

**An in vitro analysis of the behaviour of an alumina  
based dental all-ceramic restorative system subjected  
to occlusal loads.**

by

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requirements for the degree of  
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## **Dedication**

To my wife

*Kathryn*

and my children

*Joanne, Sara and Christopher*

## Abstract

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The commercial introduction of glass infiltrated slip cast alumina ceramic technology as a core dental ceramic for crowns and bridges began in 1990. The system known as In-Ceram® provided a metal free ceramic core with a flexural strength that was reported to be between 400 and 600 MPa. The first aim of this study was to investigate in vitro fractures for this type of crown that occurred from high localized occlusal loading. The second aim was to propose suitable guidelines for the construction of crowns that are more crack resistant when loaded by a spherical object such as an opposing cusp.

The review of the literature identified Young's modulus ( $E$ ) as an important aspect for the behaviour of a brittle solid when it is exposed to load. Young's modulus is involved in the total energy of the solid, its free surface energy, its toughness, parameters involved in contact areas, stresses from indenters on those contact areas and at the interface between two materials with different moduli. This mismatch was to become more important when Finite Element Analysis was applied to the test results. A second physical property that has importance in bilaminar materials is Poisson's ratio. Less important than Young's modulus, Poisson's ratio still provides some of the answers to crack propagation at an interface, highlighted by the description of Dundurs parameters (Mencík 1996). The use of Weibull statistics provided information concerning reliability for different core to veneer thickness ratios in the experimental models and crown designs.

The first experimental part was to establish a crown model suitable for investigation. The porcelain jacket crown restoration for an upper incisor was chosen because it provided the best opportunity to vary the construction parameters of the crown without sacrificing the aesthetics. The results from this initial in vitro study established the thickness ranges for the standard design of these crowns, the loads that could be expected and the two types of fracture patterns that might occur when the bilaminar system is loaded via a hard steel ball indenter in the middle of the palatal surface.

The data from chapter three were then compared with two basic bilaminar and monolithic disc designs to establish the relevance of the dimensions of the ISO standard test specimen design for flexural strength. The results of this chapter confirmed that the loads were in the same range for the bilaminate discs, and that the flexural strengths for the two component parts were in agreement with other known results. The use of Finite Element Analysis (FEA) was employed to provide an additional method for testing the model to establish areas when principal stresses might lie and how they were distributed. It was found that the ISO flexural strength test was not ideal for testing bilaminates because it did not account for the large mismatch in Young's modulus between the two component

materials. The FEA revealed an unexpected increase in tensile stress on the bottom surface of the In-Ceram when it was the bottom layer in a bilaminate. The FEA did confirm the difference between a flat punch and a ball indenter for stress intensities around the contact area with the ball indenter producing higher stresses for a given load.

The effect of supporting the ceramic system with different layer thicknesses was then tested. This was the second step in gaining information about the design criteria for a more crack resistant crown. The physical tests were made using a 5 mm diameter steel ball on discs that were supported on a flat steel base. The FEA results indicated again that the tensile stresses on the bottom surface of the In-Ceram were higher when it was used in combination with the Vitadur Alpha veneer, but that the level of stress was much lower than that experienced in the unsupported state. An FEA model using the properties of dentine as the support base also confirmed a significant influence from this medium. As the Young's Modulus of the support and the relative thickness of the In-Ceram increased, the tensile stress on the bottom surface decreased. As the support increased and the veneer thickness decreased, the stresses on the top surface of the Vitadur Alpha increased, and the nature of the damage changed from a simple cone crack followed by radial cracking, to a complex cone crack and crushing of the veneer porcelain. The high pressures in the thinner veneer layer crushed the contact area. The radial cracks followed. The higher pressures on the thinner veneer layers caused those samples to break explosively. The stresses at the surface were calculated using a novel analytical program called "Elastica".

The information was then applied to the design of the crowns, where layer thicknesses varied slightly as the crowns were made under normal dental laboratory conditions. Statistically the modified crowns performed better under the loading conditions used. The results from the modified crown tests confirmed the data in chapter five, where it was shown that increasing the In-Ceram thickness and decreasing the Vitadur Alpha thickness at the point of contact could provide a more reliable crown during function.

The implications from this in vitro study are:

1. There is a significant influence in the type of support beneath these crowns at the point of occlusal loading.
2. The surface of the core preparation may have a significant influence on the fracture resistance. This is with respect to the adhesion of the luting agent to the tooth surface irrespective of whether the actual adhesion is to dentine or an artificial core such as gold, ceramic or resin.
3. The overall design of the core layer with particular respect to the area beneath the contact area has been shown to influence the initiation of fractures and in some

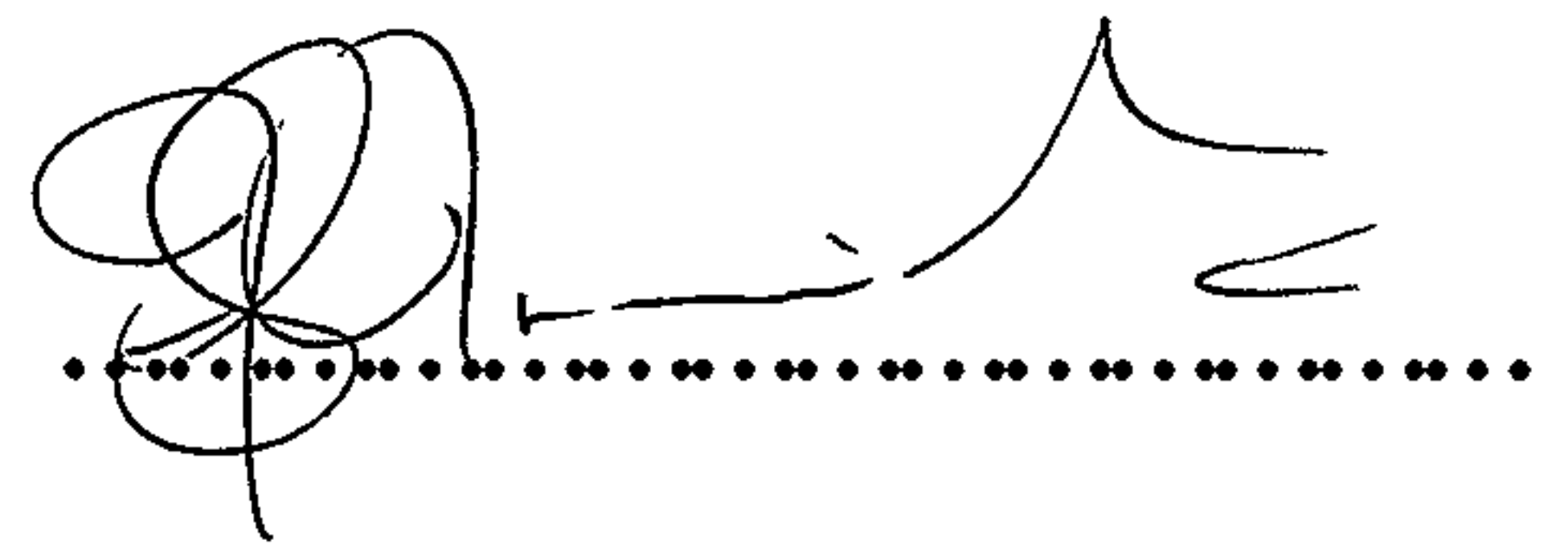
respects the propagation of those fractures, although it is recognized that once the porcelain has fractured the crown is, for all intents and purpose, a failure.

## Declaration

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### Candidate's Certificate

This is to certify that the work presented in this thesis was carried out by the candidate in the Discipline of Biomaterials Science, Faculty of Dentistry, University of Sydney, and has not been submitted to any other university or institution for a higher degree.

A handwritten signature in black ink, consisting of a large, stylized 'J' followed by a series of loops and a final flourish. The signature is written above a horizontal dotted line.

James Ironside, 2001

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I would like to acknowledge and thank the following people for their assistance, support and guidance in the completion of this thesis.

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## Publication and Scientific Meetings

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### Journal Publications

Ironside JG The In-Ceram metal-free ceramic system for Fixed Prosthodontics. Australian Prosthodontic Journal 1993; 7: 55 – 61.

Ironside JG Light Transmission of a Ceramic Core Material used in Fixed Prosthodontics. Quint. Dent Tech. 1993; 16: 103 -106

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Ironside JG and Styles GT. Core framework design to change resistance to crack initiation. J. Dent. Res. 1995b; 74: 522, Abstr. No. 975.

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Ironside JG Proos K and Swain MV. Finite element analysis of bilaminar dental ceramic discs. To be Published 2001

Ironside JG. The effect of porosity on the flexural strength of In-Ceram. J. Dent. Res. 1995a; 74: 522, Abstr. No. 973.

### Presentations

#### *International Meetings*

**Ironside JG** Strengthening and Toughening Philosophies for the New Ceramic Systems. The IADR 72<sup>nd</sup> General Session and Exhibition, March 10 –13, 1994, Seattle, Washington State, USA.

**Ironside JG** The effect of Porosity on the Flexural Strength of In-Ceram. The IADR 73<sup>rd</sup> General Session and Exhibition, June 28 – July 1, 1995, Singapore.

**Ironside JG** Crack Initiation and Propagation in an Alumina Based Ceramic Crown. The IADR 74<sup>th</sup> General Session and Exhibition, March 13 –17, 1996, San Francisco, CA, USA.

#### *National Meetings*

**Ironside JG** Biomechanics and the New Ceramic Restorations. The IADR (ANZ Division) 37<sup>th</sup> annual meeting, August 24 – 28, 1997, University of Otago, Dunedin, New Zealand.

**Ironside JG** and Watson TF. Damage Following Finishing Procedures: A Confocal Microscope View. The IADR (ANZ Division) 34<sup>th</sup> annual meeting, September 25 – 28 1994, Melbourne, Australia.

Swain MV, **Ironside JG** and Mori T. A Simple Precise Method for Measuring the Mechanical Properties of Teeth. The IADR (ANZ Division) 33<sup>rd</sup> annual meeting, September 1993, Perth, Australia.

# CHAPTER 1

---

## Introducing Ceramics to Dentistry

---

### 1.1 Introduction

#### 1.1.1 The role of Ceramics in Dentistry

Ceramics have aroused the interest of man for thousands of years. The word ceramic comes from Greek and literally means "earthen". McLean (1979) provides a concise history of ceramic use in modern dentistry. The use of ceramics for the restoration of teeth has been a part of dentistry's modern period of evolution. This period started in the late seventeenth century but major advances have mainly come about this century. The demand for aesthetic restorations led to improvements in ceramic formulation and firing techniques. McLean (1979) has largely summarized the earlier types of ceramic systems. Essentially the type of ceramic restoration to be discussed will be what has been traditionally known as the Porcelain Jacket Crown, often referred to simply as a "PJC". These crowns are quite distinct from Metal-Ceramic Crowns (MCC) because there is no underlying metal framework to provide a stiff and tough foundation for the porcelain that provides the colour and form that in turn make the crown look and function like a natural tooth.

However, these types of crowns do require some form of toughened dental ceramic to support the outer glassy veneer that provides the final colour and form. The major strength-limiting property of dental ceramics has been the presence of surface microcracks (Southan et al 1972 and 1973). The original Porcelain Jacket Crown used a type of porcelain that was higher than normal for its alumina content (McLean 1965, 1987), the alumina being recognized as a material that would increase the tensile strength. These materials were known as "Aluminous" porcelains and provided a stronger core layer on which the veneering porcelains could be fired. Like most toughening procedures in ceramic materials, the inclusion of high amounts of alumina increased the opacity, which meant that the core material needed to be as thin as possible to allow maximum thickness of the veneer to reproduce the aesthetics of natural teeth. This limited their application as they were relegated to use where the forces of occlusion were low, such as the incisor teeth. Conditions such as bruxism also precluded their use. Examples of other systems

that will be described in chapter two, involved monolithic construction of toughened glassy ceramics.

The application of the crown to the tooth involves significant preparation of the remaining tooth to provide sufficient space to construct the crown, allowing it to be blended back into the dentition. An example of typical crown preparations and the final result appear in figure 1.1 below. These crown preparations were prepared by myself and the technical work was constructed by Mr. Paul Wagner, Wagner King Ceramics, Sydney. Diagrammatic examples for this type of bilaminar construction appear in figure 1.2 below.

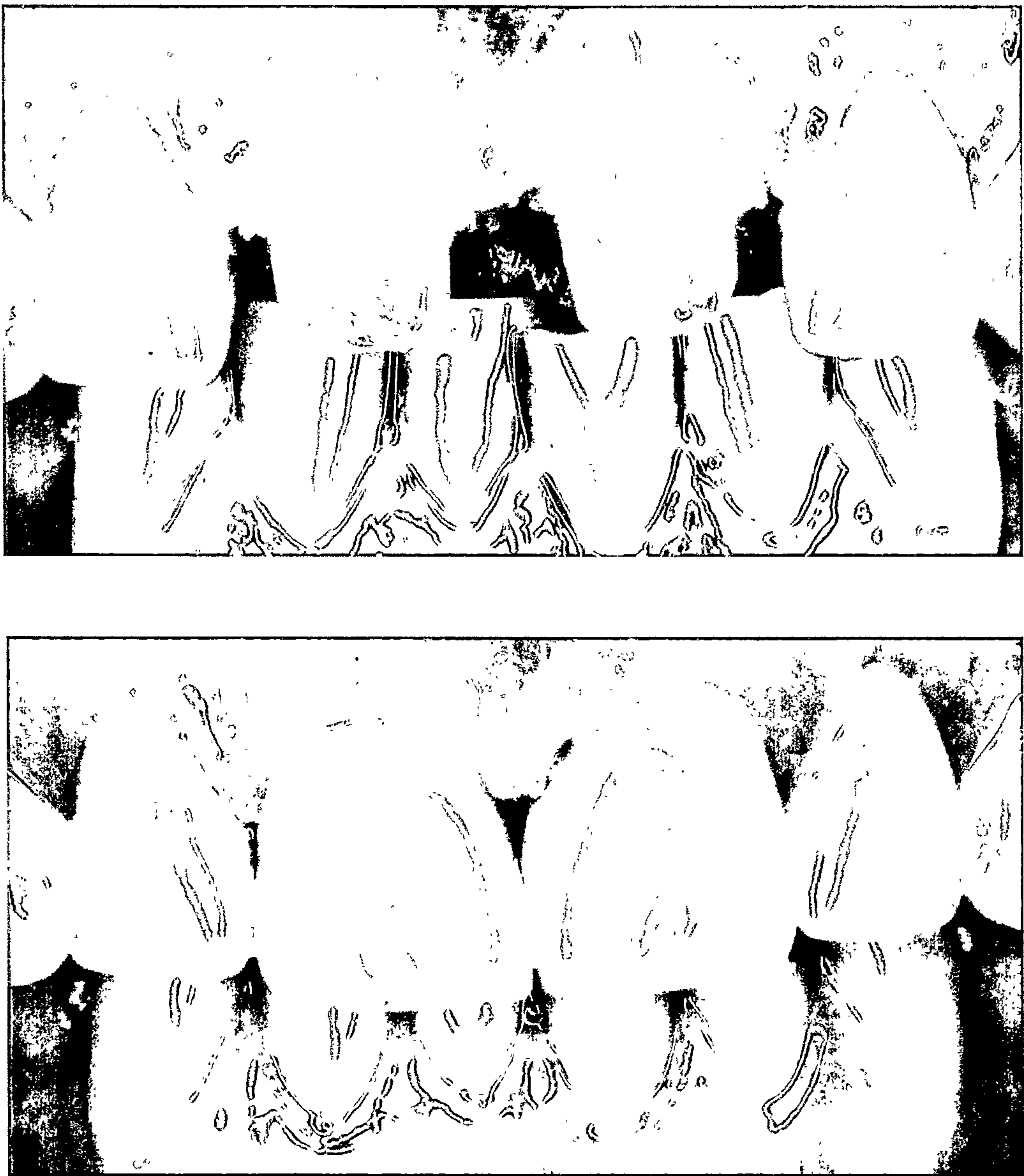
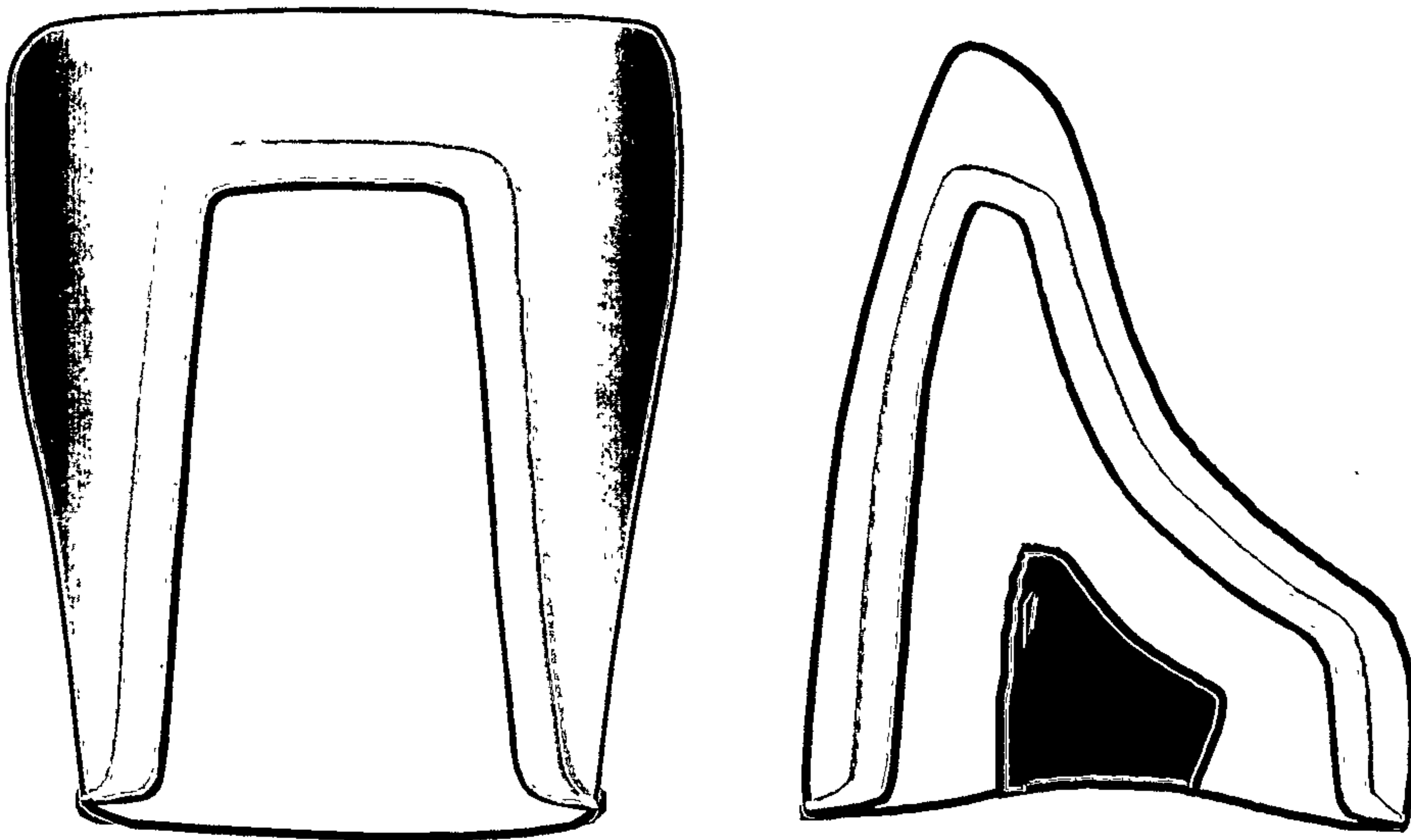
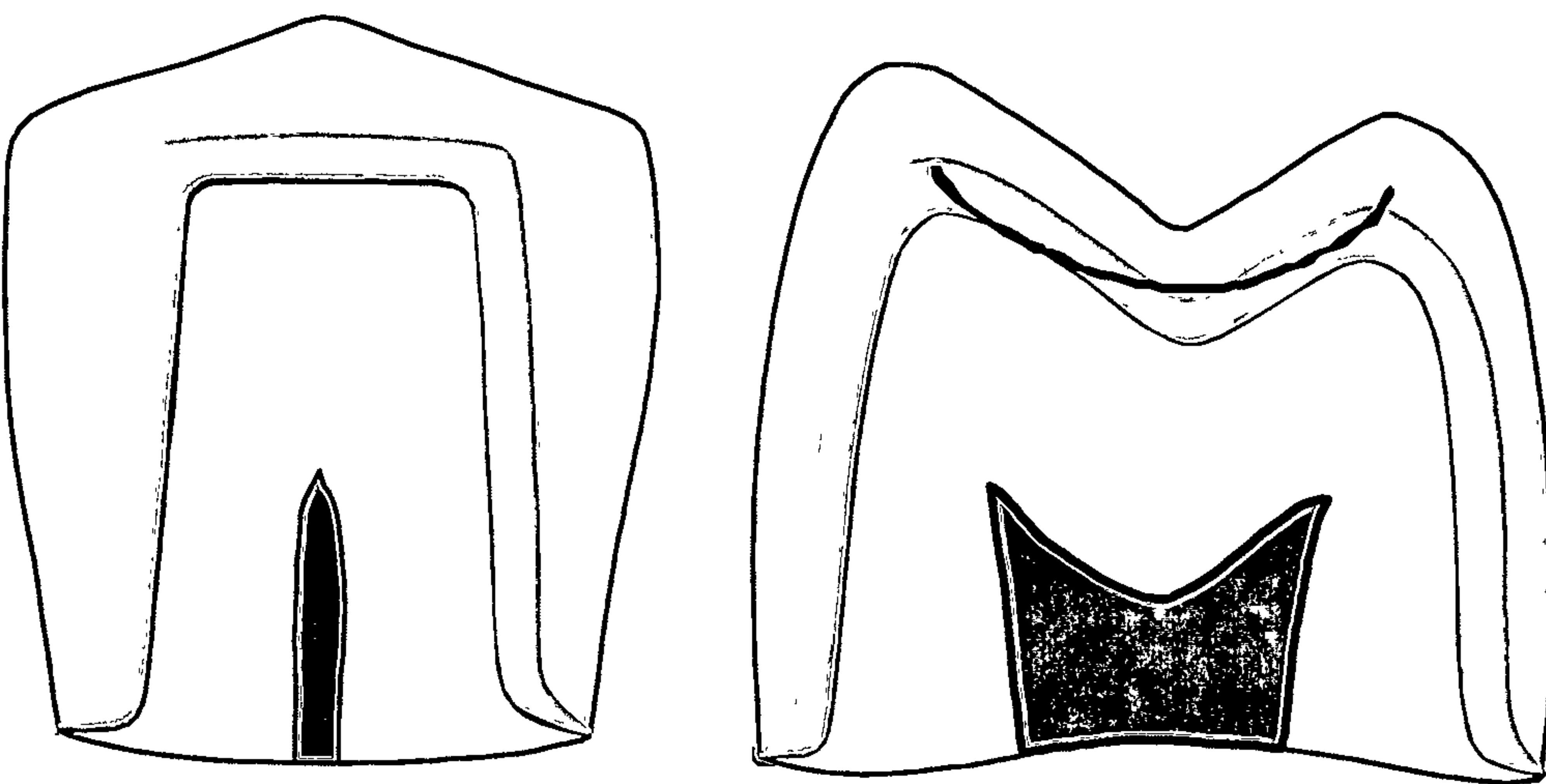


Figure 1.1. Top. Clinical preparation of two upper central incisors prior to the placement of In-Ceram Porcelain Jacket Crowns. Bottom. The In-Ceram crowns in place after cementation with Glass Ionomer Cement.



a. Anterior Crown: Labial view on the left and Proximal view on the right. In these diagrams the Vitadur Alpha veneering porcelain is the outer layer, the In-Ceram core material is the next layer and the third innermost section represents the dentine or supporting core.



b. Premolar Crown – Buccal view and Proximal view. In these diagrams the Vitadur Alpha veneering porcelain is the outer layer, the In-Ceram core material is the next layer and the third layer represents the dentine.

Figure 1.2. Diagrams of In-Ceram® restorations. Diagram (a) represents an anterior crown and diagram (b) a posterior crown.

