/water mixture resulted in an excessive drop of K_I values from the dry state to ethanol/water mixture saturated state for Estilux Posterior and Occlusin samples, while little effect of fluids on K_I could be observed on P10 and P30. The investigators tried to theoretically predict the wear of the composites, based on the assumption that microcracking occurs in the subsurface layer due to cyclic and impact stresses. Based on that assumption, three criteria for good wear resistance would be: (a) high fracture toughness (high critical SIF, K_{IC}) and larger threshold crack length (a_t); (b) small inherent flaw size (a_o) and (c) high crazing stress (σ_c). Based in these assumptions and the results of this study, the wear resistance of tested commercial composites should be: Occlusin > P10 > Estilux Posterior > P30 = Ful-Fil > Profile > Silux --~ Isomolar > Concept.

In a study from 1991, Higo et al. [50] used a fracture mechanics approach to investigate the fracture toughness behavior of three commercial composite resins for dental use named Clearfil photo posterior, P-50 and Occlusin. The outcome of that study was that Occlusin exhibited higher fracture toughness values than any other resin when employing a ring specimen test procedure. However, when an indentation method was used, comparable fracture toughness values for all three resins were produced.

As a fracture mechanics approaches became more popular in attempts to estimating life-times of dental restorative materials, it became important to have available data on the fatigue behavior of these materials. At the end of the 90s, efforts at estimation included several untested assumptions related to the equivalence of flaw distributions sampled by shear, tensile, and compressive stresses. However, environmental influences on material properties were so far not accounted for to any greater extent, and it was unclear if fatigue limits existed. In a study by Baran et al. [51], they characterized the shear and flexural strengths of three resins used as matrices in dental restorative composite materials by use of Weibull parameters. They found that shear strengths were lower than flexural strengths, liquid sorption had a profound effect on characteristic strengths, and the Weibull shape parameter obtained from shear data differed for some materials from that obtained in flexure. In shear and flexural fatigue, a power law relationship applied for up to 250 000 cycles; no fatigue limits were found, and the data thus implied only one flaw population is responsible for failure. Again, liquid sorption adversely affected strength levels in most materials (decreasing shear strengths and flexural strengths by factors of 2–3) and to a greater extent than did the degree of cure or material chemistry.

In a study by Manhart et al. [52], they determined some mechanical properties of three packable composites (Solitaire, Surefil, ALERT), a packable ormocer (Definite), an advanced hybrid composite (Tetric Ceram) and an ionreleasing composite (Ariston pHc) in vitro (Table 3). As seen from that table, the properties of these composites differed significantly, which could be related to differences in filler particle size and shape distributions among the different materials. Their study suggested that fracture and wear behavior of the composite resins would be highly influenced by the filler system. They found that ALERT had the highest fracture toughness value, but also the highest wear rate, which they related to the fiber like particles used in that material. Overall, Surefil demonstrated good fracture mechanics parameters and low wear rate, which they suggested could be related to their more particle shaped filler particles. This study suggested that fracture and wear behavior of the composite resins are highly influenced by the filler system.

Composite material	Flexural strength (MPa)	Flexural modulus (GPa)	Fracture toughness K_{IC} (MN $m^{-1/2}$)	Mean wear rate (μm^3 $cycle^{-1}$)
Solitare	81.6 (10.0)	4.4 (0.3)	1.4 (0.2)	1591
Definite	103.0 (19.9)	6.3 (0.9)	1.6 (0.3)	2763
Surefil	132.0 (14.3)	9.3 (0.9)	2.0 (0.2)	3028
ALERT	124.7 (22.1)	12.5 (2.1)	2.0 (0.2)	8275
Tetric Ceram	107.6 (11.4)	6.8 (0.5)	2.0 (0.1)	5417
Ariston pHc	118.1 (10.5)	7.3 (0.8)	1.9 (0.2)	7194

Table 3. Some properties of six dental composite materials [52].