The critical problem for all ceramic materials, not the least those used in dentistry is the huge difference in theoretical strength, based on the covalent nature of their structure, and the usual strengths found in general use (7000-70000 MPa versus 7-700 MPa) (Drummond et al. 1991). This was originally determined by Griffith, who reported that the theoretical strength for all solids could generally be regarded as a constant with a value approximating  $E/10$  (Lawn, 1993). This work led to the concept of "Griffith flaws", a large number of submicroscopic flaws with an approximate length of 2 micrometres and a wall separation of approximately 50 nanometres. For example, if we consider dental porcelain with a strength of 70 MPa, then using the following formula to approximate crack length:

$$K_{Ic} \sim \sigma \sqrt{\pi c}$$

$$1 \times 10^{-6} = 70 \times 10^{-6} \sqrt{\pi c}$$

$$1/70 = \sqrt{\pi c}$$

$$1/500 = \pi c$$

$$1/1500 = c$$

$$c \sim 70 \mu\text{m}$$

According to the following formula from Lawn (1993) this will give a theoretical stress-concentration factor in the order of 100.

$$\sigma_c / \sigma_A \approx 2c/b = 2(c/p)^{1/2}$$

Where:

---

$\sigma_A$  = Applied stress (N)

$\sigma_c$  = Stress at C

$p$  =  $b^2/c$

$b$  and  $c$  are the semi-axes of an elliptical cavity.

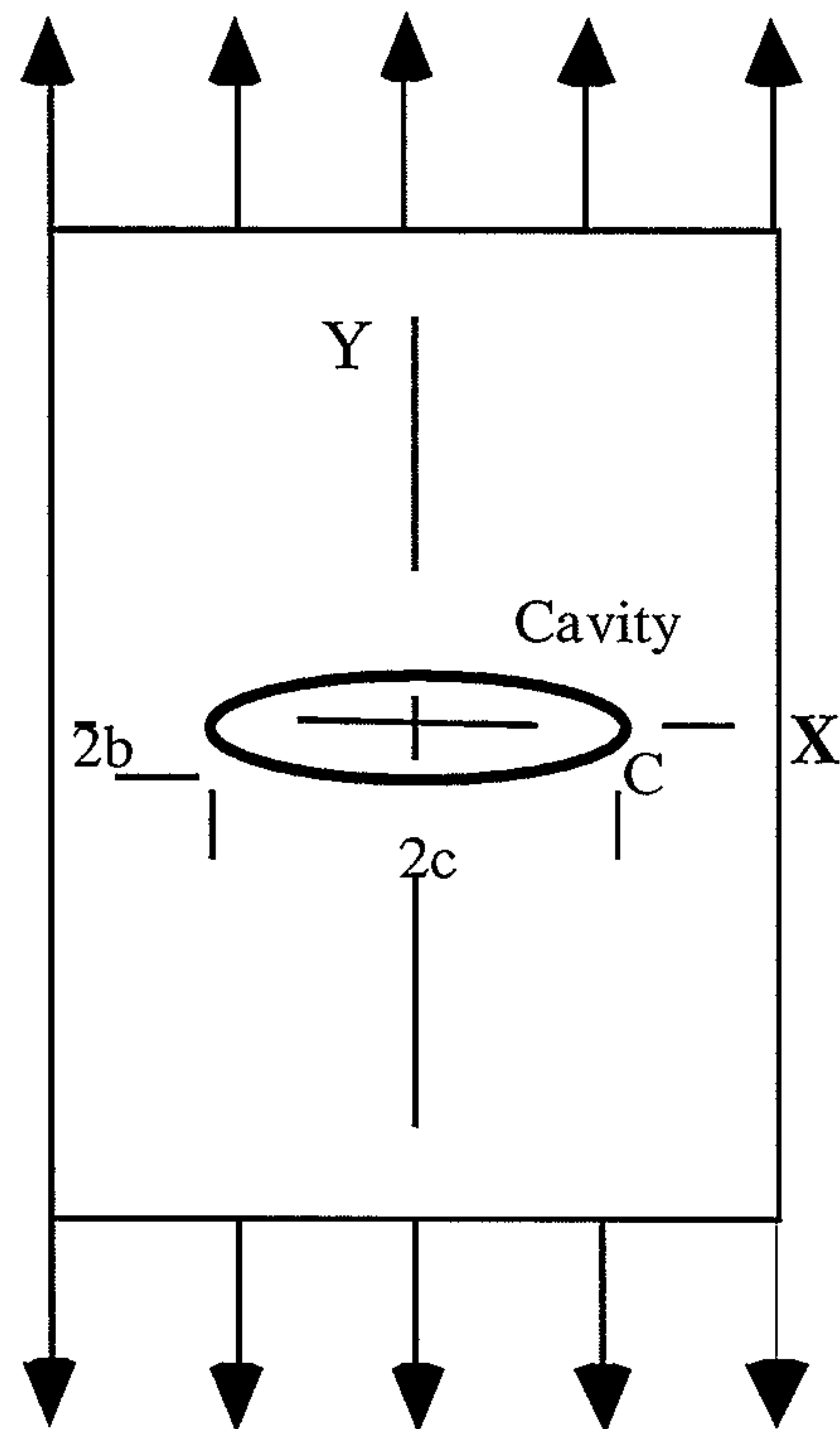


Figure 1.3. Diagram of a plate containing an elliptical cavity subjected to uniform tension.

Until the introduction of In-Ceram®, the other attempts to improve on the strength and toughness of the traditional aluminous porcelain jacket crowns were not successful, (Dickinson et. al., 1989), (Fan 1988)

Some of the undesirable aspects for the use of ceramics in the posterior teeth have been their extremely brittle and abrasive nature (Wiley 1989). In the past manufacturers have quoted hardness values when predicting the potential clinical wear of new ceramics. This was disputed in an article by Seghi et al., (1991) where their results demonstrated that Knoop hardness values for the ceramics tested showed a poor correlation to opposing tooth wear. They suggested that wear rate discrepancies between the materials tested may be due to different ceramic microstructures and or related to fracture toughness. Lawn (1993) suggests that the most important factors in resistance to wear in brittle ceramics are

toughness and to a lesser degree, hardness. The greater the toughness, the greater the wear resistance for ceramics. Hardness also plays a significant role in wear; for example metals have significantly higher toughness than ceramics, but generally much lower hardness and significantly more wear.

### **1.1.2 The increase in All-Ceramic Restorations in Occlusion.**

The literature has innumerable case reports where all-ceramic restorations are used in areas where they will be subject to high occlusal forces. Examples where "etched porcelain onlays" are used in the treatment of bruxism are increasingly common, (Glaser et al. 1991). It is disconcerting that such articles do not put values on the forces these systems could be subject to or perhaps it is simply a case that we still do not fully understand the complexities involved when using such treatments.

An article by Banks (1990) provides a comprehensive review of the new role of ceramics in posterior occlusions. This article encompasses the views relating to bonding ceramic inlays, onlays and conservative crowns to tooth structure. The article also looks at indications, contraindications, advantages and disadvantages. The conclusion which can be made from the article is that, like other materials, ceramic restorations are not about to provide all the answers to restorative needs, but where the rules for their application are adhered to they can be a successful means of treatment. A review article by Broughton (1990) indicated that clinicians were at odds where problems for these systems occurred. Some felt that the problems were operator related, others felt that they were related to the dental laboratory procedures and a few felt that they could come from both.

The paper by Morena et al., (1986) indicated several different dental ceramic systems, (a feldspathic porcelain, an aluminous porcelain and a fine-grained polycrystalline core material) that were likely to experience failure in the mouth. These failures were the result of the effects of corrosion and fatigue at loading levels that could reasonably be expected within the oral environment. An important point from their paper is the recognition of the importance of a critical flaw size and the need to attenuate crack tips by blunting or deflection. These flaws could conceivably be introduced during either the laboratory phase, or during clinical placement procedures and hence the degree of confusion reported by Broughton (1990). Bridgework using all-ceramic techniques has been indicated however there has been early recognition of the limitations, (Campbell et. al., 1988).

Further review articles have appeared in the literature since, as newer dental ceramic systems have become available. One such article by Denry (1996) provides an insight to the range of crystalline structures and flexural strengths now available. A summary in table 1.1 from her article appears below. The range of principal crystalline phases

reviewed, included leucite, alumina, fosterite, zirconia, mica hydroxyapatite, lithium disilicate, sanidine and spinel.

**Table 1.1 Comparative flexural strength values for all ceramic materials**

| Material                        | Product           | Modulus of Rupture (MPa) | Reference               |
|---------------------------------|-------------------|--------------------------|-------------------------|
| Aluminous Core (Pt foil)        | Vitadur-N core    | 123                      | Seghi et al., 1990b     |
| Aluminous Core                  | Hi-Ceram          | 139                      | Seghi et al., 1990b     |
| Zirconia-reinforced             | Mirage II         | 70                       | Seghi et al., 1990b     |
| Injection Moulded Aluminous     | Alceram           | 162                      | Wohlwend et al., 1989   |
| Heat-pressed Leucite-reinforced | IPS Empress       | 97                       | Seghi & Sorensen, 1995  |
|                                 |                   | 160 – 180                | Dong et al., 1992       |
| Leucite-reinforced              | Optec HSP         | 104                      | Seghi et al., 1990b     |
| Slip-cast alumina               | In-Ceram          | 446                      | Seghi & Sorensen, 1995  |
| Slip-cast alumina               | In-Ceram Spinell  | 378                      | Seghi & Sorensen, 1995  |
| Slip-cast alumina               | In-Ceram Zirconia | 604                      | Seghi & Sorensen, 1995  |
| Mica glass-ceramic              | Dicor             | 125                      | Seghi et al., 1990b     |
| Hydroxyapatite glass-ceramic    | Ceraperl          | 150                      | Hobo & Iwata, 1985      |
| Lithia based glass-ceramic      | Experimental      | 188                      | Anusavice et al., 1994b |
| Machinable glass-ceramic        | Dicor MGC         | 229                      | Seghi & Sorensen, 1995  |
| Machinable ceramic              | Vita Mark II      | 122                      | Seghi & Sorensen, 1995  |
| Magnesia Core (glazed)          | Experimental      | 269                      | Wagner et al., 1992     |

## 1.2 The Aim of this Study.

### 1.2.1 Introduction

The motivation for this thesis came from the introduction of the In-Ceram glass infiltrated crown and bridge system, which brought with it a small number of early clinical failures. In some cases these failures actually occurred when the patient first closed down on the tooth at the cementation stage or within a few weeks post cementation. Given the recorded

high flexural strength for this dental ceramic core material – the highest recorded at the time of its introduction, the idea was to investigate the possible cause for the early failures and look for ways to reduce these fractures.

In general terms the investigation began by looking at the samples under optical and scanning electron microscopy, noting the fracture site and repeating the fracture pattern with a group of similar crowns in vitro. A further development comes in the form of a simpler model based on a ceramic disc, measuring load to failure under bending stresses. This will be followed by developing the disc design to resist fracture initiation and then applying that principle to a new crown design.

The failed restorations were sent to me for investigation by the Dental Technicians who made them. It should be remembered from the outset that one of the main reasons these restorations are used is for their aesthetic qualities. A metal ceramic crown may still be functional if the outer veneering porcelain delaminates in whole or in part, but these restorations are considered to have failed if part of the outer veneering porcelain comes away. An example of this type of failure can be seen in figure 1.4. One longer-term failure came from my own clinical patients after it had been in service for a period of four and a half years. This crown appears in figure 1.6.

### **1.2.2 Investigation**

The crowns were investigated by optical and scanning electron microscopy. The results indicated that the fractures could have been initiated from contact induced fracture similar to that of a spherical indentation such as that from an opposing cusp. Examination of the occlusion with the crown on its master model and articulated with the opposing model indicated the possibility that an opposing cusp caused the fracture. Examples of clinical failures are presented below in figures 1.4 to 1.7. The diagram demonstrating this complex fracture pattern appears in figure 1.8.

The individuality of crown shapes both externally and internally further complicates both the fracture pattern and the resistance to crack initiation. Kelly (1999) demonstrates results for a monolayered ceramic which indicate cracks starting from the underlying surface. This is also the case with Malament et al., (1999). The In-Ceram system on the other hand is a bi-layered alumina based system.

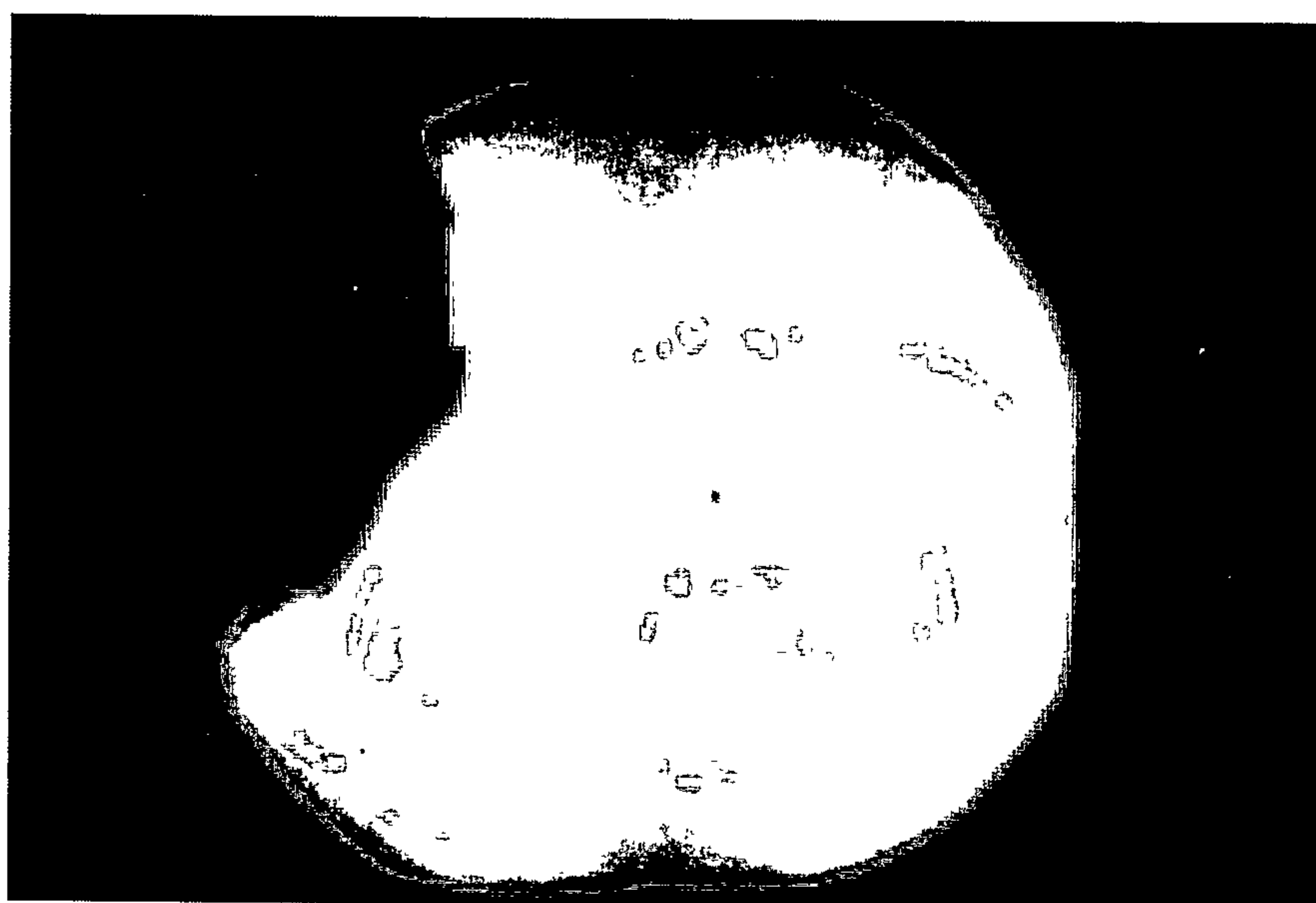


Figure 1.4 a

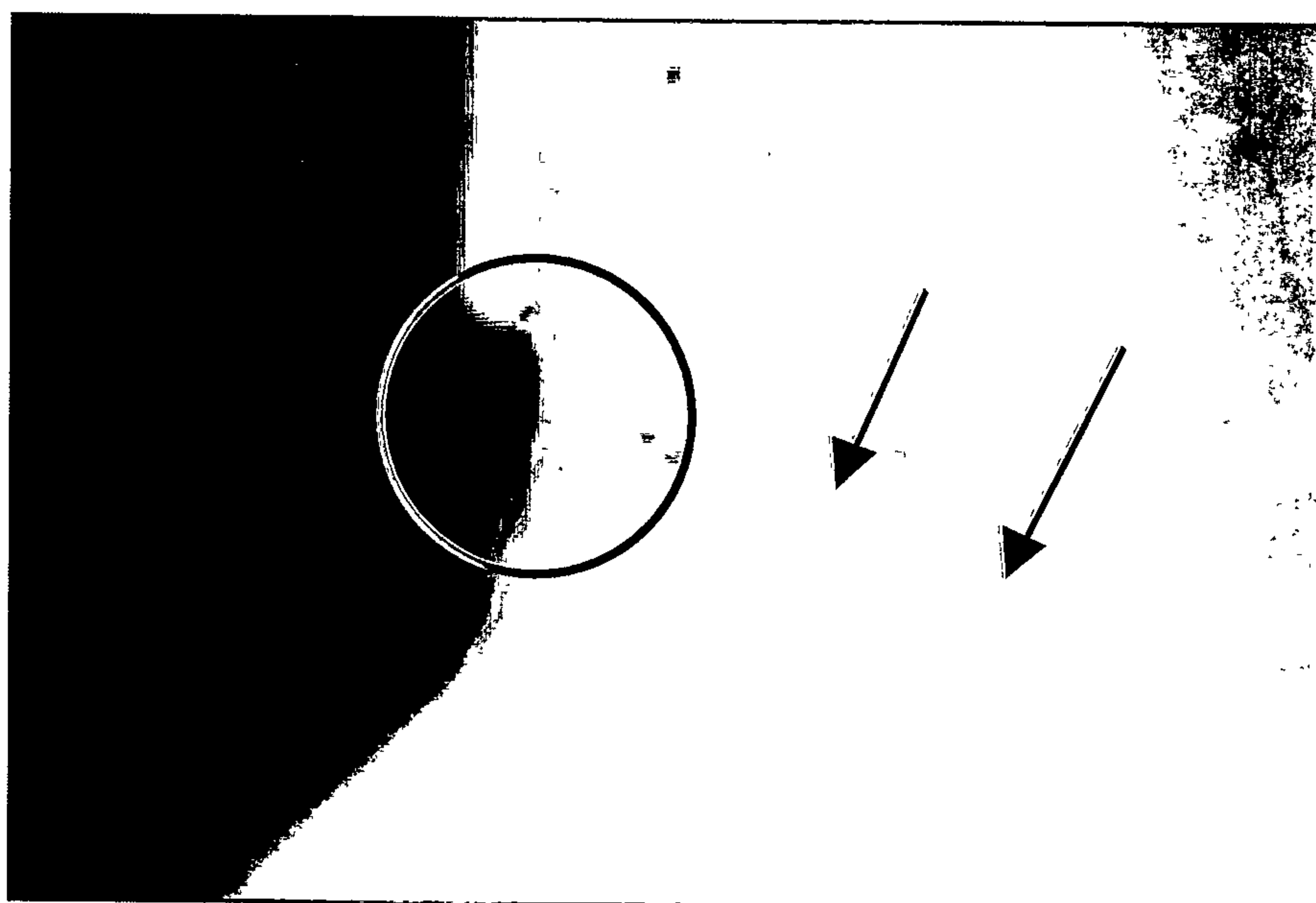


Figure 1.4 b

Figure 1.4. Clinical failure of an In-Ceram® Crown from a lower right first molar. Figure 1.4 a shows the whole occlusal view with an area at the right hand side of the photograph where the veneering porcelain has delaminated but the core remained intact. The red circle in figure 1.4 b highlights the circular indentation from the initial crack and the arrows indicate a line from a radial crack. The veneering porcelain below the crack has also delaminated.

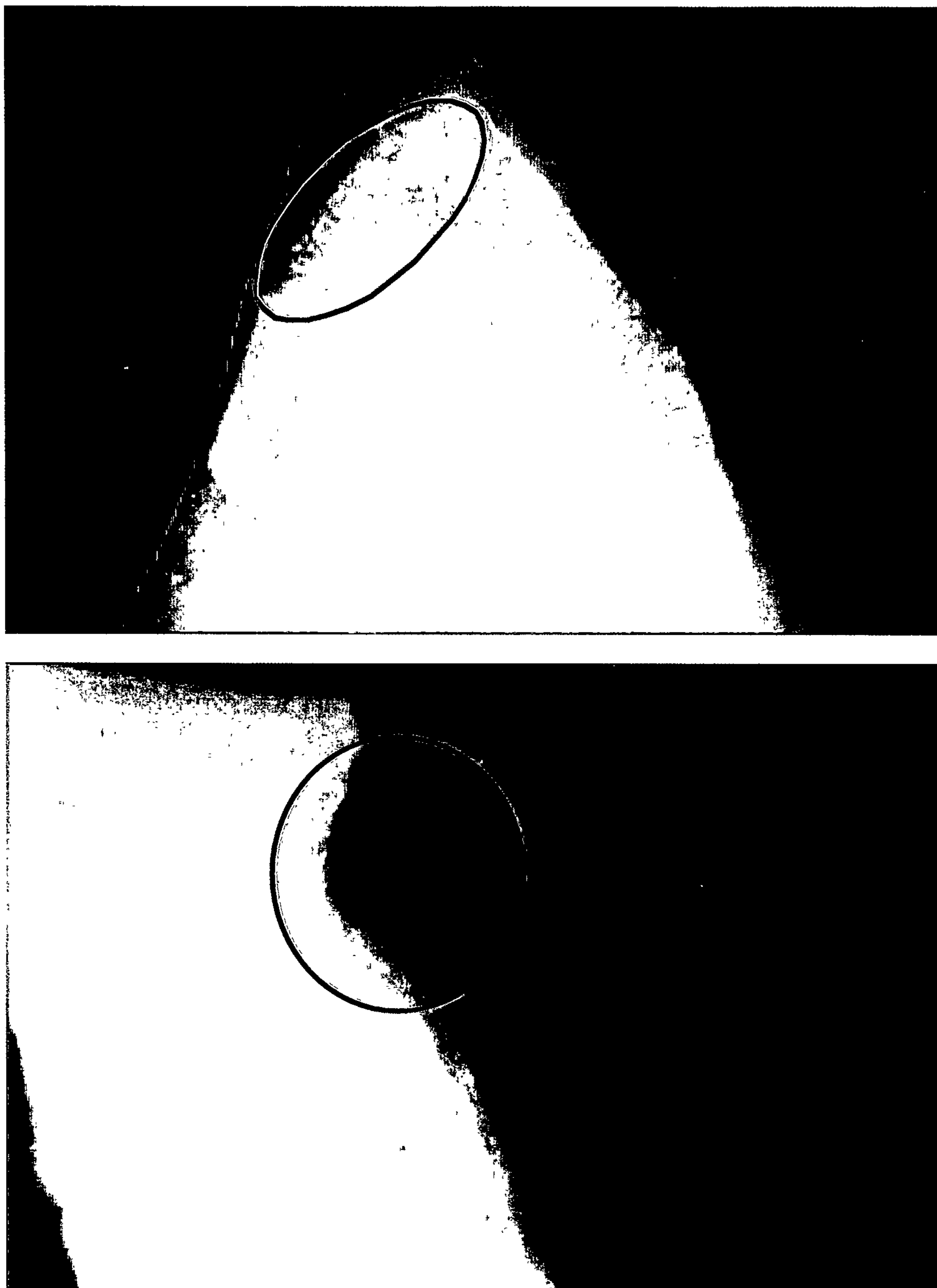
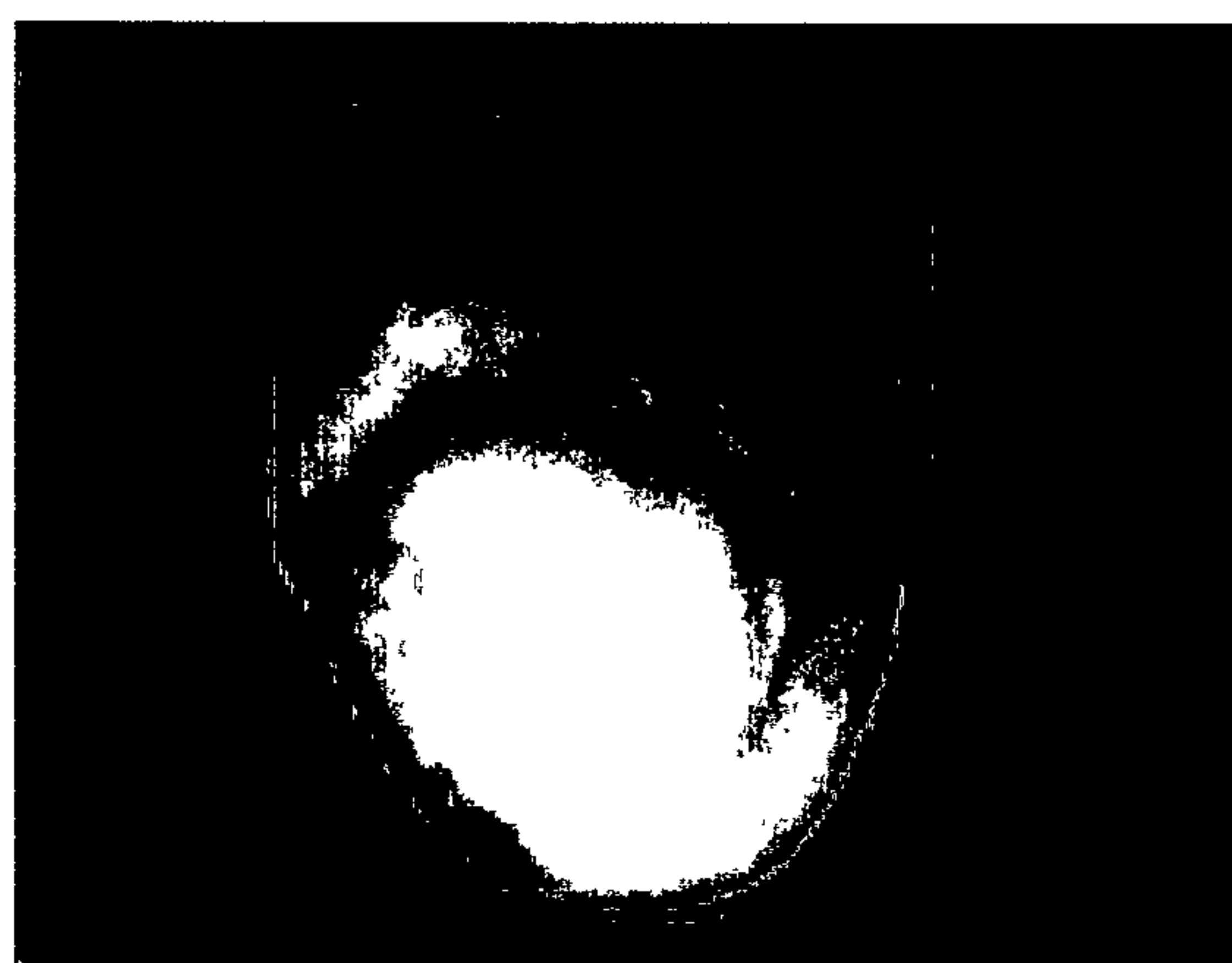