Considering the importance of being able to perform life-time predictions of dental composites, McCool et al. [53] , continued their research from 1998 [51], by comparing the lifetime predictions resulting from two methods of fatigue testing: dynamic and cyclic fatigue. To do so they made model composites, in which one variable was the presence of a silanizing agent. They tested their specimens in 4-point flexure, using a cyclic fatigue frequency of 5 Hz, while their dynamic fatigue testing spanned seven decades of stress rate application. Data were reduced and the crack propagation parameters for each material were calculated from both sets of fatigue data. These parameters were then used to calculate an equivalent static tensile stress for a 5-year survival time. The 5-year survival stresses predicted by dynamic fatigue data were approximately twice those predicted by cyclic fatigue data. In the absence of filler particle silanization, the survival stress was reduced by half. Aging in a water-ethanol solution reduced the survival stresses by a factor of four to five. One of the conclusions drawn from this study is that cyclic fatigue is a more conservative means of predicting lifetimes of resin-based composites.

The notion that there is a correlation between wear resistance and fracture toughness was to some degree rejected by Ruddel et al.[54]. In their study they produced pre-polymerized fused-fiber filler modified composite particles and determined their effectiveness by incor-

porating these fibers into composites. The results revealed that these particles decreased both flexural strength and fracture toughness, but improved wear performance. The SEM evaluations did not suggest that porosities had been incorporated during particle incorporation. Instead, fractures were transgranular through the reinforcing particles. Microscopic flaws observed in the new particles most likely explain the lower strength and toughness values. This study is important, because it shows that a composite with improved wear resistance could also suffer from an increase in fracture risk.

During the past 10 years, it has become clear that fracture is a major reason for clinical failure of dental composites. Many clinical fractures are likely to be preceded by slow sub-critical crack propagation. To study the slow sub-critical crack propagation, Loughran [55] used notched composite (Z100, 3M ESPE) specimens and fatigued them in a four-point bending test using a load cycle at 5 Hz between 25 and 230 N until failure. Displacement and load were recorded during the fatigue tests and used to derive crack propagation based on beam-compliance. What they found was that the number of cycles until failure ranged between 34 and 82,481. In the last 1500 cycles prior to final fracture, the beam compliance increased consistently, indicating sub-critical crack propagation. From the compliance change they calculated that the crack length increased 8% (77 ± 14 μm) before final failure. The crack growth rate during sub-critical crack propagation was determined as a function of the stress intensity for the last 1500 cycles before fracture. The importance of this study was that they found that the fatigue lifetime varied widely, and that stable crack growth existed prior to fracture consistently. This consistency allowed formulation of stress-based crack propagation relationships that can be used in concert with numerical simulations to predict composite restoration performance. The large variation found for specimen lifetime was attributed to the initiation process that precedes sub-critical crack propagation.

As mentioned earlier, during the early 80s, dentists regarded poor wear resistance tendency to be associated with recurrent caries and restoration discolorations as the key shortcomings with dental composites. Today, that perception has changed quite considerable. By improved filler technology and silanization methods, the poor wear resistance is no longer a major clinical problem. Improved adhesives, now making it possible to bond composites to both enamel and dentin, have decreased the risk for recurrent caries. The use of more stable chemicals and smoother composite surfaces caused by the use of finer filler particles has decreased the magnitude of restoration discolorations. In other words, what were regarded as major shortcomings with posterior composites are no longer regarded as major weaknesses. Of course, these shortcomings have not yet been completely eliminated, so there is still room for improvements. However, as the composites have been improved, another shortcoming has been identified as now being the biggest problem, namely fractures[48]. In a recently published clinical study [56], in which two composites were evaluated over a 22-year period, the authors claimed that the most common reason for failures of posterior composites were fractures. That study suggests that further understanding of the fracture mechanical behavior of dental composites is needed.