Figure 1.5. Clinical failure of an In-Ceram® Crown. The top photograph demonstrates the mesial view of the incisal edge/corner of this In-Ceram crown, which fractured from occlusal trauma due to bruxism. The bottom photograph demonstrates the inciso-palatal view and shows the left half of the original circular contact point. This crown was sectioned with a high-speed diamond bur to facilitate its removal. The red outlines indicate the original contact/fracture area.



a. Proximal view



b. Palatal view



c. Inside palatal view

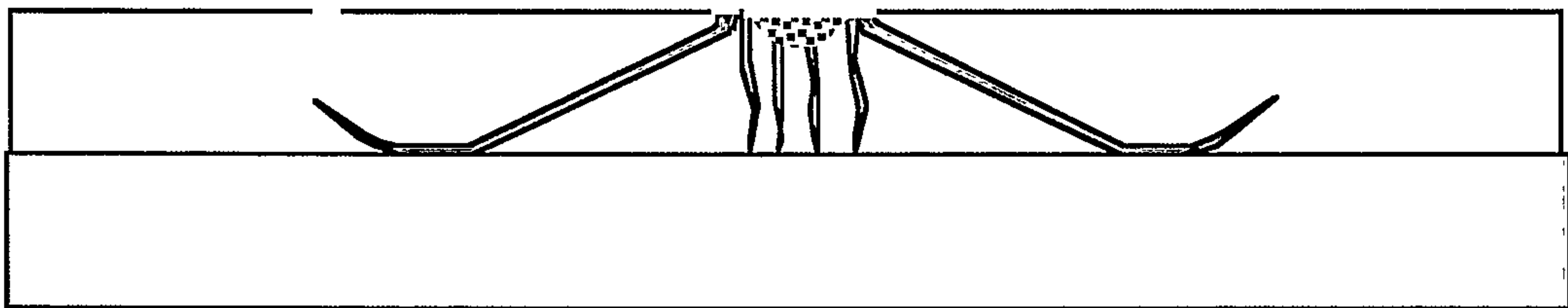


d. Close-up of fracture site

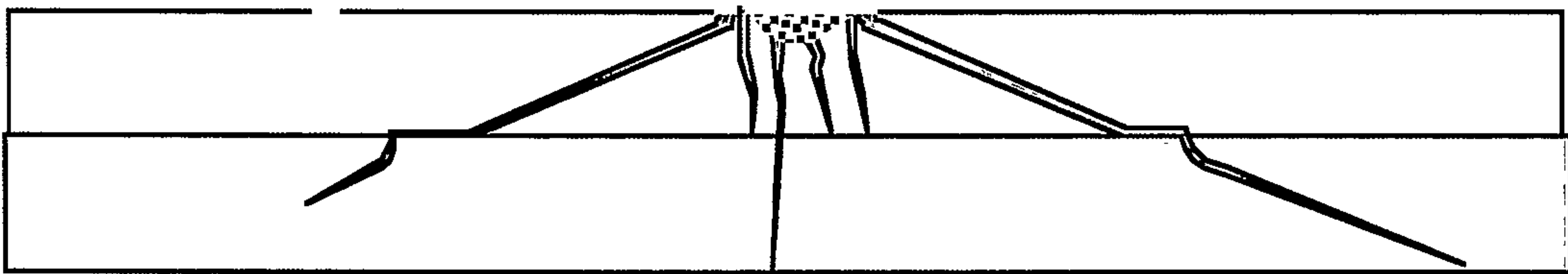
Figure 1.6. Clinical failure of an In-Ceram® Crown. Views (a) and (b) demonstrate the short distance from the fracture site to the gingival margin of the crown. View (c) demonstrates the anatomy of the veneer on the palate for this particular crown. View (d) demonstrates the initial fracture site (arrowed) through the veneering porcelain to the right of the black marker.



Figure 1.7. Clinical failure of an In-Ceram® Crown. This posterior crown had been in service for four and a half years. The initial fracture site (arrowed) is over the mesio-buccal cusp. After the initial surface fracture the radial cracks developed and ultimately the buccal wall of the crown fractured away. Some of the glass ionomer cement remains on the tooth.



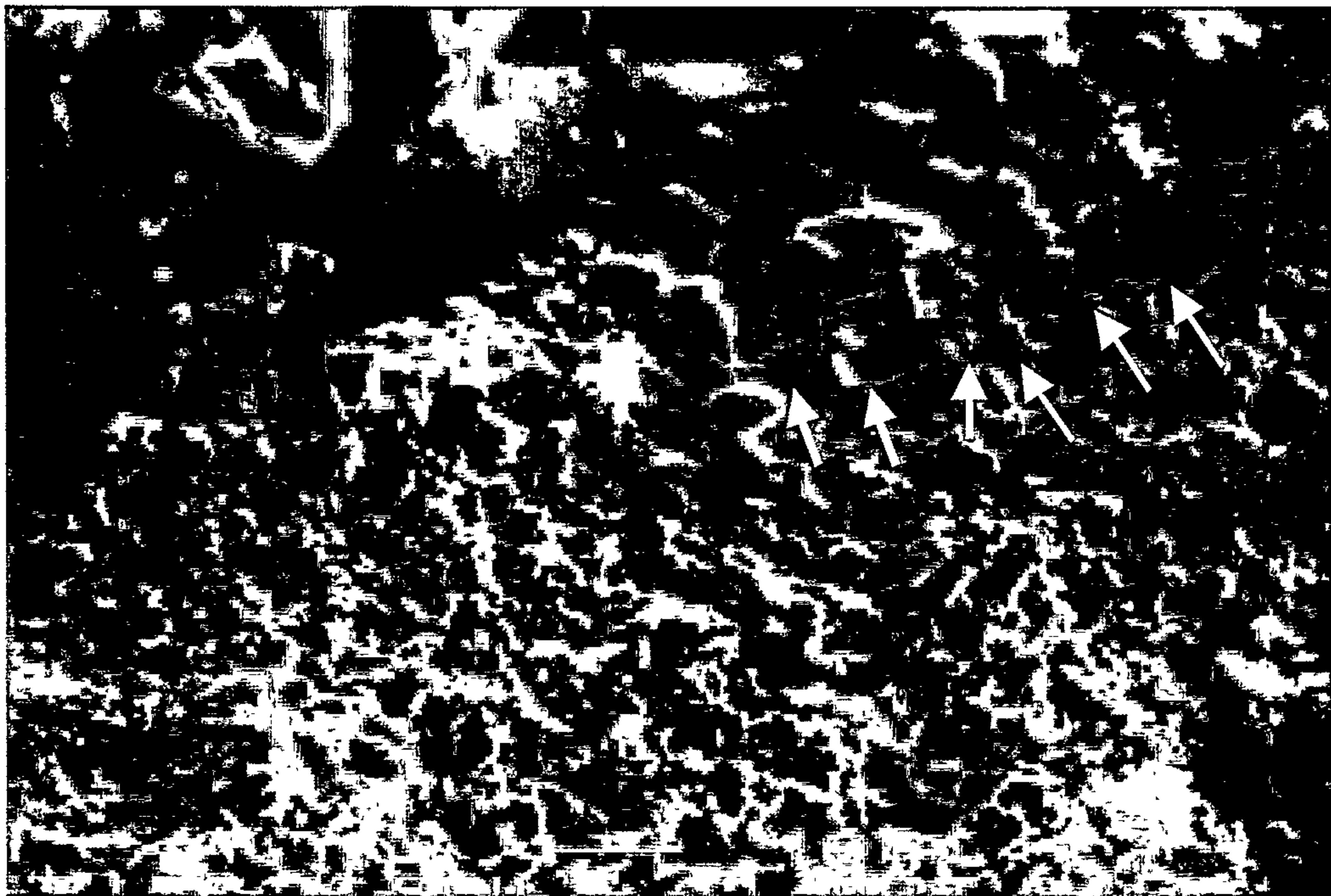
a.



b.

Figure 1.8. Diagram of the complex fracture patterns seen in the clinical examples. In part (a) the representation is for the pattern seen when the veneering porcelain fractures away from the core leaving the core intact which is seen in figure 1.5 and part (b) is for the situation where the core also fractures such as in the example in figure 1.7.

The advantage gained from the clinical crowns was that in each case the point of occlusion could be identified and in each case it appeared to coincide with the crack initiation site. This was particularly evident with the first example in figure 1.4. The fact that the crack initiations appeared to be inconsistent with current fracture theory reported for porcelain jacket crowns might also reflect the vastly superior strength of the In-Ceram core material compared with those core materials previously available. The example in figure 1.4 demonstrated areas of delamination between the layers but it also demonstrated a crack plane beneath the contact point where a radial crack passed through both layers without deviation. The scanning electron micrograph below in figure 1.9. shows both the crack plane of a radial crack and evidence of a tortuous delamination marked by the white arrows.



— 10 μm

Figure 1.9 Scanning electron micrograph of the fracture plane perpendicular to the crown surface and below the contact area. The top left diagonal third is the glassy Vitadur Alpha and the bottom right diagonal two thirds is the In-Ceram core material. Original magnification 1000x

### 1.2.3 Hypotheses.

The results of investigation by optical microscopy indicated a circular crush mark at the surface and cracks not unlike a Hertzian cone crack proceeding through the feldspathic covering layer, i.e. the veneering porcelain.

This initial investigation led to the following hypotheses:

1. That loading by spherical indentation such as might occur in occlusion from an opposing tooth might lead to failure of the crowns at an early stage of their service.
2. Also that the bilaminar layering of the In-Ceram crown can be made more resistant to crack initiation where a load is applied using a spherical indenter, by increasing the core to veneer porcelain ratio, that is, making the core thicker.

The structures of several dental porcelains will be reviewed in chapter 2 to demonstrate the wide range of approaches to both improving the toughness and also aesthetics. This chapter will also review basic mechanical concepts for stresses and crack initiation.

The concept of spherical indentation as a method for crack initiation is introduced and explored in chapter 3 using a series of anterior crowns to repeat the failure and provide information concerning expected failure load values and their consistency. Weibull statistics will be employed to evaluate these crowns for their reliability. The information will then be applied to a simple bilaminar disc model in chapter 4 in an effort to repeat the result and then provide a platform to investigate ways to improve resistance to crack initiation.

The concept that increasing the core to veneer porcelain thickness ratio will increase the resistance to crack formation and catastrophic crack propagation is based on the knowledge that the fracture toughness of the body porcelain is low compared to the core ceramic and that the modulus of elasticity of the core ceramic, is much higher than the veneering porcelain. Also the size of the crack, relative to the microstructural features has an effect on the overall crack propagation through the bilaminar system of this type of crown. If the strain within the body porcelain can be contained under the stress of the occlusal load, by the core porcelain, then the risk of initiating a crack can be reduced. These aspects will be examined in chapters 5 and 6 where a range of core to veneering ratios will be tested both mechanically and via finite element analysis. The final chapter is a summary for the new design concept.

#### **1.2.4 The ceramic system employed for this study.**

This study will focus primarily on the glass-infiltrated alumina ceramic core system, In-Ceram® and its veneering porcelain Vitadur Alpha®. However the concepts developed should be relevant to all crowns that employ a bilaminar structure where the core material is much stronger and stiffer than the overlying glassy veneer.

The alumina version of the In-Ceram material has been indicated by the manufacturer, Vita Zahnfabrik, for use as a restorative material for anterior and posterior single unit crowns, inlays and onlays, and three unit anterior fixed partial dentures. There are two other versions. The first uses Magnesia-Aluminate Spinel as the crystalline part and this material is primarily aimed for anterior crowns and possibly inlays and onlays. It is more translucent. The most recent version uses a blend of alumina and zirconia for the crystalline part. This material is aimed at posterior three unit fixed partial dentures or bridges. All three variants are designed as core ceramics to support a more glassy porcelain that provides the aesthetic and functional final form of the prosthesis.

To construct the core material, the process requires the crystalline phase to be applied first and sintered. This produces a weak and porous framework. A Lanthana based glass material is then infiltrated into the core framework under high temperature and the final product is a very strong and relatively tough ceramic core to support the veneering porcelain during the final construction and firing processes.

#### **1.2.5 The following study**

The following chapters will begin with a review of dental porcelain and an insight into the directions for increasing their strength and toughness. The next part will review fracture processes. Chapter three will look at in vitro failure of anterior porcelain jacket crowns using the In-Ceram® system which will lead into the development of a model to replicate these failures. The next step is to develop a method to use In-Ceram® in a way that is more resistant to crack initiation and then apply that information to the crown model used in chapter three.

# CHAPTER 2

## Dental porcelain, its failures and mechanical testing

### 2.1 Introduction

This chapter introduces the basic chemical and structural aspects of a number of different dental porcelains that have been developed for dentistry over more recent times. The scanning electron micrographs are used to illustrate examples of the wide variety of crystalline complexes used to improve strength and toughness. The latter half of the chapter revises some of the mechanical concepts that will be useful in the thesis investigation. The principal areas of interest are the initiation of cracks from blunt indenters, the role of the interface between the two ceramic layers that are found in In-Ceram® and the stresses involved with the formation of radial cracks from the internal surfaces of the crown. According to Rhee et al. (2001) the two important modes for failure are the Hertzian-like cone crack in the near contact region and the radial cracks that can initiate at relatively low loads. The third area, delamination is not considered as a principal mode of contact damage, but as a secondary to either Hertzian or radial cracks, clinically it can be equally destructive.

#### 2.1.1 The Composition of Dental Porcelain

Several summaries for the composition of dental porcelain have been written. They cover the composition of feldspathic porcelain as veneering porcelain in all-ceramic and metal-ceramic crowns, examples being Southan (1970), McLean (1979) and Binns (1983). They describe a history of modifying the basic Potash Feldspar-Quartz-Kaolinite mix by the removal of mullite and free quartz, while increasing sodium oxide and alkaline earth oxides as bivalent glass modifiers, to improve translucent properties while trying to maintain strength. Fluxing agents have also been added to lower the melting temperatures and make them easier to handle in the dental laboratory. These materials are thus substantially glassy and Binns (1983) describes their classification as porcelain as "somewhat of a misnomer".

The  $K_2O$  content was also varied to accommodate the need to match the coefficient of thermal expansion for metal alloys used in dental metal-ceramic techniques. The increase in  $K_2O$  content allowed the formation of a greater proportion of leucite crystals,

(coefficient of thermal exp.  $27 \times 10^{-6}/^{\circ}\text{C}$ ). This led to an overall rise in the coefficient of thermal expansion to something in the order of  $13.5 - 15.5 \times 10^{-6}/^{\circ}\text{C}$ . These materials are not compatible with the alumina cores used for porcelain jacket crowns. The residual stresses from the firing process are so great that the veneering porcelain cracks and breaks off during the cooling down process.

The phase diagram below in figure 2.1 demonstrates the complexity of the ceramic formulations found in natural clays and helps us to understand where the dental ceramics come from.

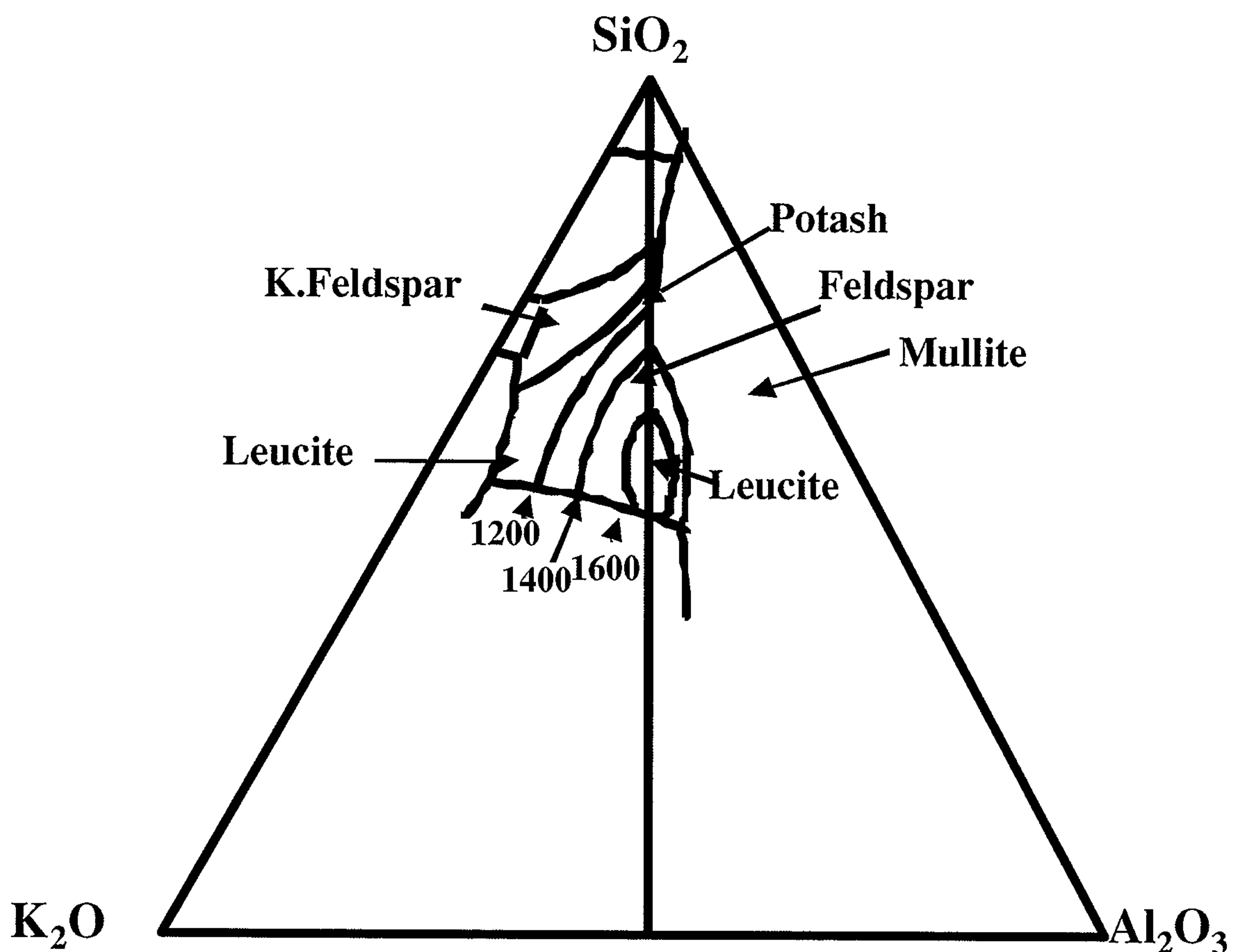


Figure 2.1. This is a  $\text{K}_2\text{O}-\text{Al}_2\text{O}_3-\text{SiO}_2$  phase diagram from Binns (1983). The numbers represent the temperature gradients at which melting occurs.

Table 2.1 from Binns (1983) gives an indication of the variability of high purity potash feldspars that could be used for dental porcelain manufacture.

It is also noted that both feldspar and leucite undergo phase transformations during the firing processes for Dental porcelains, (Barreiro et al., 1989)

**Table 2.1      Chemical Compositions (%) of High-Purity Naturally occurring Feldspars**

| Source                         | Idaho | Sweden | Morocco | Morocco | Portugal |
|--------------------------------|-------|--------|---------|---------|----------|
| SiO <sub>2</sub>               | 65.1  | 69.2   | 64.5    | 64.4    | 64.3     |
| TiO <sub>2</sub>               | <0.01 | <0.01  | <0.01   | <0.01   | <0.01    |
| Al <sub>2</sub> O <sub>3</sub> | 18.8  | 16.7   | 18.9    | 18.7    | 19.2     |
| Fe <sub>2</sub> O <sub>3</sub> | <0.01 | 0.06   | <0.01   | <0.01   | 0.03     |
| CaO                            | 0.05  | 0.10   | 0.05    | 0.02    | 0.05     |
| MgO                            | <0.02 | <0.02  | <0.02   | <0.02   | <0.02    |
| Na <sub>2</sub> O              | 3.46  | 2.73   | 1.91    | 1.45    | 2.99     |
| K <sub>2</sub> O               | 11.8  | 10.9   | 13.7    | 14.4    | 12.5     |
| P <sub>2</sub> O <sub>5</sub>  | 0.19  |        | 0.38    | 0.37    | 0.37     |
| Loss                           | 0.17  | 0.21   | 0.37    | 0.28    |          |

From the same author we can also obtain an indication of the typical chemical composition of the more traditional Porcelain Jacket Crown porcelains.