3.3. Fracture mechanics aspects of cements and adhesives

In order to attach restorations such as composite fillings, inlays/onlays, crowns and bridges, different cements/adhesives have been used in dentistry through the years. The oldest but still used cement is the zincphosphate cement, which was introduced about 150 year ago and consists mainly of a zincoxide powder mixed with phosphoric acid. During setting, that cement goes through an acid-base reaction during which a salt and water is formed. The way this cement works is simply by etching the surfaces of the tooth and the surface of the restoration the cement is in contact with, a process that occurs as the cement sets, whereupon zincphosphate crystallites precipitate into the etched surface regularities as the cement sets. With that mechanism a mechanical interlocking is established, explaining the retention of the cemented restoration.

In addition to the zinc phosphate cement, other cements such as silicate, zincsilico phosphate, polycarboxylate and glass ionomer cements have been used. In the case of the silicate and zincsilico phosphate cements, phosphoric acid is used in both cases, while the powders of these two cements are either a silicate glass powder or a mixture of that powder with a zinc phosphate powder. When it comes to the polycarboxylate and the glass ionomer cements, the powders are either the zinc oxide powder or the glass powder used in the silicate cement, while the acid has been replaced with a polyalceonic acid. The polyalkeoinic acid, often polyacrylic acid, is capable of reacting with the powder through an acid-base reaction, but also with the dentin or enamel surface. During that reaction the $–COO^-$ of the polyalkeonic acid can interact with ions such as the Ca^{2+} present in the tooth surface and form some ionic interaction. Compared to the zinc phosphate and silicate cements, the polycarboxylate and glass ionomer cements were introduced to dentistry during the 60s and the 70s. Regarding the ability to bond to hard tooth tissues, it is generally assumed that zinc phosphate, zincsilico phosphate, and silicate cements only bond via micromechanical retention, while polycarboxylate and glass ionomer cements bond both via a micromechanical retention as well ionic surface interaction.

The idea to develop some kind of chemical bond to dental hard tissues was however introduced before the zincpolycarboxylate and glass ionomer cements had been invented. The first idea to use some kind of chemical interaction to form a bond to the hard dental tissues was introduced during the late 40s when Hagger [57] suggested that a molecule that had a phosphate group capable of interaction with Ca^{2+} at the tooth surface and a methylmethacrylate group capable of forming a covalent bond to a curing methacrylate based filling materials could form such a bond. Unfortunately, the molecule Hagger used to achieve such a bond did not show to be very efficient. However, when Buonocore in 1955 [58] explored the possibility to first etch the enamel surface with a phosphoric acid, then rinse and dry and coat the acid roughened surface with a curable resin, it became possible to achieve a predictable bond to enamel.

Buonocore's idea was not widely accepted initially, because dentists feared that the phosphoric acid, particularly if it came in contact with exposed dentin surface, would cause pulp irritation and eventually pulp death. Such pulp reactions were known to occur,

particularly when the more slow setting silicate cement was used. As a consequence it would take several years until Buonocore's acid-etch approach took off. A major contributor for teaching dentists how to use enamel etching and composite resins was 3M, who during the 60s had expanded their products to dentistry.

To spread the usage and the acceptance of enamel etching and resin bonding as well as their composite resin, 3M sponsored a symposium entitled "The Acid Etch Technique" in 1975. The presentations presented at that symposium were published in a book [59] that was then widely distributed by representatives for the company. By having prominent researchers presenting papers related to the acid etch technique, a lot of misperceptions could be eliminated and the enamel etch technique became generally accepted [60]. When it came to testing enamel bonding, most in vitro studies relied on morphology achieved by use of SEM and different strength tests of which shear bond testing soon became the most popular.

Even though enamel bonding was a major advance in dentistry, the ability to bond to dentin was not resolved when enamel bonding took off. Because most surfaces exposed during tooth preparations of cavities and crowns consist of dentin, a reliable dentin bonding was still needed in order to truly bond different restorative materials to dentin. However, dentin bonding was much more complicated to achieve than enamel bonding. In contrast to enamel, dentin is a living tissue and therefore much more demanding than enamel when it came to biocompatibility of chosen materials. Besides, dentin contains much more water, making it difficult to adapt more or less hydrophobic materials to the dentin surface.

Parallel to these events, Bowen had already during the 60s initiated research to develop resin systems capable of bonding to cut dentin surfaces [61]. The basic principle behind his ideas was that the adhesives should contain a reactive group capable of reacting with Ca-ions present on the tooth surface, and then react with the resin when the resin cured. When these adhesives, often referred to as the first generation of dentin adhesives were explored, it soon became clear that a cut dentin surface was coated with a so called smear layer. That smear layer consisted of a few microns thick layer of smeared collagen in which fractured hydroxyapatite crystallites were embedded. It was soon clear that the first generation of adhesives developed a weak bond to the tooth surface, and that the bond was weak and worked for a short time period only, mainly because the bonds formed to the smear layer, or the bonds between the smear layer and the dentin were too weak to resist loading.

During the 70s, the dental community discussed the effect the smear layer had on bonding and whether or not it should remain on the dentin surface. Some researchers viewed it as beneficial, since the vital dentin channels were sealed off, decreasing the risk of pulp irritations caused by the restorative material. As a consequence, somewhat more acidic adhesive systems were developed, capable of removing some of the smear layer but retaining some smear serving as protective layer. The adhesives that fell into this class are often referred to as the 2nd generation adhesives.

At the end of the 70th, a major break-through occurred. That break-through consisted of a clinical study performed by Fusayama at al. [62], in which they claimed that by etching both

enamel and dentin, they were able to bond composites to dentin without having any problem with pulps responding to the etching procedure. There is no doubt that Fusayama et al.'s finding was looked upon with enormous skepticism. Their explanation that resin infiltrated the tubules and thereby formed resin tags that contributed to the retention was also questioned. It was first when Nakabaiashi [63] came out with his hybridization explanation, suggesting that the resin infiltrated the etched dentin surface and formed a hybrid layer consisting of partly dissolved dentin, as dentin etching started to become accepted .

These two studies[62, 63] opened the door for more aggressive dentin etching resulting in the 3rd generation adhesives. Etching dentin with phosphoric acid was still not the general trend. Instead, weaker conditioners such as EDTA and citric acids were used[64]. However, at the end of the 80th, some bonding systems had occurred on the market that used the same etchant for both enamel and dentin. The success of these adhesives, the so called 4th generation adhesives, took of during the early 90s, when both Kanca [65] as well as Gwinnett [66] in two independent studies claimed that by leaving the dentin moist before priming, they could better infiltrate the collagen layer with the primer and thereby achieve better bonding to dentin.

Simultaneous with these trends related to bonded composite, it had also been noticed that by etching the surface of ceramic restorations located at the dentin surface with hydrofluoric acid and then silane coat the etched surface, it was possible to bond ceramic restorations to tooth surfaces. Such an approach resulted in a significantly lower risk of ceramic fracture than compared to the use of more traditional cements, including polycarboxylate and glass ionomer cements. By use of the information presented under the ceramic section in this chapter, it is quite easy to explain why resin bonded ceramics performed so well by considering fracture mechanics. In the case of the more traditional cements, they can be described as having brittle properties with limited ability to form strong bonds to the ceramic surface. In the case of the phosphoric acid based cements they did not form any strong bonds to the tooth surface neither. By realizing that even a ceramic restoration can flex during chewing, one can visualize the development of shear stresses at the ceramic-cement interface, and that these stresses can trigger a crack growth along the ceramic-cement interface. In the case of the resin bonded ceramics, the shrinkage of the resin cement initially induced some compressive stress in the ceramic surface adjacent to the resin cement. If a crack propagates to the resin interface in such a case, the more ductile nature of the resin cement will not as easily allow the crack to propagate along the ceramic-cement interface. Besides, after the load has been removed, the resin will because of its polymerization shrinkage, try to force the fractured ceramic in contact with its fractured surfaces. Thus, in this case, a ceramic fracture may occur, but in contrast to a fracture in a ceramic cemented with more traditional cement, one may not end up with a detectable catastrophic failure.