**Table 2.2      Typical Chemical Composition (%) of Jacket-Crown Porcelains**

|                                | Firing Temperature (°C) |             |           |
|--------------------------------|-------------------------|-------------|-----------|
|                                | 1250 / 1300             | 1060 / 1100 | 900 / 980 |
| SiO <sub>2</sub>               | 70.6                    | 63.7        | 67.3      |
| Al <sub>2</sub> O <sub>3</sub> | 17.2                    | 19.5        | 10.8      |
| CaO                            |                         |             | 2.2       |
| Na <sub>2</sub> O              | 2.8                     | 2.2         | 4.6       |
| K <sub>2</sub> O               | 9.4                     | 8.2         | 7.9       |
| B <sub>2</sub> O <sub>3</sub>  |                         | 6.0         | 6.8       |

The strong fluxing influence from the addition of  $B_2O_3$  can be seen in the second and third columns of table 2.2 where the melting temperature has been significantly lowered.

### **2.1.2 The basic design for a Porcelain Jacket Crown**

The diagram shown in figure 2.2 demonstrates the basic outline of the traditional porcelain jacket crown and its relationship / connection to a tooth. The crowns themselves originally consisted of an aluminous porcelain core and feldspathic porcelain as a veneering material to cover the core, giving the crown form and aesthetics. At the shoulder the crown had a minimum thickness of about 1 millimetre, with no part of the crown thickness being less than 1 millimetre. Finally the crown was luted to the tooth using Zinc Phosphate cement. Today there are a number of design variants based on whether the system in use has a core or is made entirely of the one material i.e. monolithic. A sample of the types of materials available will be discussed to give an indication for the range of crystalline formulations. Similarly the luting agents themselves range from the traditional Zinc Phosphate cements to the water based Glass ionomer cements and finally the wide range of dental composite resins.

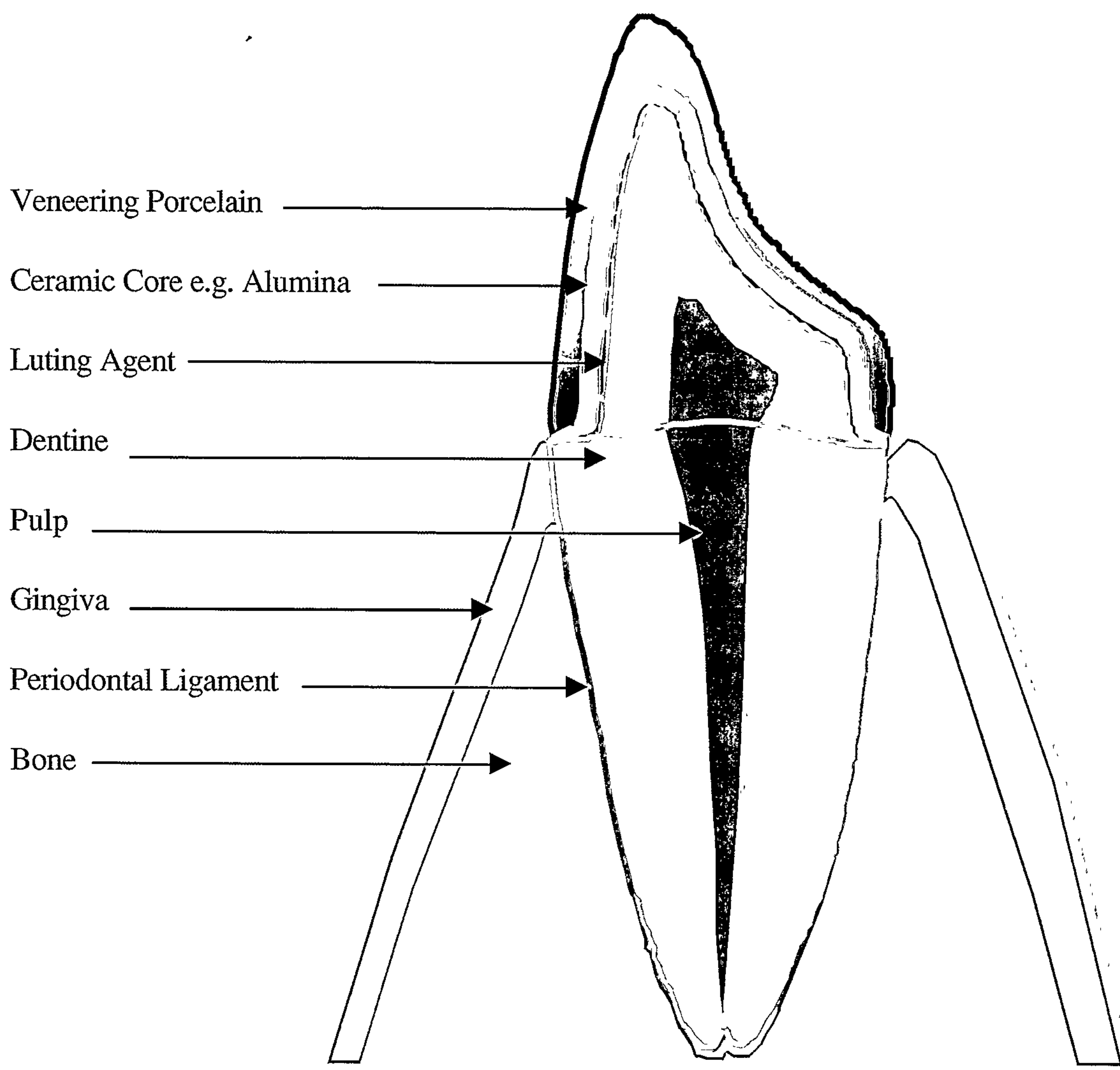


Figure 2.2. This is a simple schematic representation of a bilaminar crown such as a porcelain jacket crown, luted to a vital tooth, which is in turn connected to supporting alveolar bone, by a periodontal ligament.

### 2.1.3 The failure mechanisms

The latter part of this chapter will review the possible mechanisms involved with failures in porcelain. The relationship between the mechanical properties, principally Young's Modulus, and the ceramic structure will be presented in an attempt to explain the initial failure mechanism for these crowns.

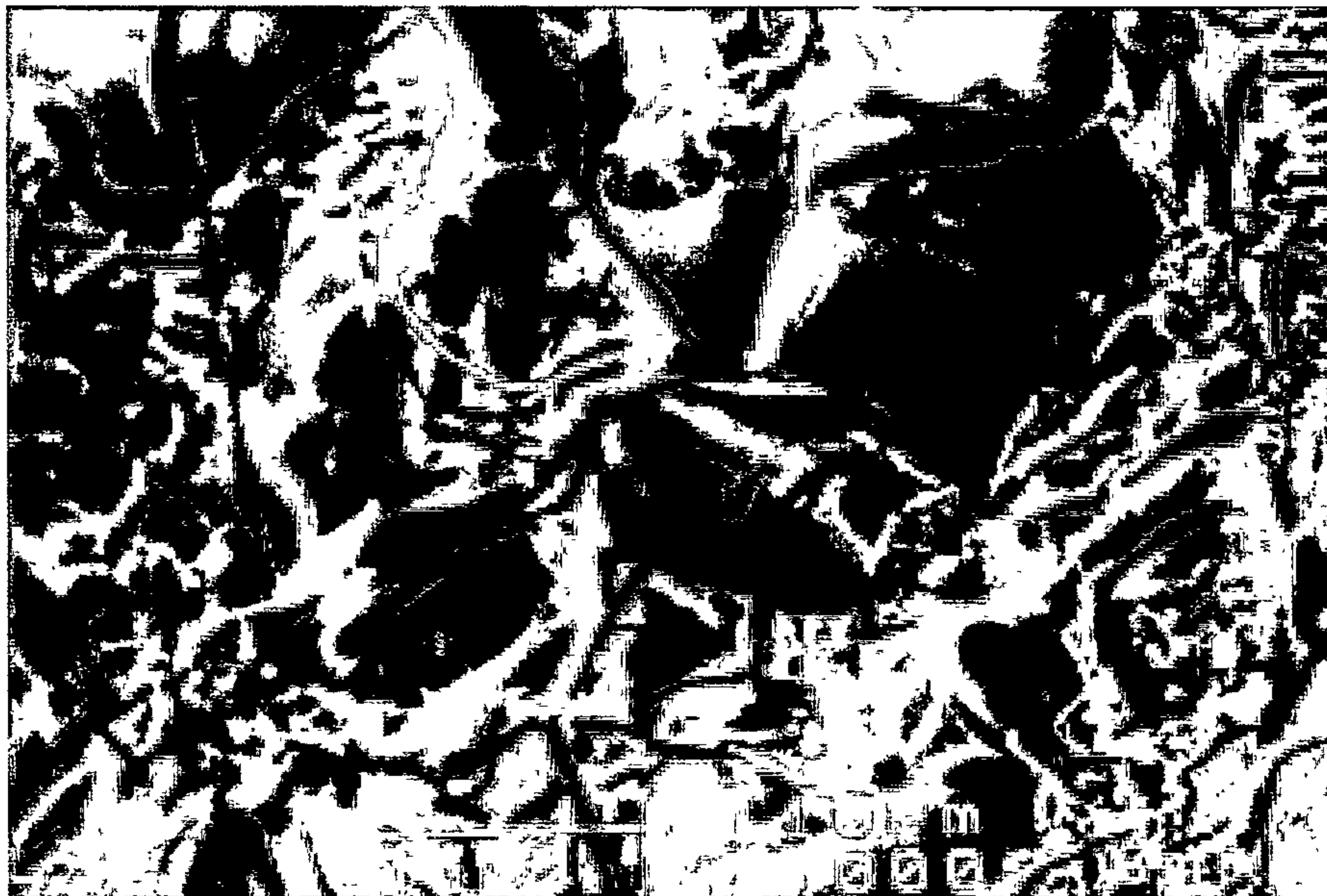
The contacts in the dental occlusion have been observed to resemble spherical indenters (Dietschi et al. 1990) and so the In-Ceram system will be investigated using spherical indenters. Teeth start out with naturally rounded cusps and with age and other variables

such as bruxism, erosion, abrasion and foreign materials such as restorations; they change shape and sometimes become quite flat. For this reason, contact mechanics and the cracking that develops will also be reviewed. This will include the early work of Griffith, Inglis, Auerbach and Hertz.

## 2.2 Examples of Dental Porcelains

### 2.2.1 The Feldspathic Porcelains

The feldspathic porcelains used in all-ceramic systems have coefficients of thermal expansion ranging from  $5.5 - 7.5 \times 10^{-6}/^{\circ}\text{C}$  when used over castable glass and alumina based core materials (Binns 1983), to  $16 \times 10^{-6}/^{\circ}\text{C}$  when used in the newer pressed leucite systems. Figure 2.3 demonstrates an example of the structure of this type of material. Figure 2.4 is an earlier version of this type of material.



————— 10  $\mu\text{m}$

Figure 2.3. Scanning electron micrograph of Vitadur Alpha®. Original Magnification 2000 x



Figure 2.4 Scanning electron micrograph of a feldspathic veneering porcelain, Vitadur N®. Note the lack of distinct crystalline structure and the glassy appearance of this porcelain. Original Magnification 1000x.

### 2.2.2 The Leucite Systems

Leucite has been widely used as a constituent of dental ceramics to modify the coefficient of thermal expansion; however the different thermal expansion coefficients of these crystals from the matrix lead to significant residual stress developing. The leucite while cooling contracts away from the matrix and undergoes extensive twinning and micro-cracking if the size of the grains exceeds a critical value. This is most important where the ceramic is to be fused to metal. The material demonstrated in figure 2.5 is a leucite-modified ceramic that is designed to be fused to a metal substructure.

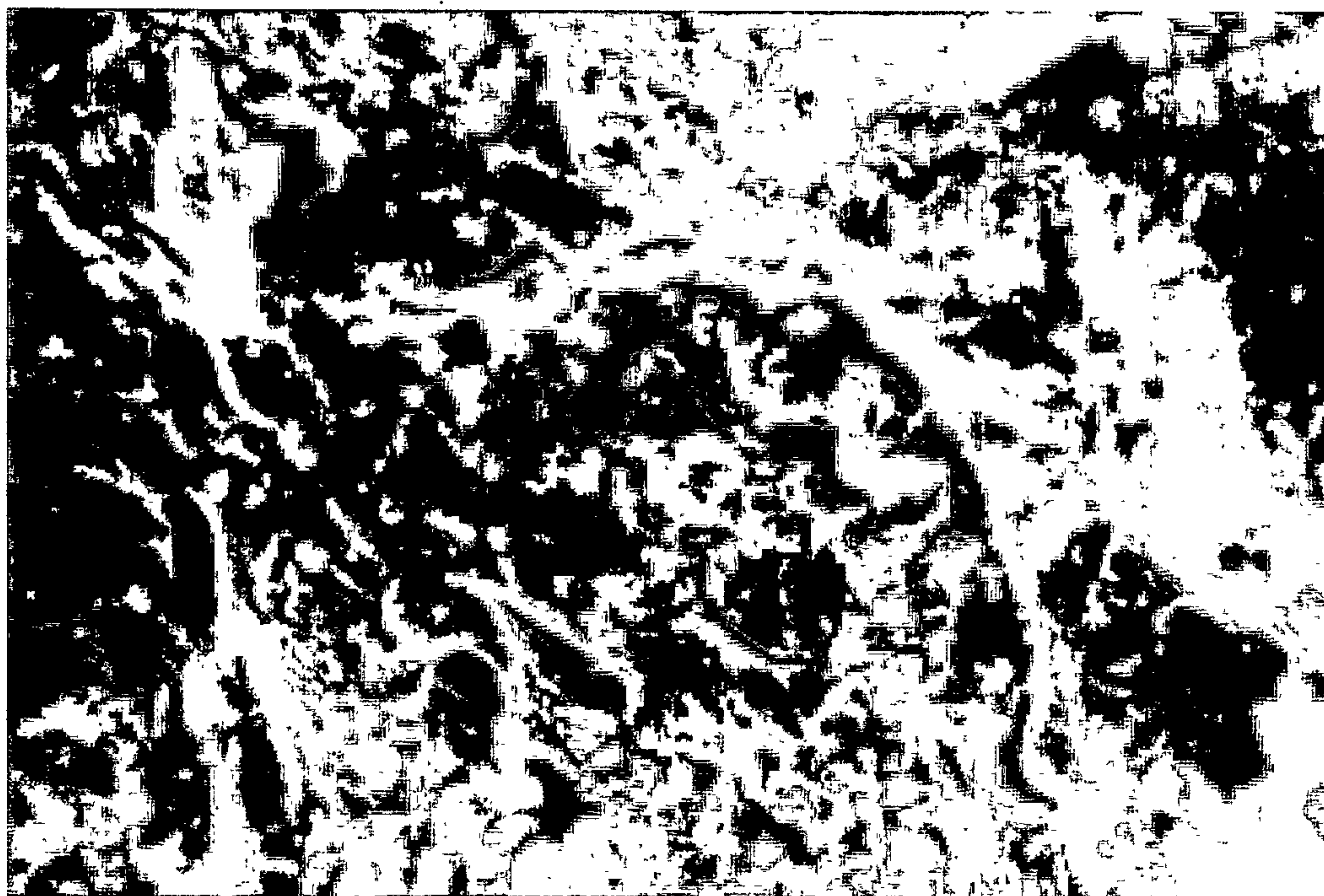
The recent introduction of the pressed leucite reinforced ceramic system, IPS Empress, has leucite in a different role. This material relies on an increased volume of fine leucite particles to increase flexural strength because of the significant residual stresses and

associated crack deflection mechanisms possible. A scanning electron micrograph demonstrating this structure can be seen in figure 2.6. This type of ceramic can be used as a monolithic construction of the crown without the need for a veneering layer. The final colours are created with coloured glazes.



————— 10  $\mu\text{m}$

Figure 2.5. Scanning electron micrograph of a leucite containing veneering porcelain, Duceram®. Original magnification 2000x.



— 1  $\mu\text{m}$

Figure 2.6. Scanning electron micrograph of a leucite containing core porcelain, IPS Empress®. This unpolished view of a fractured surface shows a cluster of leucite crystals in a cavity in the centre of the micrograph. Original magnification 10,000x.

### 2.2.3 The Castable Glasses

The development of glass ceramics by the Corning Glass Works in the late 1950's has led to the creation of a dental ceramic system based on the strengthening and toughening of glass with the nucleation and growth of various forms of mica, (Grossman 1985). The Dicor® crown system has a base chemical composition of  $\text{SiO}_2$ ,  $\text{K}_2\text{O}$ ,  $\text{MgO}$ ,  $\text{Al}_2\text{O}_3$ , and  $\text{ZrO}_2$  as well as a fluoride component and a fluorescing agent, (Grossman 1985). The result is a tetrasilicic mica glass-ceramic. An advantage of this system was its hardness value. Measured using a Knoop indenter, the hardness was similar to dental enamel which was seen as advantageous with respect to wear, (Naylor et. al., 1991)

The dental laboratory construction uses the lost wax system to produce a glass casting of the restoration. The casting is then heat-treated or "cerammed", during which tetra silicic fluoromica crystals are formed to increase the strength and toughness of the glass ceramic. This procedure is designed to take place within the economic confines of a commercial dental laboratory. A second dental version was developed to be used for CAD/CAM dental procedures. This cerammed glass is provided in an already heat-treated state from the manufacturer. Although intended for similar dental restorations, they have quite different structures as can be seen in figures 2.7 and 2.8.



Figure 2.7. Scanning electron micrograph of the castable glass ceramic, Dicor®. This demonstrates the crystalline structure near the surface of the casting. Original magnification 20,000x.



\_\_\_\_\_ 1  $\mu$ m

Figure 2.8. Scanning electron micrograph of the machinable glass ceramic, Dicor MGC®. This micrograph demonstrates the fluoro-mica platelets deeper into the Dicor® structure. Original magnification 20,000x.

## **2.2.4 The Alumina Based Systems**

The driving force for the development of new alumina based cores has been the immense difference in reliability between metal-ceramic systems and all-ceramic systems and a public perception that metal-free restorations are more aesthetic. The disadvantages of the metal ceramic systems include radiopacity, some questions centring around metal biocompatibility and lack of natural aesthetics; important features in today's consumer conscious dental market.

### **2.2.4.1 The Aluminous Jacket Crown**

The modern Aluminous Jacket Crown, probably more commonly known as the Porcelain Jacket Crown (PJC) was popularized in the mid 1960's by McLean (1979). This report also points out the importance for the use of alumina in dental ceramics and how it modifies the flaw systems at the surface and within the ceramic. The alumina is added to provide dispersion strengthening to resist the deepening of surface micro-cracks. During firing for the manufacture of the porcelain powder, the glassy phase is melted and flows around the alumina crystals forming an ionic bond between the matrix (glass) and the crystals (alumina) (Hondrum 1992). Basically the alumina crystals are still bound within a glassy matrix but at a concentration of 40% to 50% by volume which increases the elastic modulus of the core material by approximately 50% (Jones 1988). By using the increased alumina concentration, these tough crystals act as barriers or deflectors for crack pathways, requiring more energy to be expended for the crack to proceed (Hasselman 1966) and Dinsdale (1967). The aluminous porcelains are also very prone to strength degradation when they contain porosity, McLean (1979).

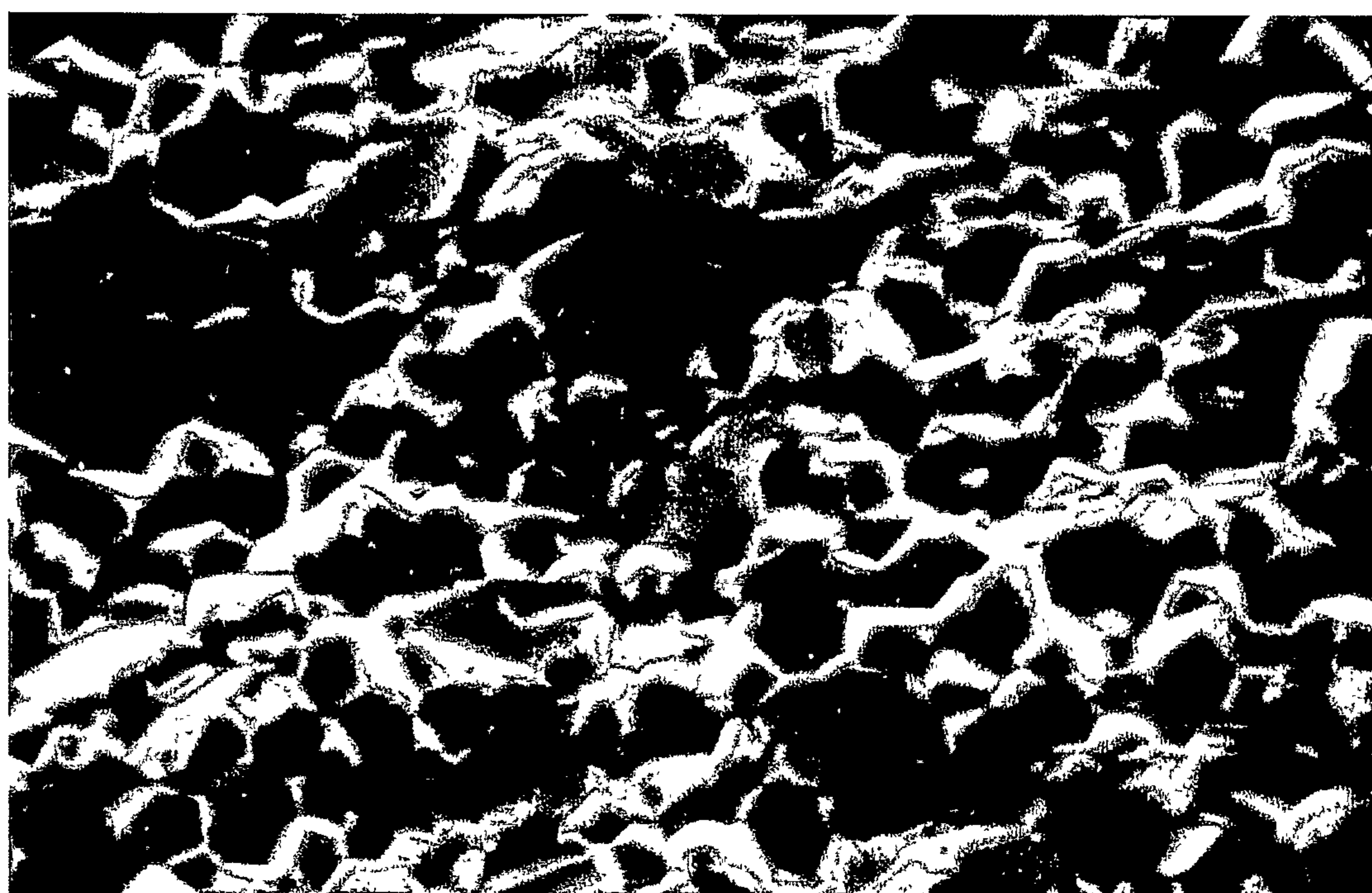
### **2.2.4.2 The Pure Alumina Cores**

The attraction for alumina is provided by the vast knowledge for its use industrially and that it can be made to fit accurately. It can be easily adapted, is biocompatible and has high bending strength. This is coupled with the knowledge that in a sintered state the alumina has a degree of translucency. (Prestipino 1993)

#### **2.2.4.2.1 Heat Cured After Pressing**

The Nobel Biocare company from Sweden have introduced two systems that essentially use a system of pressing alumina onto a metal die, removing the pressed shape from the die and then sintering it. One system is used to make alumina profiles that are then used as cores to build up ceramic superstructures for single tooth implants, CeraOne®, and the second is to make cores for conventional crowns, a process known as Procera®.

Essentially there is no difference between the alumina in these two systems. Both systems use pure alumina for the core. Figure 2.9 demonstrates classical hexagonal alumina crystals within this material. Unlike the other materials, there is no glassy phase present. Veneering porcelains such as Vitadur Alpha® and Duceram® are then fired onto this alumina core to provide the colour and form for the restoration.



— 10  $\mu$ m

Figure 2.9 Scanning electron micrograph of the alumina structure of a CeraOne® core. The original magnification was 5000x.