From a fracture mechanics point of view, there is no doubt that the adhesive joint is the most challenging region. The reason relates simply to practical problems such as minimizing the incorporation of defects in this region. In addition, the fact that the adhesive shrinks and

induces shrinkage stresses between the tooth and the adhesive, as well as between the adhesive and the restorative material, does not make the situation manageable, which is further complicated by differences in mechanical/physical properties of the different materials forming a joint. In the following section we will approach the adhesive joint in an attempt to identify different challenges associated with this region.

When it comes to the failure mechanism at the dentin resin interface, there are certain questions that need to be addressed. These questions include: (1) does failure at the human dentin-resin interface occur by a cohesive or an adhesive mechanism? (2) is the failure mechanism accompanied by a plastic deformation, and if so how important is it? To address these questions, Lin and Douglas [67] performed a computational analysis and fractography of two different bonding systems: Scotchbond- (SB2) and Scotchbond-Multipurpose (SBM). The difference between these two systems is that SB2 consists of a mixture of primer and a so called bonding resin, while SBM uses the same primer and bonding resin, but in contrast to SB2 they are placed as separate systems on the dentin surface. According to their estimates, the dentin-resin interracial fracture toughness (G_{IC}), for the SB2 and for the SBM were 30.22 ± 5.61 and 49.56 ± 7.65 J m^{-2}, respectively, which were significantly different ($p < 0.01$). Both SB2 and SBM interfaces with dentin displayed significant degrees of plasticity (0.15 and 0.19) which were beneficial to crack resistance. Thus, correcting for the plasticity, the G_{IC} for SB2 and for SBM increased to 42.83 ± 7.75 and 74.97 ± 10.47 J m^{-2}, respectively. The fractography of the two systems reflected these numeric differences. SB2 showed largely interfacial adhesive failure, while SBM showed adhesive-cohesive failure with occasional dentin adhesions attached to the composite interface and vice versa.

In another study, Toparli [68] determined the reliability and validity of the adhesive bond toughness of dentin/composite resin interfaces from the standpoint of fracture mechanics. The fracture toughness (K_{IC}) and fracture energy (J_{IC}) values of two different composite resins (Brilliant Dentin and P50) were determined. The fracture toughness and energy values obtained experimentally for Brilliant Dentin were found to be higher than those for P50. It was seen that calculated J values (J_{adh} and J_{res}) changed with the crack length; but the effective fracture energy (J_{eff}) was independent of the crack length, as expected. The applied fracture energy (J_{appl}) and effective fracture energy (J_{eff}) are considerably smaller than the experimentally determined J_{IC} values of composite resins. The important finding was that the bonded interface tends to produce microscopic flaws which could act as critical stress risers promoting interfacial failures. The initiation and propagation of such flaws under the mastication forces can be followed by fracture toughness (K_{IC}) or fracture energy (J_{IC}) in linear elastic fracture mechanics (LEFM).

The effect of crack growth at a resin bonded metal interface after storage in water was studied by Moulin [69], who found that the adherence energy dramatically decreased with time in water. The slope of the regression straight line appeared to be a good criterion for evaluating the durability of the alloy/adhesive interface. The study revealed the importance of silica coating the metal surface and, especially, the effectiveness of the Rocatec system upon the degree of hydrolytic degradation. The study showed how the development of cracks depends upon surface treatment.

Adhesion at the titanium–porcelain interface using a fracture mechanics approach has also been used to investigate the bonding mechanism of such systems [70]. In that study they used specimens of five different titanium–porcelain and one base metal–porcelain bonding systems on which they performed a four-point bending interfacial delaminating test. The pre-cracked specimen was subjected to load and the strain energy release rate (G) was calculated from the critical load to induce stable crack extension in each system. The strain energy release rate of titanium–porcelain with a Gold Bonder interface layer was highest among the five different systems. No attempt was made to explain the experimental findings.

In two studies by Ichim et al. [71, 72] they looked at a typical non-carious cervical lesion, a so called abfraction, treated with a glass ionome or a combination of glassionomer and composite. The approach they used was that they used a nonlinear fracture mechanical approach simulated by use of FEA. They used a novel Rankine and rotating crack model to trace the fracture failure process of the cervical restorations. The approach involves an automatic insertion of an initial crack, mesh updating for crack propagation and self contact at the cracked interface. The results were in good agreement with published clinical data, in terms of the location of the fracture failure of the simulated restoration and the inadequacy of the dental restoratives for abfraction lesions.

In their second study [72] they investigated the influence of the elastic modulus (E) on the failure of cervical restorative materials and tried to identify an E value that would minimize mechanical failure under clinically realistic loading conditions. What they found was that the restorative materials currently used in non-carious cervical lesions are largely unsuitable in terms of resistance to fracture of the restoration. They suggested that the elastic modulus of such a material should be in the range of 1 GPa rather than several GPa that is usually the case.

Despite an obvious advantage to approach adhesives and their performance from a fracture mechanics point of view, traditional bond studies usually focus on bond strength values. By comparing such strength values, it is noticed that large variations exist among different reports. These variations are due to differences among operators, but also on the day a certain tester performed a test. The standard deviation is 25-50 % of the mean value, which suggests that defects present in the adhesive region may be of a bigger concern than the true adhesive strength.

In an attempt to resolve the questions related to the large variability in strength values and their clinical meaning, The Academy of Dental Materials at their annual meeting in 2010, focused that meeting on the value of bond strength measurements. In one presentation, Scherrer et al. [73] presented a literature search based on all dentin bond strength data obtained for six adhesives evaluated with four tests (shear, microshear, tensile and microtensile) and critically analyzed the results with respect to average bond strength, coefficient of variation, mode of failure and product ranking. The PubMed search was carried out for the years between 1998 and 2009. The six adhesive resins that were selected included three step systems (OptiBond FL, Scotch Bond Multi-Purpose Plus), two-step (Prime & Bond NT, Single Bond, Clearfil SE Bond) and one step (Adper Prompt L Pop). By pooling the results from

the 147 references, it was revealed an ongoing high scatter in the bond strength data regardless which adhesive and which bond test was used. Coefficients of variation remained high (20–50%) even with the microbond test. The reported modes of failure for all tests still included a high number of cohesive failures. The ranking of the adhesives seemed to be dependent on the test method being used. The scatter in dentin bond strength data, independent of used test, confirmed Finite Element Analysis predicting non-uniform stress distributions due to a number of geometrical, loading, material properties and specimens preparation variables. The study reopened the question whether an interfacial fracture mechanics approach to analyze the dentin–adhesive bond would not be more appropriate for obtaining better agreement among dentin bond related papers.

In another paper presented at that meeting, Soderholm [74] emphasized the benefits of using fracture mechanics approaches when it comes to studying dental adhesives. In his review, different general aspects of fracture mechanics and adhesive joints were reviewed, serving as a foundation for a review of fracture toughness studies performed on dental adhesives. The dental adhesive studies were identified through a MEDLINE search using "dental adhesion testing AND enamel OR dentin AND fracture toughness" as search strategy. The outcome of the review revealed that fracture toughness studies performed on dental adhesives are complex, both regarding technical performance as well as achieving good discriminating ability between different adhesives. The review also suggested that most fracture toughness tests of adhesives performed in dentistry are not totally reliable because they usually did not consider the complex stress pattern at the adhesive interface. However, despite these limitations, the review strongly supports the notion that the proper way of studying dental adhesion is by use a fracture mechanics aproach.