#### 2.2.4.2.2 The Glass-Infiltrated Alumina System for Cores

During the 1980's, Dr. Michael Sadoun of the Paris number 5 University, in association with the German dental company, Vita Zahnfabrik, developed a slip casting system using fine-grained alumina. The cast alumina was sintered and then infiltrated with a Lanthana based glass. Figures 2.10 and 2.11 demonstrate the material before and after glass infiltration. This provided a glass infiltrated alumina core on which a feldspathic ceramic could be baked to provide the functional form and aesthetic component of the restoration, (Claus, 1990). This system will be used for the model in this thesis because the initial reports indicate that the core material known as "In-Ceram"® has one of the highest flexural strengths and fracture toughness values of all the currently available dental

ceramic systems available to most commercial dental laboratories, (Seghi et al., 1990; Wassenaar 1990, 1991a and b; Kern et al., 1994, Seghi et al., 1995). The system also has the greatest versatility for dental use. The material is also characterized by its great accuracy of fit and dimensional stability through the many firing cycles often associated with the veneer build-up, Wohlwend et al. (1989).

The use of the slip cast and lanthana based glass technique provided several advantages. The first was to allow accurate shaping of the core both before and after initial sintering. After the initial sintering when the alumina crystals are lightly sintered together, the core framework can be tried on the master dies and any further adjustments made with ease due to the low flexural strength of the material. This was measured using 12 mm discs and the bi-axial testing protocol described in chapter four, at approximately 55 MPa. The lanthana based glass has approximately 15.6 % alumina which after infiltration helps increase the alumina content of the In-Ceram. The lanthana ( $\text{La}_2\text{O}_3$ ) is said to improve the alkali resistance of the glass, Hornberger (1998). The overall alumina content has been reported variously as being between 80% and 90%, Hornberger (1998).

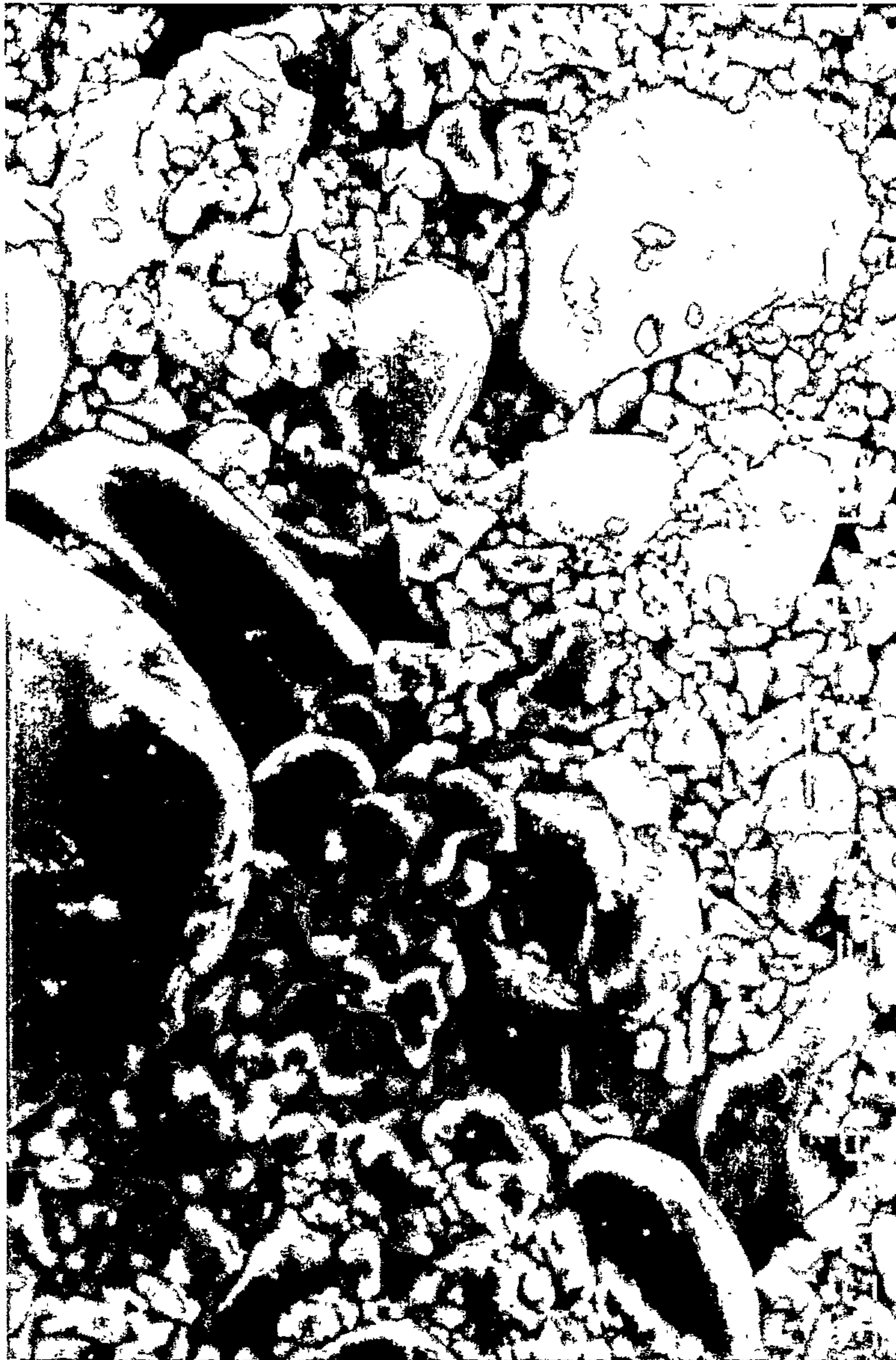


Figure 2.10 Scanning electron micrograph of the sintered alumina substructure of In-Ceram® before glass infiltration. Note the critical mix of large and small particles of alumina. Original magnification 10,000x.

The In-Ceram alumina core material has a blend of high purity alumina particles with a particle size distribution that can be seen in the graph below. The average particle size is approximately 3.8 μm, however the wide range of sizes facilitates the dense packing that can be seen in the electron micrograph above.

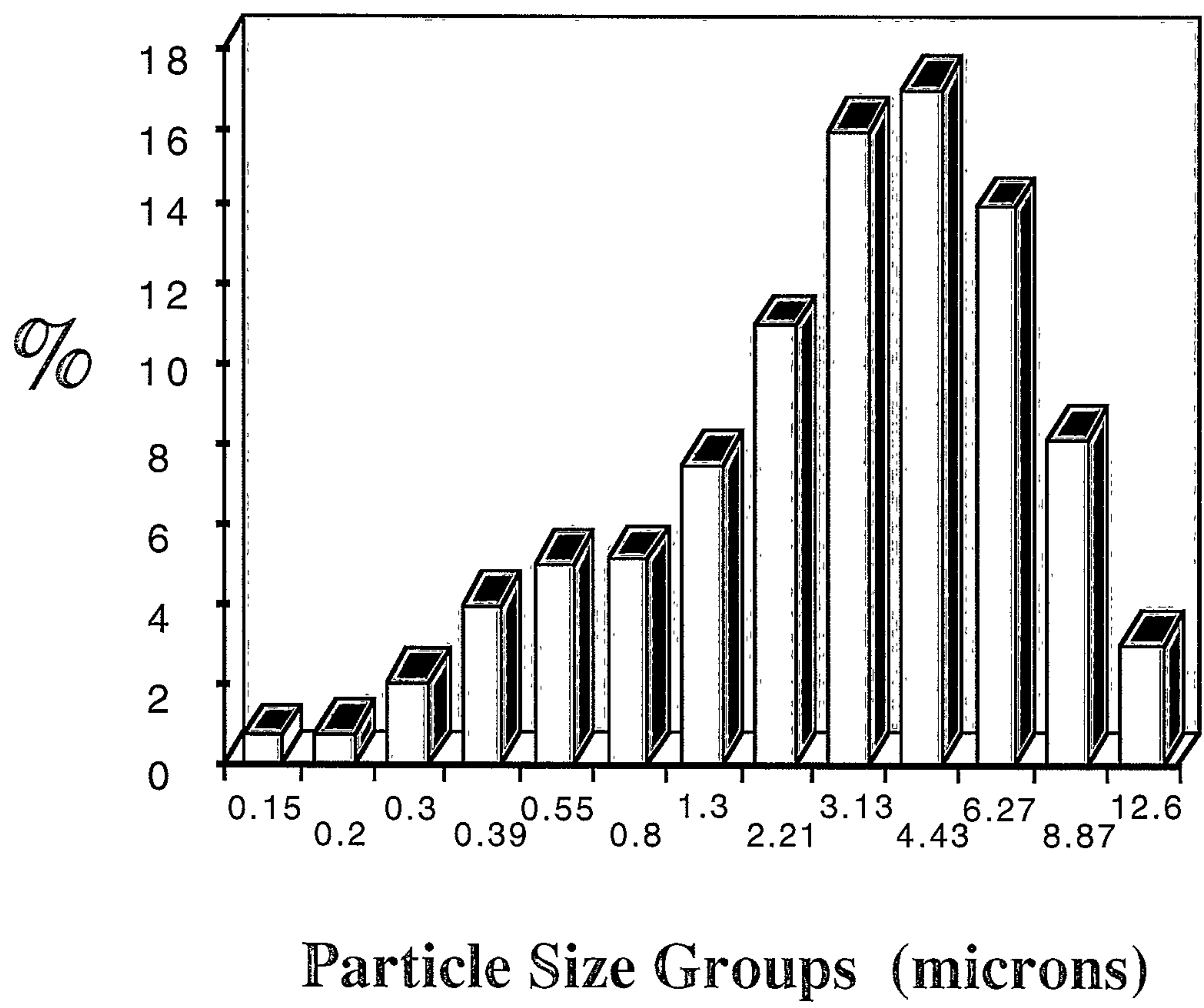




Figure 2.11 Scanning electron micrograph of the sintered alumina substructure of In-Ceram® after glass infiltration. This view of In-Ceram demonstrates the high saturation of the core with minimal porosity present. An example of a residual pore can be seen circled in the lower right corner of the electron micrograph. Original Magnification 25,000x

### 2.3 Failure Mechanisms in Dental Porcelain

Following on from the initial examples of clinical failure given in chapter one including the photographs in figures 1.4 to 1.7, there are reports that ceramic restorations such as inlays and crowns fail due to occlusal injury not unlike that of a small spherical indentation. This has led to at least one group using a small spherical indenter of 2 millimetres diameter as a testing device to produce similar failures in vitro, (Dietschi et al., 1990). Perhaps the most important consideration to note is the possible crack driving forces associated with tensile stresses formed from elastic displacement components within the field of the contact, (Marshall et al., 1979).

Another consideration is the choice of indenter to produce failure. In general, the contacts between teeth are the same as small spherical indenters, changing with age and wear, and becoming broader, leading to flatter and larger areas of contact. The cusps of teeth are naturally rounded. As teeth become worn they tend to exhibit small milled facets and so the contacts can become much broader. A spherical indenter rather than a sharp indenter would therefore be the system of choice when measuring the behaviour of the ceramic system in question.

### 2.4 Fracture Mechanics

This area of physics quantifies the effect of cracks on materials. It answers specific problems concerning whether cracks will extend or arrest in physical or solid structures, (Ewald and Wanhill 1984). Dental ceramics are essentially brittle materials which justifies the use of linear fracture mechanics. During loading these ceramics obey Hooke's law, where stress and strain are proportional and it is only a very small field around the advancing crack tip that possibly undergoes plastic deformation, (Mencík 1996). This field can also be considered at the molecular level in terms of size.

In linear elastic fracture mechanics, there are considered to be three independent kinematic movements of the upper and lower crack surfaces with respect to each other. This is demonstrated in figure 2.12. These are categorized as simple crack opening (mode I), opening by shear (mode II), and opening by tearing (mode III) (Owen and Fawkes, 1983).

*Mode I (opening mode):* in which the two crack surfaces are pulled apart in the  $y$ -direction, but where the deformations are symmetric about the  $x$ - $z$  and  $x$ - $y$  planes.

*Mode II (shearing mode):* in which the two crack surfaces slide over each other in the  $x$ -direction, but where the deformations are symmetric about the  $x$ - $y$  plane and skew symmetric about the  $x$ - $z$  plane.

*Mode III (tearing mode)*: in which the crack surfaces slide over each other in the  $z$ -direction, but where the deformations are skew symmetric about the  $x$ - $y$  and  $x$ - $z$  planes.

The most important case in practice is mode I. This mode is usually indicated by the  $I$  in the fracture mechanics formulae, written as a subscript to the stress intensity factor  $K$ , i.e.  $K_I$ . In the presence of thermal or residual stresses, or in non-homogeneous materials, mode II is often present (Mencík 1996). Considering the complex structures used in some of the current dental ceramics, it is possible that all three cases can be superimposed. The glassy materials such as Vitadur Alpha® are most likely to fail entirely in type I mode whereas the more complex In-Ceram® may contain aspects of all three. Refer to figures 2.3 and 2.11. for examples of the simple structure of Vitadur Alpha® versus the complexity of In-Ceram®. The diagram in figure 2.13 summarizes the respective structures of these materials. The emphasis is on the presence of glass matrix surrounding the filler particles in Vitadur Alpha, versus the intimate contact between particle in In-Ceram. Cracking in Vitadur Alpha would probably travel through the glassy matrix, whereas cracking in the In-Ceram would need to break the joins between the alumina particles; overcome the residual stress mismatch between the two phase, and overcome friction between opposite sides of the crack. The latter aspect of reinforcement could be caused by the presence of elongated alumina crystals lying across the crack plane or binding and rubbing where the crack plane is forced to undergo a severe change in direction.

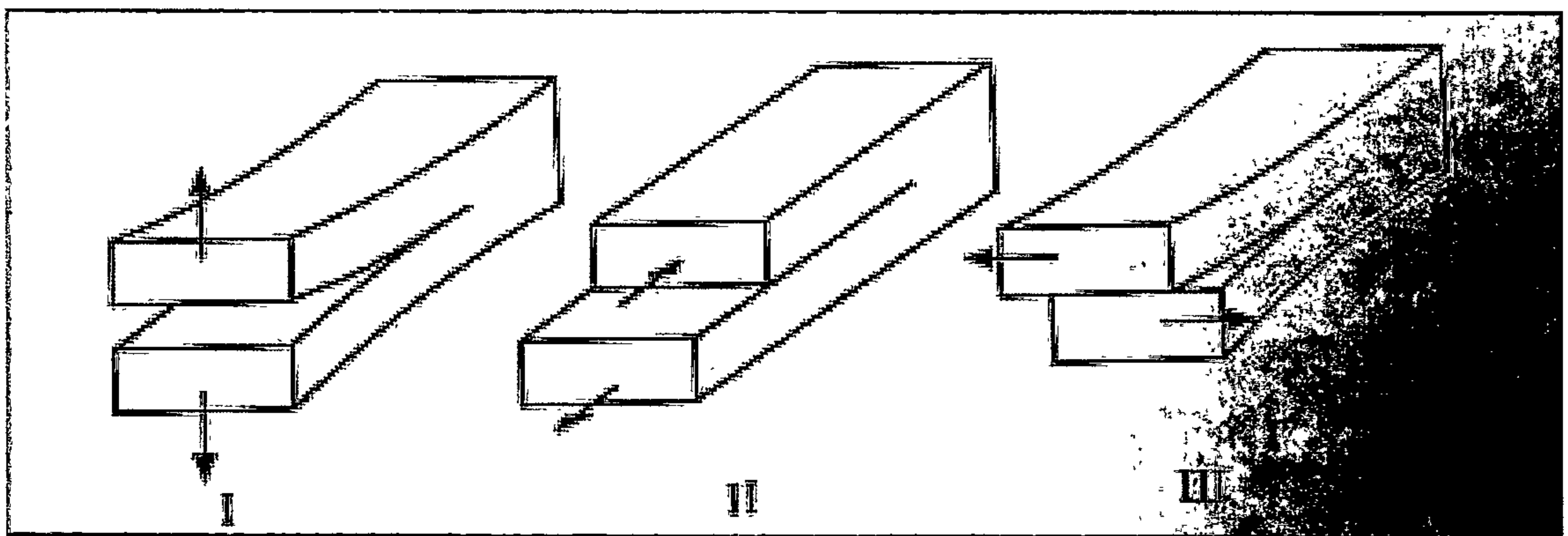


Figure 2.12 Three principal modes of fracture: (a) Opening mode I; (b) Shearing mode II; (c) Tearing mode III

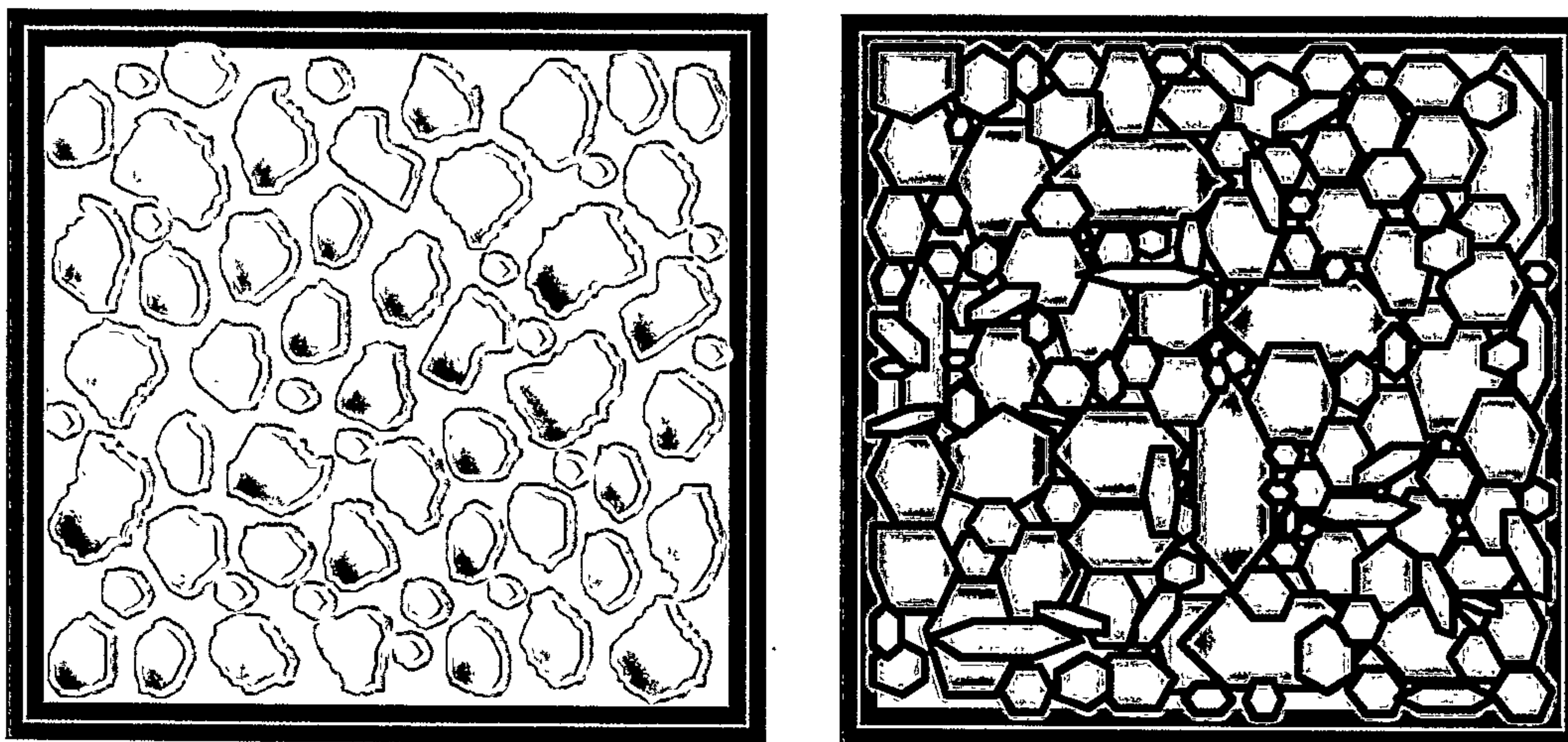


Figure 2.13 This schematic diagram refers to the glassy matrix and crystalline filler of Vitadur Alpha® on the left and In-Ceram® on the right.

### 2.4.1 Crack Initiation

This study aims to investigate the process of crack initiation from spherical indentation *in vitro*. However, classical work, from Hertz, Griffith and others needs to be reviewed, in order that an appropriate insight to the contact induced tensile stresses and crack extension criteria in particular may be developed. For a Hertzian crack to pop in, the tensile stress at the contact surface needs to be exceeded, However for a body of finite dimension as we find in the thickness of a dental crown, while the contact load is being applied, the body of the material is undergoing strain. The surface on the opposite side of the object will be flexing and developing its own tensile stresses and when they exceed the flexural strength (tensile strength) of the material, fractures from this surface are likely to occur.

The pioneering work in the field of contact induced fracture by Griffith, Inglis, Auerbach and Hertz has again been critically reviewed by Lawn (1993). The following paragraphs attempt to both summarize this work and introduce it to support later arguments describing a method of failure for the In-Ceram ceramic system. The clinical failures from figures 1.4 to 1.7 would seem to indicate both Hertzian cone cracks and radial cracks are involved.

#### 2.4.1.1 The Griffith Concept

This concept is based on energy theories that extension of a crack is reliant on classical mechanics and thermodynamics. For this summary I have sought extensively from Lawn (1993). The Griffith energy balance concept for equilibrium fracture considers the overall energy of the system as  $U$  and then considers the component parts if the crack is to undergo extension. The components are the mechanical part, which is the load applied from outside the system, e.g. a cusp or an indenter about to bear down onto the porcelain.

The second part is the surface energy required to form the new surface area as the crack extends. This is where molecular bonds are broken by the strain of the applied load and need to reform to provide a new stable surface for the porcelain either side of the crack. Therefore the mechanical energy can be expressed as:

$$U_M \quad (2.1)$$

This is also broken down into component parts of the strain potential energy of the elastic medium ( $U_A$ ) as well as the potential energy of the outer loading system. As these two are opposites, because action and reaction are opposites, then the potential energy of the outer load ( $U_E$ ) is seen as the negative of the strain in the loaded body. Therefore:

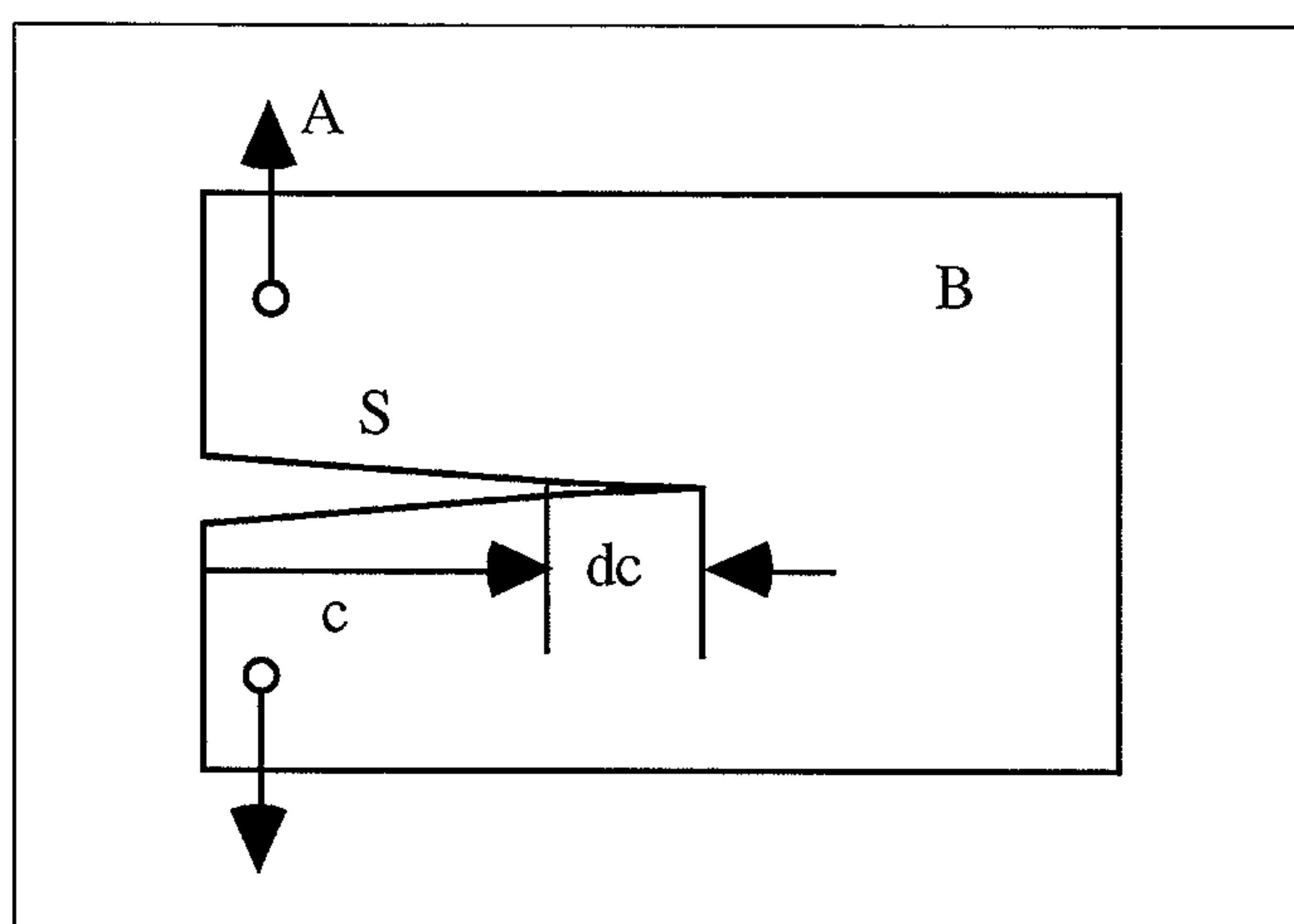
$$U_M = U_E + U_A \quad (2.2)$$

The term for the energy expended to create the new crack surfaces is also introduced as:

$$U_S \quad (2.3)$$

Therefore for the equation to remain balanced the surface free energy plus the mechanical energy must equal the overall energy of the system:

$$U = U_M + U_S \quad (2.4)$$



System Boundary

Figure 2.14. Schematic illustration which describes a static plane-crack system, with incremental extension of crack length  $c$  through  $dc$ : B, elastic body; S, crack surface; A, applied loading. (Adapted from Lawn 1993)

The thermodynamic equilibrium is reached when there is a balance between the mechanical and surface energy terms over a virtual crack extension, which is represented by the length  $dc$  in figure 2.14. For this to happen either the load applied ceases or no longer has enough energy to overcome the constraints of the crack by the elastic body. These constraints being provided by the stiffness of the material as a product of its molecular bonding. This also assumes that the material is free to move at its outer surfaces and is not in itself constrained by an outer reinforcing material. The mechanical energy then decreases as the crack extends:

$$dU_M/dc < 0 \quad (2.5)$$

Correspondingly, the surface energy will generally increase with crack extension as the cohesive forces of molecular attraction across crack extension  $dc$  are overcome to create the new fracture surfaces:

$$dU_S/dc > 0 \quad (2.6)$$

This means that  $U_M$  favours crack extension and  $U_S$  opposes it.

Given this, then a formal statement for the Griffith energy balance concept can be expressed as:

$$dU/dc = 0 \quad (2.7)$$

Griffith then explored the concept of a crack in uniform tension using glass as his model because it was considered isotropic and basically follows Hooke's law until failure. This then provided relationships to determine surface energy of the crack, which Griffith expressed in terms of unit width of front as:

$$U_S = 4c\gamma \quad (2.8)$$

where  $\gamma$  is the free surface energy per unit area. The total system energy becomes:

$$U(c) = -\pi c^2 \sigma_A^2 / E' + 4c\gamma \quad (2.9)$$

This leads to the critical conditions for failure to occur by combining the formal Griffith energy balance equation or equilibrium condition with the total system energy:

$$\sigma_F = (2E' \gamma / \pi c_0)^{1/2} \quad (2.10)$$

where  $\sigma_F$  is the critical stress at failure.

Further experimentation with glass tubes and rods determined that the stress component normal to the crack plane was important and led to the expression:

$$\sigma_F c_0^{1/2} = 0.26 MPa.m^{1/2} \quad (2.11)$$

Following from this it is relatively easy to see that the stiffness or more precisely the Young's Modulus (E) is an important aspect in the overall strength of a ceramic. In addition the crystalline content, its density and geometry can further add to the strength of the material in a number of ways. The first aspect of crystalline content needs mention because this method involves using a relatively tough ceramic, where the crack will deviate around stronger particles, following a less cohesively bonded material. The deviations extend the overall crack length and surface area for a given depth of penetration measured perpendicular from the surface. The multi-angular pathway provides a second possibility for crack growth retardation, friction. The friction comes from crack walls binding as they pull apart, but are prevented from moving freely by the very nature of their tortuous path. And finally there is the possibility that the particles themselves may traverse a crack line and tend to hold it together in a fashion similar to a reinforcing fibre.

#### 2.4.1.2. Fracture Toughness.

The role of fracture toughness as a physical parameter for the study of dental ceramics has not been widely appreciated until the 1970's. Since then a number of authors, such as Dunn et al., (1977) Anusavice et al., (1991), Denry et al., (1993) have used fracture toughness measurements in their assessment of ceramic properties. The concept of fracture toughness and the idea of energy expenditure are important in considering the long-term serviceability for dental restorations.

Where there is a crack present in a given material, and it is stable, there must be forces that are insufficient to encourage the advance of the crack tip. When the crack advances, more energy is released than is absorbed ( $U_s$  being that part of the energy balance equation representing the energy released to create the new surfaces). This happens when enough energy is built up due to stress and overcomes the ability of the material to absorb it one way or another, that is, a crack reaches a critical size, and fast fracture occurs. The toughness of a material is the energy absorbed in creating unit area of crack. An alternative approach to the thermodynamic relationship developed by Griffith is the stress intensity concept. This relates to the magnitude of the stress at the crack tip. The stress intensity factor is that point where the combination of stress and crack length reach a critical level from which fast fracture takes over. This is described by Irwin (1957) as:

$$K_1 \geq K_{1c} = [Y(\theta)]\sigma\sqrt{a} \quad (2.12)$$

where  $K_{Ic}$  is in many cases a material constant and is considered a measure of the toughness of the material.

The term for *toughness* is  $G_c$  and the units are  $J\ m^{-2}$ . The term for *fracture toughness* is  $K_c$ , also known as the critical stress intensity factor and the units are generally  $MN\ m^{-3/2}$  or  $MPa\ m^{1/2}$ . When fast fracture occurs,  $K = K_c$ . (Ashby et al., 1980). Fast fracture can also be viewed as the release of mechanical energy to produce kinetic energy, which is dissipated by the energy consumption in forming the area of the new fracture surfaces. The release and conversion of energy then relates back to the energy balance work of Griffith where the expression for the critical condition, or stress level for failure to occur is expressed in terms of the energy of the system. The mechanical-energy-release rate is known as  $G$  and can be expressed as:

$$G = -dU_M/dC \quad (2.13)$$

with  $C$  the crack interfacial area.  $G$  has the dimensions of energy per unit area. The process of forming the new surfaces is governed by a “work rate”, which is an expression of the energetics required to overcome the intersurface forces and separate the newly formed crack surfaces. The work rate per unit area is represented by the term  $R_0$  and expressed as:

$$R_0 = +dU_s/dC \quad (2.14)$$

The relationship between the mechanical-energy-release rate as a measure of toughness ( $G_c$ ) and the critical stress intensity factor ( $K_c$ ) then becomes:

$$G_c = K_c^2/E' = R_0 \quad (2.15)$$

When using brittle substances such as dental ceramics for crown and bridge restorations where the size of the system is quite small, an expression relating strength with the critical stress intensity factor and defect or crack is given by:

$$K_{Ic} = 1.24\sigma\sqrt{a} \quad (2.16)$$

where  $\sigma$  is the applied stress, and

$a$  is the dimension of the surface crack.

Some typical values of the critical stress intensity factor for dental ceramic materials are listed in table 2.3 below.

**Table 2.3      Examples of typical stress intensity factors for Dental Porcelain and a selection of other comparable materials. (Wiskott et al. 2001)**

| Material                   | Fracture Toughness, $K_{Ic}$ (MPa m <sup>1/2</sup> ) |
|----------------------------|------------------------------------------------------|
| Feldspathic Ceramics       | 1                                                    |
| Alumina                    | 4.1                                                  |
| In-Ceram (Alumina version) | 4 - 6                                                |
| Tooth dentine              | 3.1                                                  |
| Bone                       | 3.6                                                  |
| Dental Composite Resins    | < 4                                                  |
| Dental Amalgam             | 2.9                                                  |
| Glass Fibre Epoxy          | 60                                                   |
| Hard Steels                | 40                                                   |
| Ti6Al4V                    | 80                                                   |

Hassan et al., (1981) measured human enamel fracture toughness using a micro-indentation technique and found a range from 0.7 to 1.27 MPa m<sup>1/2</sup>. This is similar to the values for feldspathic ceramics used in dentistry as veneering porcelains.

An experimental slip cast alumina using boro-silicate glass as opposed to the lanthana based glass of In-Ceram® was found to have a fracture toughness of 3.8 MPa m<sup>1/2</sup>, (Wolf et al., 1993).

A further method for measuring fracture toughness uses indentation techniques (Evans and Charles 1976, Marshall et al., 1979, Jones et al., 1988b, Kvam 1992). However the techniques can be influenced by the very nature of the material, whether it is homogenous or not and the preparation of the surface. It is also dependent on accurately determining the elastic/plastic indentation fracture mechanics of the material, Anstis et al., 1981, Chantikul et al., 1981).

**2.4.1.3 Hertzian Fracture.**

Using the spherical indenter as a model the first step is to examine the stresses produced at the contact point and how they build to overcome the tensile strength of the material. Rather than looking at a pre-existing crack and its extension, the work of Hertz looked at

the introduction of cracks in brittle materials. Hertz's description of the cone crack has become a model for crack initiation, particularly in monophase materials such as soda-lime glass.

The classic Hertzian cone crack can be seen in the following diagram, figure 2.15.

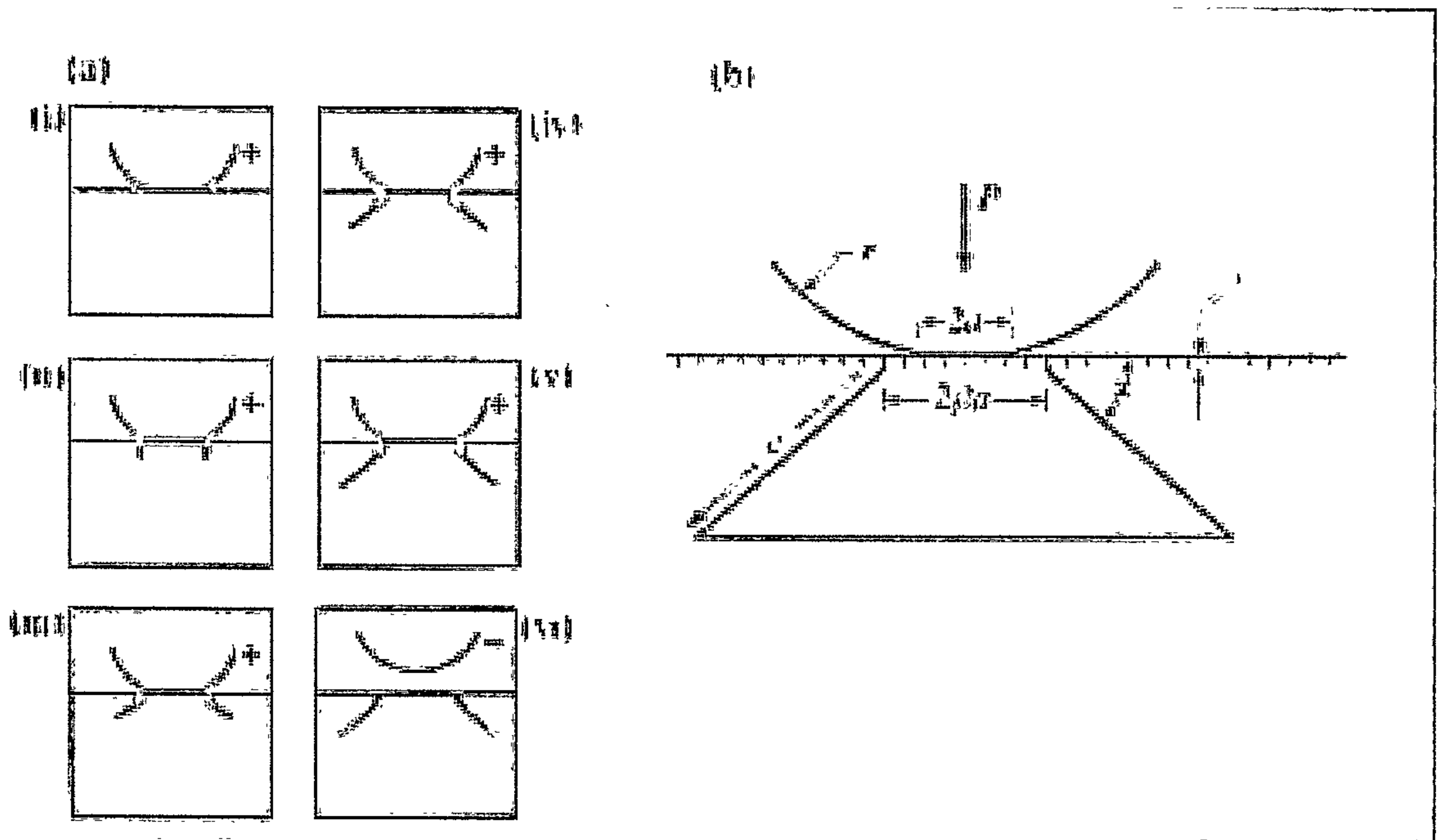


Figure 2.15. Classic description of the Hertzian cone crack system adapted from Lawn (1993). (a) His description demonstrates the evolution of the cone crack during loading and unloading. (b) The geometrical parameters.

This type of cone crack was noticed within the failed clinical examples and was easily reproduced for the in vitro trials (see chapter three).

The radius of the circle of contact  $a$  is given by equation (2.17) below:

$$a^2 = \frac{4}{3} \frac{kPR}{E} \quad (2.17)$$

$P$  is the indenter load,  $R$  is the indenter radius,  $E$  is the Young's Modulus for the specimen and  $k$  is an elastic mismatch factor given by equation (2.18) below:

$$k = \frac{9}{16} \left[ (1 - \nu^2) + \frac{E}{E'} (1 - \nu'^2) \right] \quad (2.18)$$

Here,  $\nu$  is the Poisson's ratio of the specimen and  $\nu'$  and  $E'$  are the Poisson's ratio and Young's Modulus respectively for the Indenter.

Hertz also found that the maximum tensile stress in the specimen occurs at the edge of the contact circle (Fischer-Cripps 1998) on the specimen surface and is given by equation (2.19):

$$\sigma_{\max} = (1 - 2\nu) \frac{P}{2\pi a^2} \quad (2.19)$$

This is the stress that is considered as being responsible for the production of Hertzian cone cracks. It acts in a radial direction on the surface outside the indenter and decreases as the inverse square of the distance away from the centre of contact. By combining equations (2.17) and (2.19) the maximum tensile stress outside the indenter can be expressed in terms of the indenter radius  $R$ :

$$\sigma_{\max} = \left( \frac{(1 - 2\nu)P}{2\pi} \right) \left( \frac{3E}{4k} \right)^{2/3} P^{1/3} R^{-2/3} \quad (2.20)$$

#### 2.4.1.4 Auerbach

The "Auerbach Law" that developed from Auerbach's work shortly after Hertz has been referred to since as a paradox in fracture theory (Lawn, 1993) because it implied that fracture did not occur at a critical tensile stress. The Auerbach concept as its principal advocate argued that the initiation of the cone crack was due to a critical load which was proportional to the radius of the indenter, i.e.  $P_c \sim R$ . The law itself is written as  $P = AR$ , where  $P$  is the force required to produce a cone crack,  $R$  is the indenter radius and  $A$  is termed the Auerbach constant. The initial basis for explanation of this relationship was the presence of surface flaws, and that the larger the radius of the indenter, the higher the probability that it might incur a surface flaw that would become the crack initiation site, (Fischer-Cripps, 1998). Lawn (1993) points out that because fracture can initiate when the maximum stress in the Hertzian field just equals the bulk strength of the material, then the critical load  $P_c$  is proportional to the square of the radius,  $r^2$ .

#### 2.4.1.5 The effect of sliding contact

Considered in the simplest form, that is in this case that a blunt or spherical indenter passes across a flat surface following initial contact, that a series of partial cone cracks develop in the trailing path due to the tensile stresses generated (Lawn, 1993; Swain, 1994). In the oral environment, tooth-to-tooth contact is usually followed by a sliding motion. The effects of this ultimately lead to surface breakdown and wear. As reported by Lawn (1993) exposed surfaces of brittle solids are susceptible to damage by the minutest extraneous mechanical agents. The best-prepared surfaces of these brittle solids (i.e. in this case Dental Ceramics) will contain defects from mechanical preparation. Lawn further

points out that under normal conditions even dust particles and other hard materials in the environment will cause minute Hertzian elastic-plastic contact flaws. Considering the function of the mouth as a means for the initial pulverizing of food stuffs and the contact between teeth that occurs during bruxism, there is ample opportunity for these microflaws to propagate. As an example we need to consider figure 2.16 below adapted from Lawn (1993)

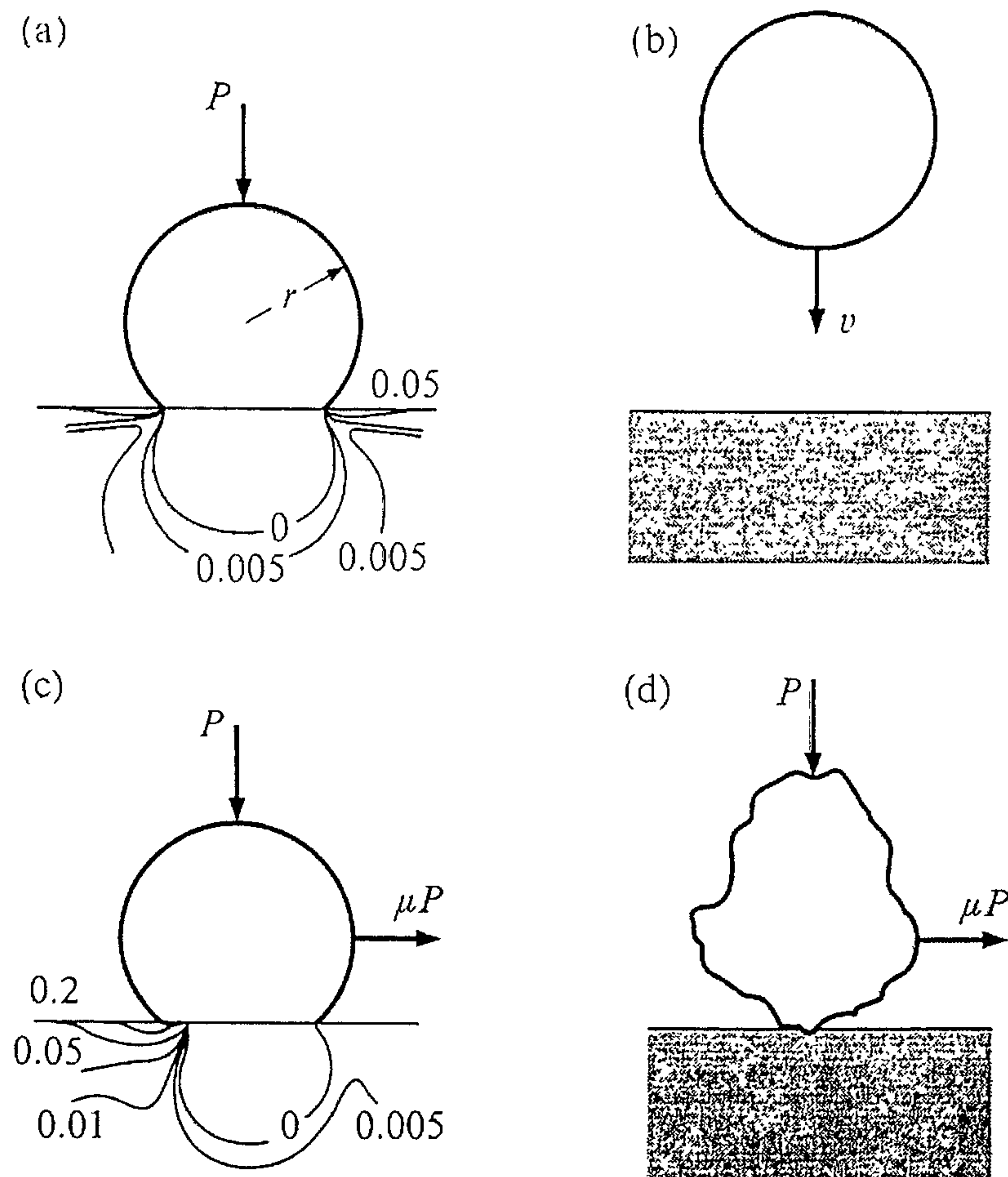


Figure 2.16 Contact damage model: (a) static blunt particle, sphere radius  $r$  and load  $P$  (Hertz); (b) impacting sphere, at normal velocity  $v$ ; (c) sliding sphere, with tangential force  $\mu P$  ( $\mu$  = friction coefficient) superimposed on normal force; (d) irregular particle. Equi-stress contours (units of mean normal contact pressure) indicated in (a) and (c). (Lawn 1993)

Based on this, Lawn points out that it is possible to pop in cone cracks at loads well below the limits of validity for Auerbach's law.

The Hertz elasticity analysis shows that the contact pressure increases monotonically with the expanding contact circle.

$$p_o = P / \pi a^2 = (3E / 4\pi k) (a / r) \quad (2.21)$$

$P_o$  is the mean contact pressure and  $P$  is the applied point load or mean contact load in Newtons.

The equation below is the relationship between contact pressure and Poisson's ratio and shows us how the maximum tensile component of the Hertzian stresses field is dependent on the mean contact pressure and Poisson's ratio.

$$\sigma_T = \frac{1}{2}(1 - 2\nu)p_o \quad (2.22)$$

As an example of the type of flaw size that can pop in from a small solid particle then Lawn solved equations (2.21) and (2.22) to produce the following equation:

$$P_{Th} = \left[ \frac{2\pi}{(1 - 2\nu)} \right]^3 \left( \frac{4kr}{3E} \right)^2 p_{Th}^3 \quad (2.23)$$

Where  $P_{Th}$  is the theoretical indenter load and  $p_{Th}$  is the theoretical cohesive tensile strength.

Then using a quartz dust particle  $r = 1 \mu\text{m}$  on glass with  $E = 70 \text{ GPa}$ ,  $\nu = 1/3$ ,  $k = 1$ ,  $p_{Th} = E/10$  the calculation of  $P_{Th}$  approximates 1 N. By extrapolating from the data in graph (a) of figure 2.16 below it can be seen that the corresponding pop-in cone crack size is about  $8 \mu\text{m}$ .