In a study by Howard and Soderholm [75] they used a fracture mechanics approach previously described by Pilliar and Tam [76-80] to test the hypothesis that a self-etching adhesive is more likely to fail at the dentin adhesive interface than an etch-and-rinse adhesive. What they found was that the fracture toughness values (K_{IC}) of the two adhesives were not significantly different. The rather high frequency of mixed failures did not support the hypothesis that the dentin-adhesive interface is clearly less resistant to fracture than the adhesive–composite interface. The finding that cracks occurred in 6–8% in the composite suggests that defects within the composite or at the adhesive–composite interface are important variables to consider in adhesion testing.

In a recently published study by Ausello et al. [81], they used FEA and fatigue mechanic laws to estimate the fatigue damage of a restored molar. The simulated restoration consisted of an indirect class II MOD cavity preparation restored with a composite. Fatigue simulation was performed by combining a preliminary static FEA simulation with classical fatigue mechanical laws. It was found that regions with the shortest fatigue-life were located around the fillets of the class II MOD cavity, where the static stress was highest.

From the above papers, it becomes clear that adhesion tests utilized in dentistry are unable to separate the effects of adhesive composition, substrate properties, joint geometry and

type of loading on the measured bond strength. This makes it difficult for the clinician to identify the most suitable adhesive for a given procedure and for the adhesive manufacturer to optimize its composition. To come to grip with these challenges, Jancar [82] proposed an adhesion test protocol based on the fracture mechanics to generate data for which separation of the effect of composition from that of the joint geometry on the shear (τ_a) and tensile (σ_a) bond strengths was possible for five commercial dental adhesives. The adhesive thicknesses (h) used varied from 15 to 500 μm, and the commercial adhesives had fracture toughness values (K_{IC}) ranging from 0.3 to 1.6 MPa $m^{1/2}$. They used double lap joint (DLJ) and modified compact tension (MCT) specimens which were conditioned for 24 h in 37°C distilled water, then dried in a vacuum oven at 37°C for 24 h prior to testing. Both τ_a and σ_a increased with increasing adhesive thickness, exhibiting a maximum bond strength at the optimum thickness (h_{opt}). For $h < h_{opt}$, both τ_a and σ_a were proportional to h, and, above h_{opt}, both τ_a and σ_a decreased with $h^{-4/10}$ in agreement with the fracture mechanics predictions. Hence, two geometry-independent material parameters, Ψ and (H_c/Q), were found to characterize τ_a and σ_a over the entire thickness interval. The results seem important, because it suggests that the adhesion tests currently used in dentistry provide the geometry dependent bond strength, and such data cannot be used either for prediction of clinical reliability of commercial dental adhesives or for development of new ones. Instead, the proposed test protocol allowed one to determine two composition-only dependent parameters determining τ_a and σ_a. A simple proposed procedure can then be used to estimate the weakest point in clinically relevant joints always exhibiting varying adhesive thickness and, thus, to predict the locus of failure initiation. Moreover, this approach can also be used to analyze the clinical relevance of the fatigue tests of adhesive joints.

In a recent paper by Kotousove [83] a conceptual framework utilizing interfacial fracture mechanics and Toya's solution for a partially delaminated circular inclusion in an elastic matrix was used , which can be applied (with caution) to approximate polymer curing induced cracking about composite resins for Class I cavity restorations. The findings indicated that: (I) most traditional shear tests are not appropriate for the analysis of the interfacial failure initiation; (II) material properties of the restorative and tooth material have a strong influence on the energy release rate; (III) there is a strong size effect; and (IV) interfacial failure once initiated is characterized by unstable propagation along the interface almost completely encircling the composite. The importance of this study is that it analyses the reliability of composite Class I restorations and provides an adequate interpretation of recent adhesion debonding experimental results utilizing tubular geometry of specimens. The approach clearly identifies the critical parameters including; curing strain, material module, size and interfacial strain energy release rate for reliable development of advanced restorative materials.