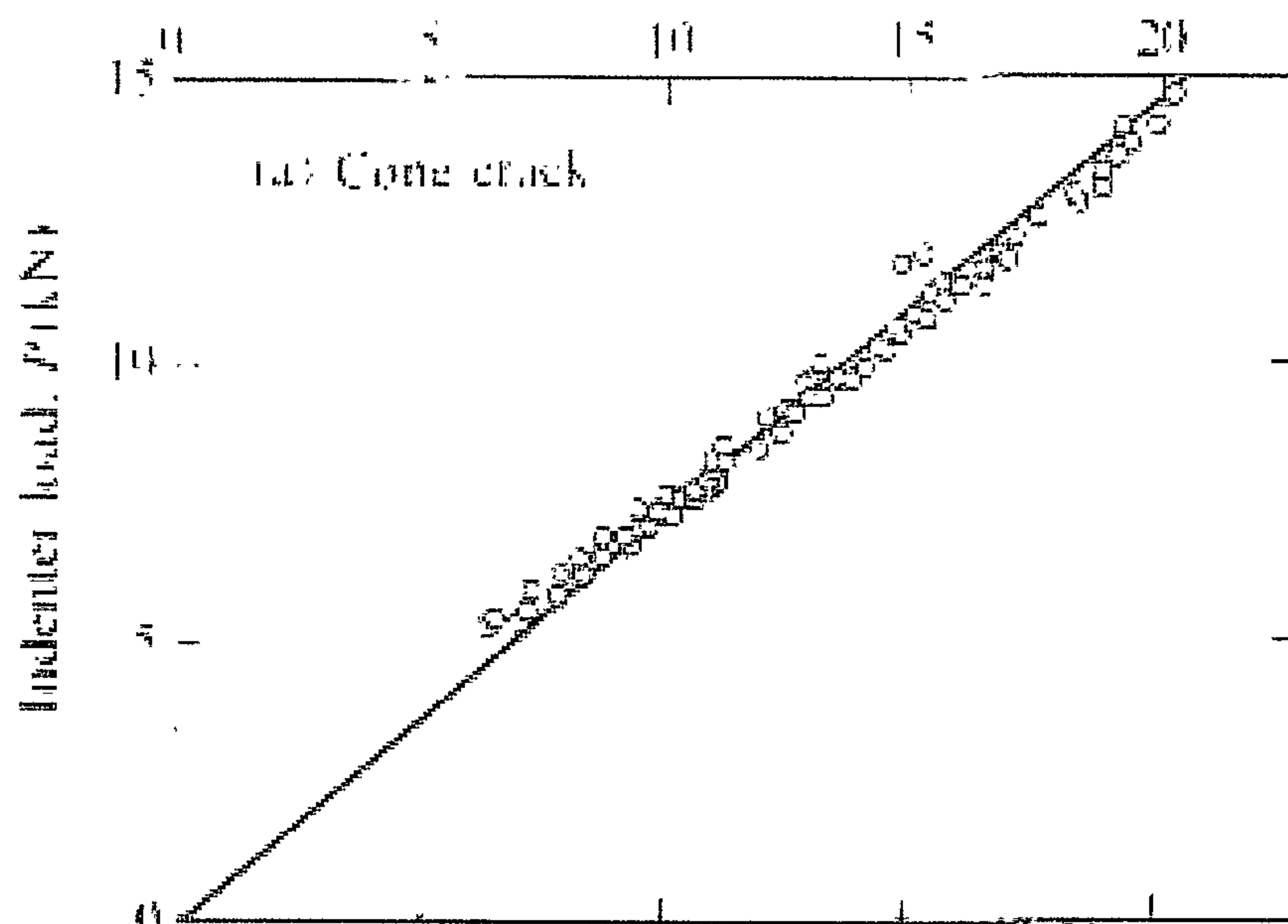


Figure 2.17 Indentation load vs characteristic crack size for cone cracks (spherical indenter with machined flat). Horizontal scale is in  $\mu\text{m}$ . (After Lawn B. R. and Marshall, D. B. (1983), in *Fracture Mechanics of Ceramics*, Eds. R.C. Bradt, A.G. Evans, D.P.H. Hasselman and F.F. Lange, Plenum, New York, Vol. 5, p. 1.)

#### 2.4.1.6 The effect of sliding contact during crown construction

The clinical implications for this are that the constant sliding contacts in the wet and often acidic environment of the mouth require a ceramic structure that is resistant to crack propagation. Fortunately, the ceramics are not soda-lime glass, however in the quest to provide colour and translucency to mimic natural teeth the veneering ceramics do tend to have lower crystalline contents and glass matrices. Therefore it is important to provide a constraining mechanism such as a rigid core material.

The flaws are further complicated by handling processes as can be seen in the photomicrograph in figure 2.18. The handling processes relate to the grinding and polishing techniques used to shape the fired porcelain. Typically after the first and subsequent firing the shape of the crown is adjusted by the technician to correct any discrepancies in form. The dentist might also grind the porcelain restoration after it has been inserted in the mouth to accommodate any discrepancies in the occlusion. In both cases the adjustments are usually accomplished using diamond burs with diameters in the order of 1 to 2 mm, in dental handpieces, at speeds often in excess of 40,000 rpm. The hard diamonds travelling at these speeds rapidly grind the porcelain away but in doing so,

remove the glazed surface and introduce numerous flaws. The depths of these flaws were measured using confocal microscopy, with some examples demonstrating crack penetrations up to 100  $\mu\text{m}$ .

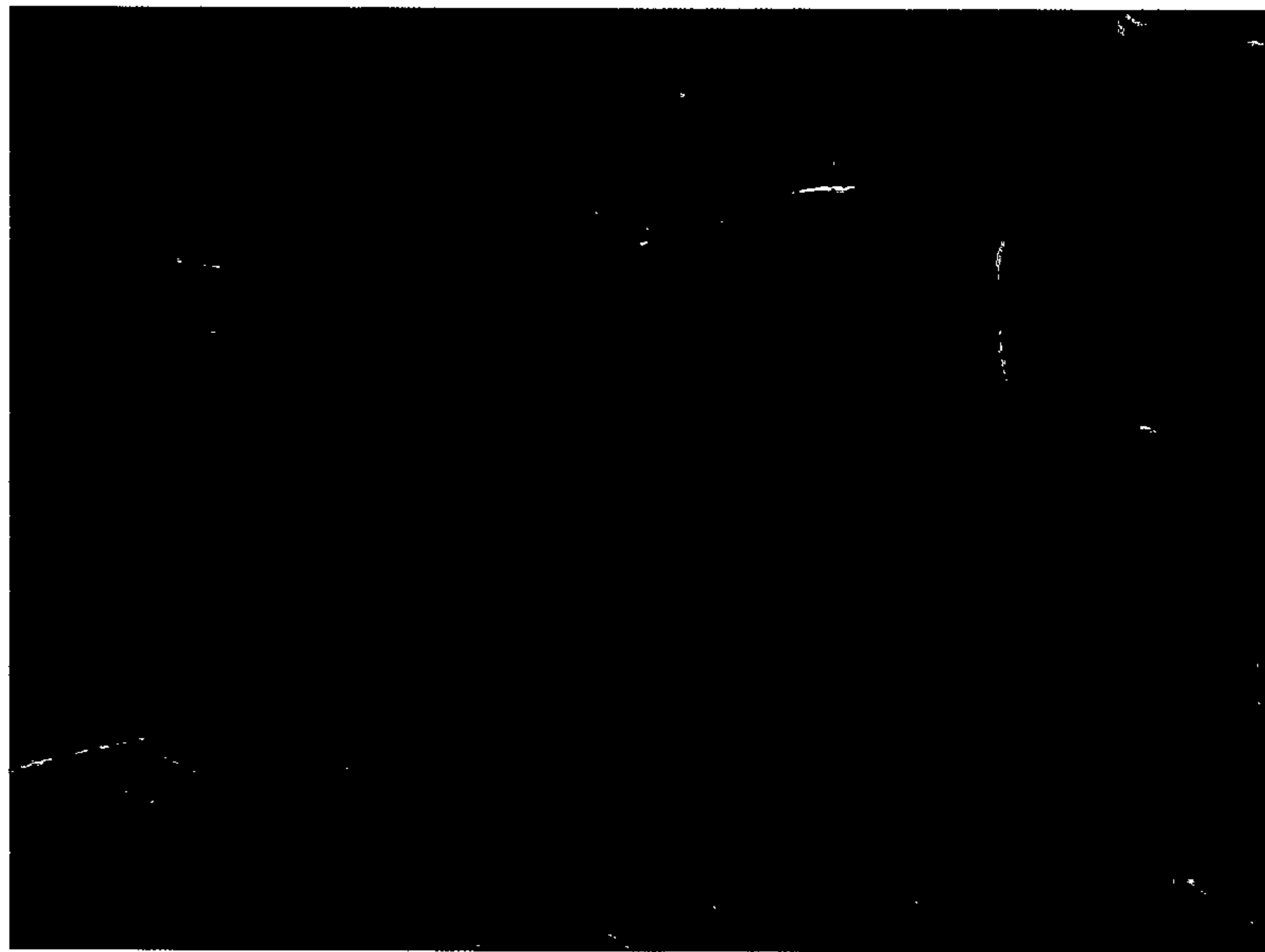


Figure 2.18 Confocal photomicrograph of Vitadur Alpha® following grinding with dental fine diamond burs and polishing with carborundum containing silicone rubber wheels. This photograph was taken with the focussing plane set at an approximate depth of 20  $\mu\text{m}$ . The cracks were stained with a fluorocine containing dye. The dye has highlighted the cracks as they cross the focal plane. Original Magnification 500x. (Ironside and Watson 1995)

#### 2.4.1.7. The effect of friction in Crack Bridging

Additional to the surface effects of friction there is also the role it may play in crack bridging. This process involves the interplay of grain boundaries behind an advancing crack tip. The grain boundaries in effect rub against grains from the opposing side of the crack, creating resistance to crack opening (Swanson et al., 1987). Friction can also act as elongated grains traverse cracks. In this instance, the "pullout" effect comes into play as these grains may be held in the opposing sides of the crack by friction.

The extent to which this relates to dental ceramics is dependent on the composition of the material. For example, the glassy veneering porcelains used for the external colour and shape of a dental crown may have very little distinct grain structure, thus providing limited opportunities for cracks to cross them or go around them and set up a scenario for a friction process. This is perhaps reflected in their low fracture toughness.

Other dental ceramics such as In-Ceram Alumina have a large number of definite grains that are elongated and do present an opportunity for friction to play a part in resistance to crack propagation.

## **2.5 Crack Propagation**

Various suggestions as to why cracks are present and can therefore propagate in dental ceramics have been explored. The presence of micro-cracks within the ceramic can be accepted given the number of mechanisms available for their production. These mechanisms include stresses from differing thermal expansion coefficients in multiphase materials, thermal expansion anisotropy in non-cubic single-phase ceramics (e.g. alumina), thermal history and phase transformations (Kase et al. 1985). The thermal history during the laboratory firing cycles used to produce dental ceramic restorations can for example alter the leucite content of a dental ceramic, which has a higher coefficient of thermal expansion than the usual surrounding feldspar glasses ( $7 \times 10^{-6}/^{\circ}\text{C}$ ). Slow cooling of the restoration increases the leucite content, producing a stressed state, (Mackert et al. 1991)

Other mechanisms include the insults from the grinding that occur during the manufacturing of dental ceramic restorations (Kase et al. 1985). Grinding with diamond or silicon carbide burs is used for shaping after firing.

### **2.5.1 The Effects of Moisture.**

There are several possible ways that moisture can affect the strength of ceramics. Listed below are several ways suggested by Bascom (1974), Gupta et al., (1981) and repeated by Drummond (1991). The most widely recognized concept is that of water molecules and strained bonds at the crack tip in glass. See figure 2.19.

#### **2.5.1.1. Possible effects of Moisture**

1. Surface corrosion leading to flaws
2. Condensing into a crack tip, exerting considerable capillary pressure, acting to open the flaw.
3. Embrittling the material around the crack tip.
4. Reducing the surface energy needed to form the new surfaces.
5. Lowering the activation energy associated with slow crack growth.

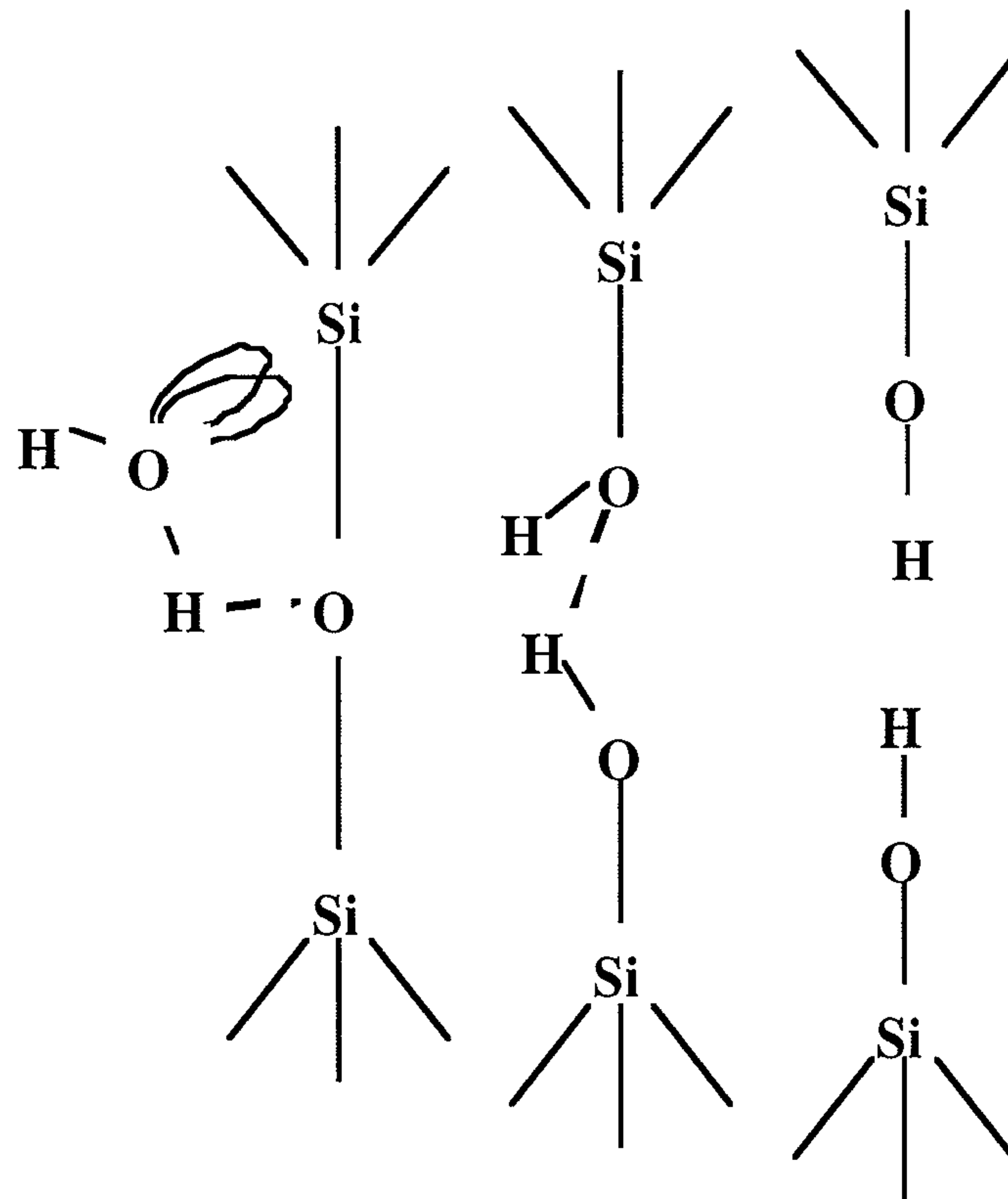


Figure 2.19. Diagram depicting the interaction of water on a strained silicon dioxide bond, causing it to separate. Adapted from Lawn 1983.

#### 2.5.1.2. The effect of moisture levels and pH on crack velocity

The following schematic diagrams of crack velocity – fracture toughness curves serve to illustrate how different glass formulations have altered crack velocities given varying conditions such as water concentration and pH. The oral environment often has changeable degrees of water concentration over ceramic restorations. For example with the mouth closed the ceramic will be completely covered by water based saliva, but during opening and more particularly, mouth breathing, the air passing through the mouth will range from ambient humidity as it enters the mouth, to 100 percent humidity as it leaves. The pH of the mouth is influenced by what the person ingests ranging from acidic carbonated drinks to basic antacids. A person may be suffering from gastric hyperacidity and reflux, which causes hydrochloric acid to enter the mouth and lower the pH to approximately 1. Increased plaque formation is associated with lowered pH.

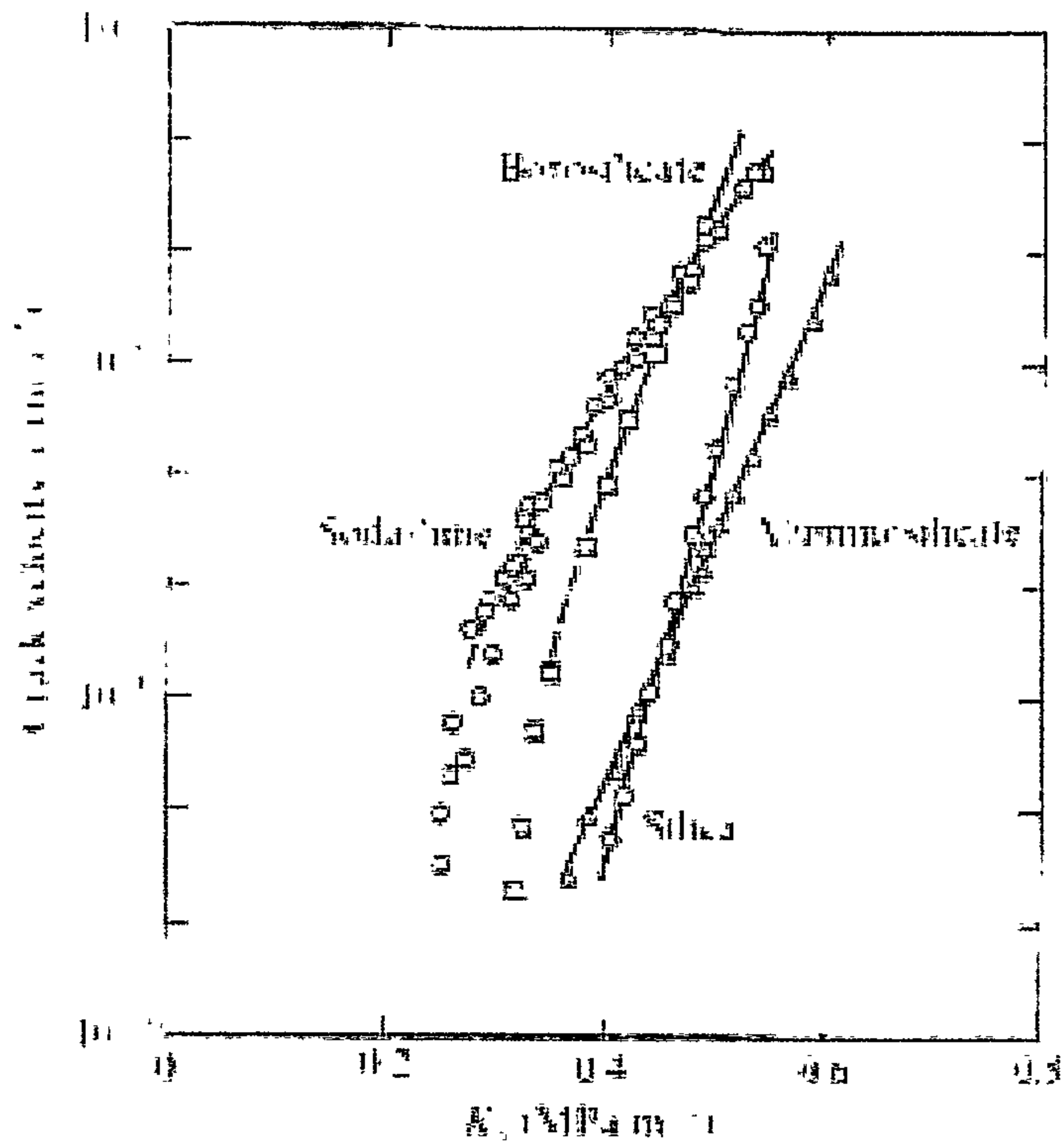


Figure 2.20 Crack velocity curves for different silicate glasses in water (Adapted from Lawn 1993)

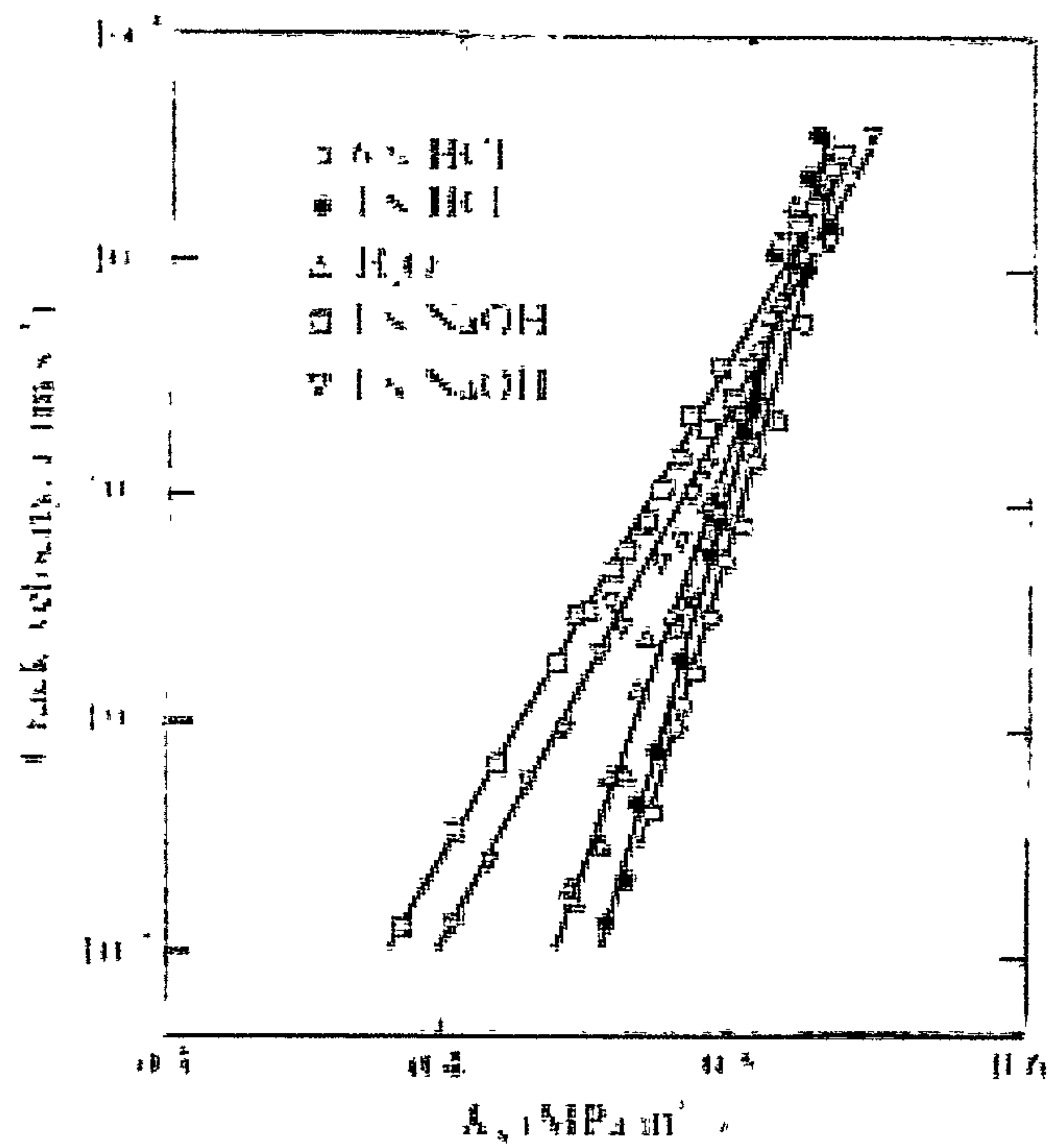


Figure 2.21 Crack velocity curves for fused silica glass in acidic and basic solutions (Adapted from Lawn 1993)

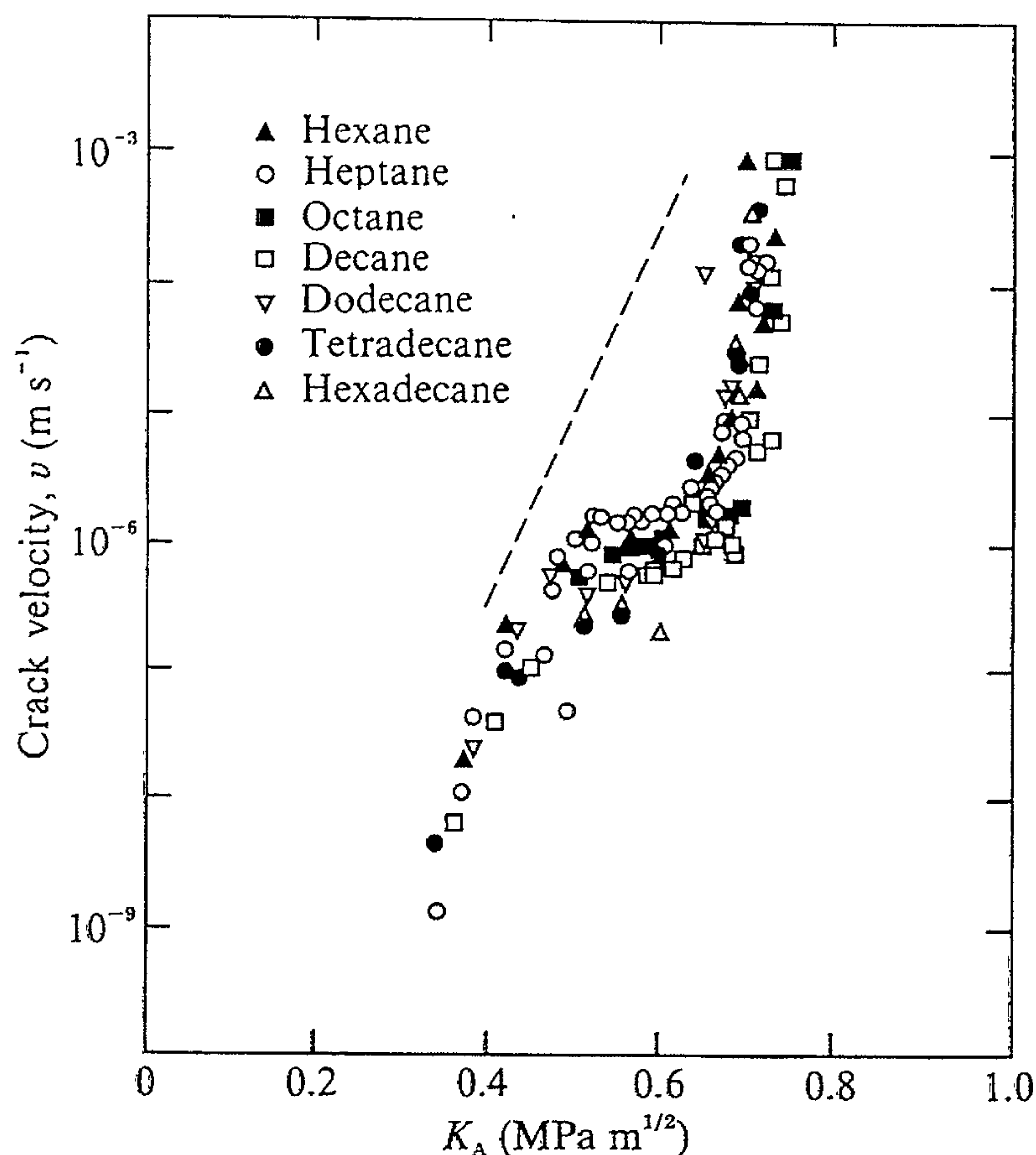


Figure 2.22 Crack velocity curves for soda lime glass in straight chain alkanes. The dashed curve is for water and is included for comparison.(Adapted from Lawn 1993)

Figure 2.20 demonstrates a broad range of glass formulations that includes those found in Dentistry for both dental composite resins and dental ceramics and how the presence of water affects velocity and its relationship to toughness. The curves in figure 2.21 demonstrate how changes in pH also affect velocity and how the rate is dependent on fracture toughness. Gehrke et al., (1991) confirm the pH role and point out that the diffusion rate of the ions for crack tip corrosion is load dependent. This is for silica glasses, which are the basic matrix for most dental ceramics used for veneering both all-ceramic and metal-ceramic crowns. The final figure in this series, fig. 2.22 is included to exemplify the diverse range of materials that can have an effect on the velocity of cracks in silicate glasses and that perhaps the ionizing effect of water is quite significant.