In a similar approach, Yamamoto [84] calculated stresses produced by polymerization contraction in regions surrounding a dental resin composite restoration. Initial cracks were made with a Vickers indenter at various distances from the edge of a cylindrical hole in a soda-lime glass disk. Indentation crack lengths were measured parallel to tangents to the hole edge. Resin composites (three brands) were placed in the hole and polymerized (two light irradiation protocols) at equal radiation exposures. The crack lengths were remeasured

at 2 and 10 min after irradiation. Radial tensile stresses due to polymerization contraction at the location of the cracks (σ-crack) were calculated from the incremental crack lengths and the fracture toughness K_C of the glass. Contraction stresses at the composite–glass bonded interface (σ-interface) were calculated from σ-crack on the basis of the simple mechanics of an internally pressurized thick-walled cylinder. The greater the distance or the shorter the time following polymerization, the smaller was σ-crack. Distance, material, irradiation protocol and time significantly affected σ-crack. Two-step irradiation resulted in a significant reduction in the magnitude of σ-interface for all resin composites. The contraction stress in soda-lime glass propagated indentation cracks at various distances from the cavity, enabling calculation of the contraction stresses.

4. Conclusion

By reviewing enamel, dentin and their interfacial bond, it is obvious that the tooth evolved in such a way it would be able to function in an optimal way without fracturing. With the sophisticated structure of both enamel and dentin, it becomes quite clear that existing man made restorative materials are far from optimal in comparison to the biological hard tissues. The crack growth risk in ceramics needs to be reduced, something that can be achieved by use of fracture tough ceramics such as alumina and zirconia. Unfortunately, as shown in Lawn et al.'s[45] paper, rather extensive removal of existing tooth structure needs to be performed in order to minimize future failures. Such an approach, though, does not make sense if one considers that a more sophisticated material is removed in order to replace it with an inferior material.

When it comes to dental composites, we have now reached a point when fractures of composites are being judged as the most common reason for composite failures [48, 56]. To come to grip with that problem, our understanding of the fracture mechanics of dental composites needs to be improved. The same is true regarding cements/adhesives. The particulate filled resins we are now using are rather primitive when compared to both enamel and dentin. However, it seems as this group of materials has the highest chance to evolve and approach the properties of enamel and dentin.

By looking at dentistry from a fracture mechanics point of view, it becomes quite clear that traditional dentistry suffer from some major processing problems. The first problem is that restorations are individual units that differ in shape and size. These different sizes and shapes result in different levels and locations of localized stresses. The second problem is that restorations are placed by individual dentists working under different conditions and introducing different amounts and types of flaws during the different dental procedures. Considering that the theoretical strength is several magnitudes stronger than the real strength values due to the presence of defects in materials suggest that processing defects, located in a material or at an interface is a significant dental problem. The third problem is partly self-inflicted. During dental education, students learn to copy the anatomy of natural teeth. The pits and fissures present in natural teeth act naturally as stress concentrators, but because of the sophisticated structure of a substrate such as enamel, such pits and fissures

may in fact act as crack stoppers. Take for example a crack that might propagate along a cusp toward the central fissure. When that crack reaches the bottom of that fissure, the thickness of the enamel decreases and one can assume that the crack will not continue to propagate up along the other cusp with increasing enamel thickness. In the case of a man-made crown or filling, the fissure will not serve the same protective purpose. However, because dental students are trained to reproduce the sharp anatomic details, sharp anatomic fissures are often regarded as a sign of good competence, while in reality such details will rather facilitate crack growth.

Based on the information provided in this chapter it seems reasonable to suggest that future dental students should receive more training in fracture mechanics in order to better understand how handling and design may affect the final outcome of a restorative procedure Besides, with such a knowledge they would be able to communicate better with other scientists and thereby facilitate the development of better restorative materials.

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