### 2.5.2. Crack Propagation - Velocity.

The crack has two aspects, which give it velocity. The first is where the inertia of the opposing sides gives the crack instability as the sides move apart, and the second comes

from a sudden impact. Dentally the instance of sudden impact and the response from the surrounding impact are important to this study. This is the type of impact where an opposing tooth cusp might impact the ceramic. It has been reported that terminal velocities of cracks in some brittle materials can reach significant fractions of the stress-wave velocity, (Lawn 1993).

The other significant aspect to stress-waves, which travel at near sonic velocities, is that they are reflected by the boundary of the system and can therefore complicate the resultant crack pattern. The suppression of stress waves and crack propagation is therefore an important aspect for ceramic restoration reliability. Presumably boundaries formed between different phases of dental ceramics and the flaws will further complicate the direction of crack propagation. In some instances these flaws could even act to deflect or arrest crack progress.

The suppression of this action by lowering the temperature past freezing point or using heat to drive off the water can be achieved in a number of applications except in the mouth. The temperature of the mouth remains at 37° Celsius for obvious biological reasons and so the presence of moisture is an integral part in allowing crack propagation to proceed.

Mencík (1996) describes the velocity for the propagation of cracks in ceramics as having three phases. The first two are dependent on chemical and temperature as well as stress but the final phase is dependent on stress alone. Figure 2.23 below is adapted from his book.

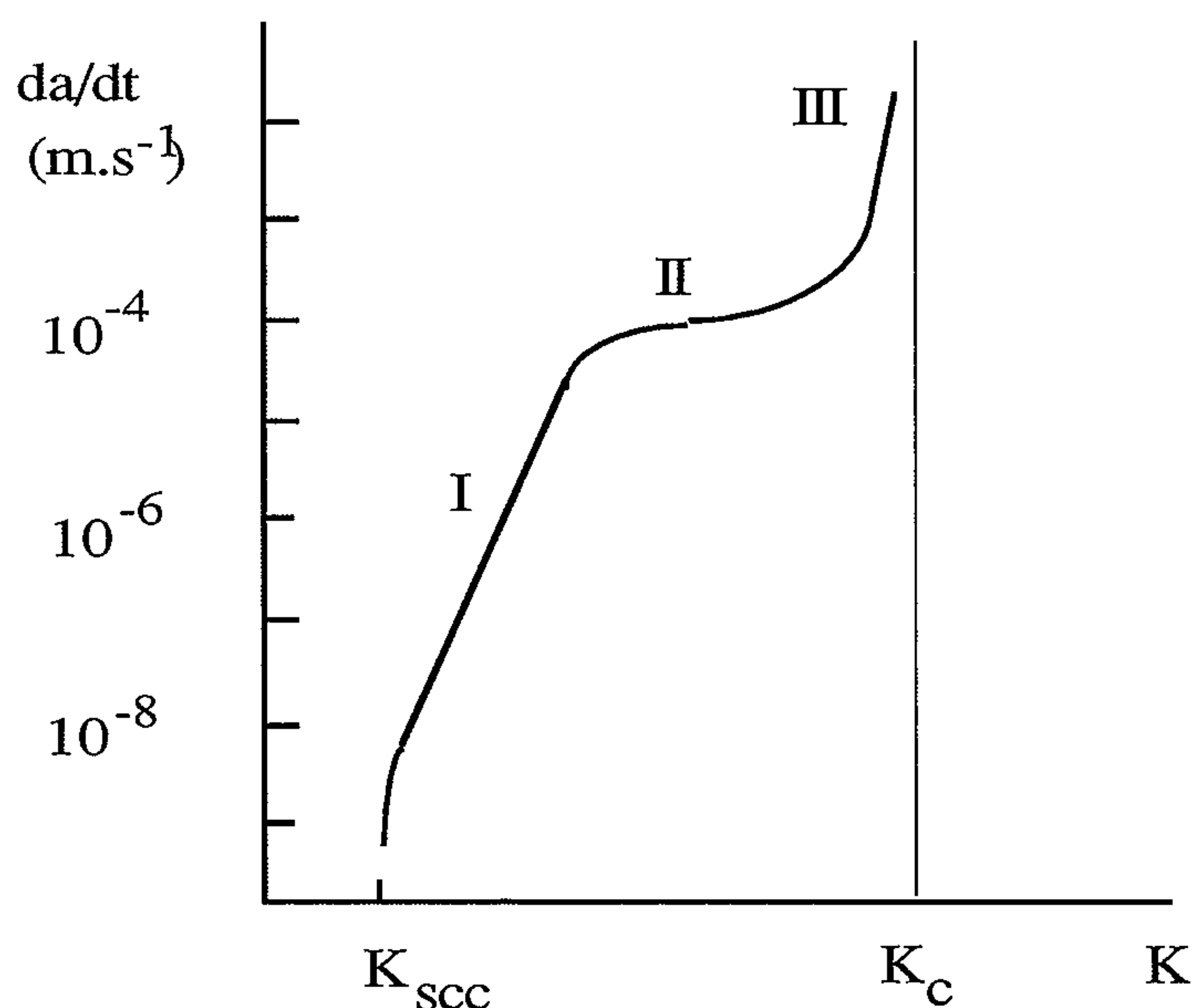


Figure 2.23. Velocity of Crack Propagation in brittle materials as a function of the stress intensity factor. (Adapted from Mencík 1996).

The three regions in the curve represent three specific stress/velocity relationships. In region I at low applied stresses, velocity is highly dependent on  $G$ ; region II, at intermediate stresses, the velocity is tending to plateau due to atmospheric environments; region III, at high stresses, the velocity is even more sensitive to  $G$  than in region I. ( $G$  is the Mechanical-energy-release rate). Llorca (1991) points out that when the cohesive forces can no longer equilibrate the stress intensity factor, the crack tip will propagate a definitive length and new bridging forces will be produced behind it.

### 2.5.3. Crack Propagation to an Interface.

The work of Davis et al., (1991) considers not only the fracture energy of interfaces, but explores how the crack gets to the interface using an elastic indentation technique. This article describes how the use of a spherical indenter in the elastic range, which is analogous to Hertzian fracture in homogeneous brittle solids can be used to relate the fracture energy to the applied load. The implication for this to dental ceramic bilayers is that the energy release rate, and total energy released as the crack extends to the interface will have a bearing on the available energy for delamination should the cone crack reach it. This work reveals that for a relatively homogenous material with cracks in the constant cone crack range, with a constant angle defined by  $\beta$ ,

$$E\Gamma_c R^3 / P^2 = \alpha(\nu) \cos \beta(\nu) \quad (2.24)$$

where  $P$  is the load,  $R$  is the cone crack radius,  $E$  is Young's modulus,  $\Gamma_c$  is the fracture energy,  $\nu$  is Poisson's ratio and  $\alpha$  is a coefficient. Experimental results for glass with  $\nu$  approximately 0.25, coupled with stress analysis have suggested that  $\alpha$  is about  $2.2 \times 10^{-3}$ .

An earlier approach to this was made by Dong et al., (1989) where polarized light studies indicated that the stresses were significantly higher at the point of contact. They concluded that failure of the interface was controlled by the initiation and propagation of a crack, from the contact point rather than a spontaneous overload failure of the entire bonded interface.

### 2.5.4. Crack Propagation at an Interface.

The other possibility that exists for this crown system and which was also observed in the clinical failures, was crack propagation along the interface between the In-Ceram® core material and the Vitadur Alpha® veneering porcelain. This was demonstrated in figure 1.4 of chapter one. The two materials have quite different mechanical properties but the assumption for the success of the system as a restorative is that they are strongly bonded together. This being the case, the strains at the interface depend on only two parameters,  $\alpha$  and  $\beta$  which are known as the plane strain Dundurs parameters (Mencík, 1996). As these

parameters require knowledge of the relative Poisson's ratios and shear moduli, then they can be written as follows:

$$\alpha = \frac{\frac{G_1}{G_2}(1-\nu_2) - (1-\nu_1)}{\frac{G_1}{G_2}(1-\nu_2) + (1-\nu_1)} \quad (2.25) \quad \beta = \frac{1}{2} \frac{\frac{G_1}{G_2}(1-2\nu_2) - (1-2\nu_1)}{\frac{G_1}{G_2}(1-\nu_2) + (1-\nu_1)} \quad (2.26)$$

Where  $G_1$  and  $G_2$ , and  $\nu_1$  and  $\nu_2$ , are the ratio of shear moduli and Poisson's ratios of the first and second materials.

Another important consideration is how the elastic moduli of the two materials combine to give an overall effective Elastic modulus,  $E_e$ . This is defined as:

$$\frac{1}{E_e} = \frac{1}{2} \left( \frac{1}{E_1^+} + \frac{1}{E_2^+} \right) \quad (2.27)$$

Where, for plane strain

$$E^+ = \frac{E}{(1-\nu^2)} \quad (2.28)$$

The general layout of this crack system is seen in figure 2.24 (Mencík, 1996)

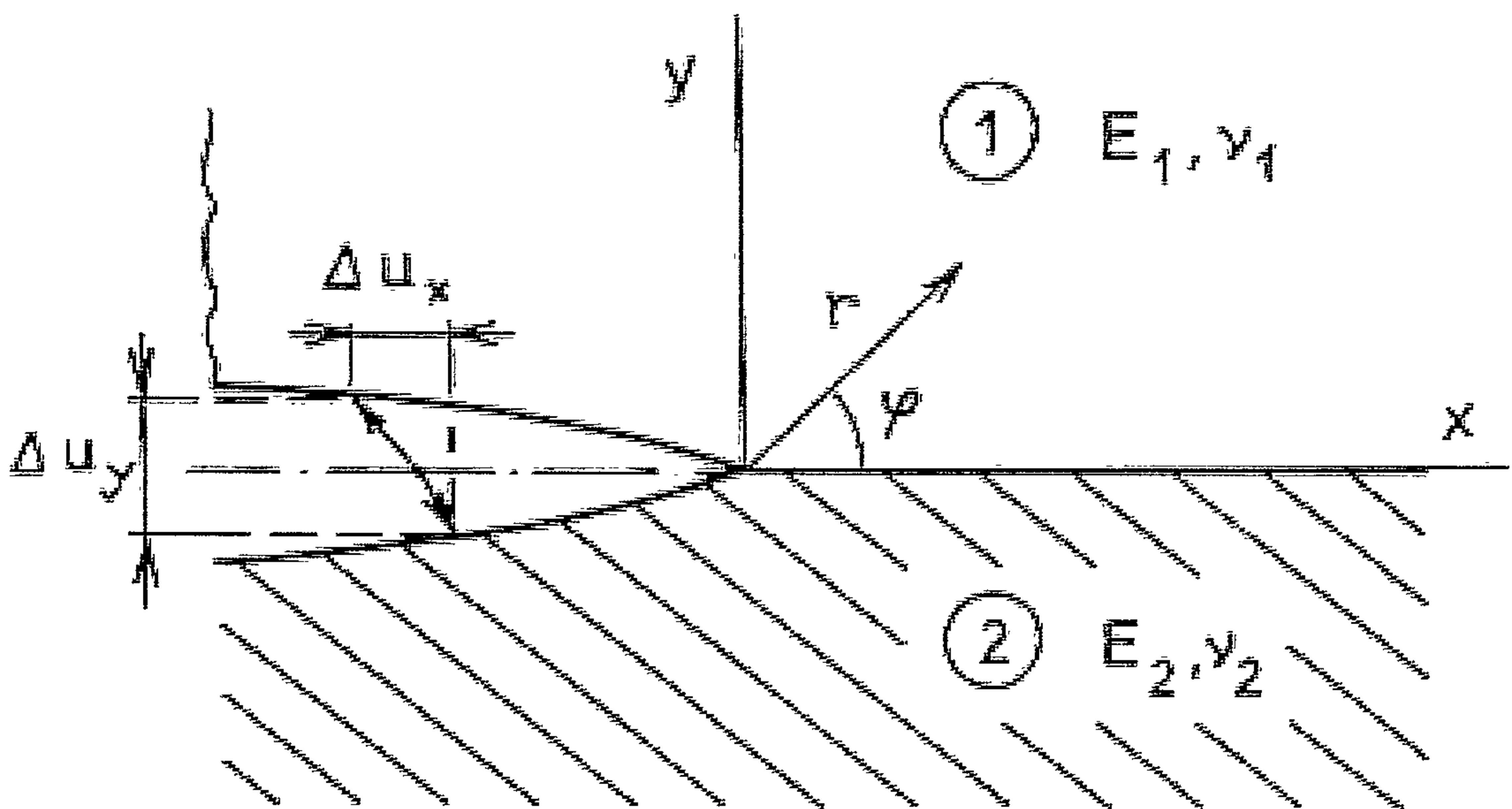


Figure 2.24 A crack at the interface of two materials where the subscripts in the Dundurs parameters and the following formulae refer to material 1 in the top layer and material 2 in the bottom layer.

At the crack tip region (Fig 2.21) there is a plane strain. As the crack progresses the

different energy release rates from the different sides cause an oscillation of the resultant stress vector. This oscillation is called  $\varepsilon$  or the oscillation index (Mencík, 1996) and is written as:

$$\varepsilon = \frac{1}{2\pi} \ln \frac{G_1 + G_2(3 - 4\nu_1)}{G_2 + G_1(3 - 4\nu_2)} \quad (2.29)$$

The stress in the crack plane in the vicinity of the tip can be expressed in the complex form in the following way:

$$\sigma = K' \frac{r^{i\varepsilon}}{\sqrt{2\pi r}} \quad (2.30)$$

where

$$\sigma = (\sigma_y + i\tau_{xy})_{\varphi=0} \quad (2.31)$$

is the vector of normal and shear stress, ( $i$  is the complex number  $\sqrt{-1}$ ) and

$$K' = K_1 + iK_2 \quad (2.31)$$

is the complex stress intensity factor. The form is different from the homogenous material and is related to the different strain releases that will result as the two layers come apart and form their own new surfaces with different areas. This leads to the shear factor. The relative plane strain displacements of two points on the top and bottom crack surfaces,  $\Delta u_x$  and  $\Delta u_y$ , can then be expressed in terms of a complex stress intensity factor,  $K'$ , as:

$$\Delta u = \Delta u_y + i\Delta u_x = \frac{8\sqrt{1-\beta^2}}{1+2i\varepsilon} \frac{K_1 + iK_2}{E_e} \sqrt{\frac{r}{2\pi}} r^{iy} \quad (2.32)$$

with  $r$  being the distance from the crack tip. The component  $\Delta u_y = u_y(r, \varphi = \pi) - u_y(r, \varphi = -\pi)$  is the real part of the right hand side, the component  $\Delta u_x = u_x(r, \varphi = \pi) - u_x(r, \varphi = -\pi)$  is the imaginary part. Mencík (1996) points out that the crack propagation problem is a mixed one of the first and second modes of crack opening and it is impossible to separate the two. However for interfaces that are significantly less tough than the neighbouring bulk materials, a crack can continue to follow an interface, even though the stress intensity

factor at the crack tip is not purely Mode I and may have substantial Mode II component (Evans and Hutchinson, 1989).

The energy release rate during crack growth along an interface is:

$$G = \frac{1 - \beta^2}{E_e} |K'|^2 \quad (2.34)$$

where

$$|K'|^2 = \sqrt{K_1^2 + K_2^2} \quad (2.35)$$

is the modulus of the stress intensity factor.

### 2.5.5. Cyclic Fatigue.

The concept of cyclic fatigue when applied to the oral environment is often complicated by the changing nature of the contact surfaces, based on the assumption that the fatigue is caused by contact between opposing elements, (i.e. teeth or restored teeth). In the young dentition, the teeth usually have well defined cusps with rounded tips (usually of the order of 2 millimetres in diameter). With use, the tips tend to become flattened and any irregularities, or asperities will be subjected to lower loads. This has been reported by DeLong et al., (1989)

Where ceramics are concerned, aspects of micro-structural shielding against crack propagation, such as friction between irregular faces of cracks in highly crystalline materials and crack bridging from elongated particles may begin to break down. Simple wear and fracture of these particles as they rub or stretch, would eventually break down their resistance. The environmental aspects of moisture, pH and the number and magnitude of the load cycles would add to the degeneration. Fracture toughness and hardness of the ceramic will affect their wear under these conditions. In general, materials with an optimum combination of high toughness and high hardness will be the most wear resistant (Evans and Marshall 1980).

## 2.6 Strengthening and Toughening Philosophies.

### 2.6.1 Strengthening Philosophies.

There have been several reports for strengthening glassy materials. One way is to remove surface defects by chemical etching or fire polishing and the other is to develop compressive stresses to the glass surface (Doremus 1973). This compressive technique has been used for toughening automotive windscreens where compressive stresses of 10

to 50 MPa may develop at the surface. There is always an opposite stress to balance the system and the windscreens have compensating internal tensile stresses. The introduction of a crack released the stress antagonists and the windscreens shattered. This second group includes systems such as thermal tempering described by Anusavice et al. (1991). Doremus (1973) also includes thermal stressing, ion exchange and surface crystallization as methods for increasing compressive stress in the surface of the glass and therefore adding to the normal fracture stress. Thermal history of the porcelain during repeated firing cycles can introduce tensions at interfacial boundaries, which may again be deleterious to the overall strength, (Fairhurst et al., 1989 Twiggs et al., 1989)

McLean et al., (1987) stated that because all dental ceramics tend to fail at the same critical strain of the order of 0.1%, any increase in strength, and also toughness, could only be achieved by an increase in the elastic modulus. This complies with his pursuit of Alumina as a strengthening material and subsequent recognition of In-Ceram as a substantial improvement in this area, (McLean 1991).

A more recent paper by Giordano et al. (1994) demonstrates the effects of ion exchange, however they point out the superficial nature of this procedure which is easily lost if the surface is refired or adjusted with grinding or polishing. Ion exchange has also been regarded as a toughening mechanism.

### **2.6.2 Toughening Philosophies.**

There are several toughening philosophies that have been investigated for hard and brittle materials, i.e. ceramics. The article by Swain (1991) summarizes these toughening mechanisms as:

- Crack deflection/ pinning;
- martensitic volume dilating transformation;
- stress-induced micro-cracking;
- crack-tip bridging due to ligament formation, fibre reinforcement or metallic particle reinforcement.

In general, the only toughening systems used in dental ceramics have been systems less likely to change the ceramic translucency requirements of the dental ceramics (Southan 1987a and b). Systems such as ion exchange have been suggested in the past (Southan 1970) and are still being explored (Dunn et al. 1977, Jones et al., 1983, Jones et al., 1988a, Seghi et al., 1990, Anusavice et al. 1992, Denry et al. 1993). Dunn et al., (1977) also point out that the process of ion exchange is time and temperature dependent, with stress relaxation occurring at higher temperatures and that certain glass formulations lend themselves more to this type of strengthening mechanism than others.

Crack tip bridging mechanisms offer increased reliability against crack propagation. The courser-grained aluminas have been shown to display this property (Hu et al., 1991). The structure of the In-Ceram core includes elongated alumina crystals which could act as crack bridging agents to increase toughness.

Most dental porcelains do undergo some degree of phase transformation toughening due to their leucite and feldspar contents. These changes occur principally during the firing processes (Barreiro et al., 1989). From Becher and Swain (1992) we learn that the formation of a transformation zone acts to shield the crack tip from the applied stress intensity,  $K_I$ .

$$K_I + \Delta K = K_0 \quad (2.36)$$

where  $K_0$  is the crack tip stress intensity required to advance the crack tip. When transformation toughening increases the fracture resistance (i.e.  $K_I > K_0$ ),  $\Delta K$  is negative: thus  $\Delta K$  equals  $-\Delta K^T$ . The crack tip propagates when  $K_I$  is raised sufficiently and the stress intensity at the crack tip equals  $K_0$ .

Thermal tempering has also been reported in a paper by Anusavice et al., (1991) and Hojjatie et al., (1993), suggesting that it inhibits crack formation but does not retard crack growth. This is in concert with Lawn et al., (1977) and Marshall et al., (1978b), where it was pointed out that compressive stresses at the surface inhibit initial crack growth, but the counterbalancing forces in the inner regions act to enhance crack growth. The tempering stresses in feldspathic porcelain were measured by Dehoff and Anusavice (1989), who found significant surface compressive stresses. The viscosities of three experimental body porcelains and three experimental opaque porcelains as a function of temperature were calculated from creep rates measured in a bending beam viscometer. Measurements were made under thermal equilibrium conditions for temperatures ranging from 550° to 625°C. Thermal expansion data measured in a differential dilatometer at slow heating rates were supplied by the manufacturer. Temperature distribution in the plates as a function of convective heat transfer coefficient, initial plate temperature, and plate thickness was calculated by use of standard numerical techniques.

Their calculations of transient and residual stress were performed for one body porcelain, for two plate thicknesses, and for three variable cooling rates. Calculated surface residual stresses were strongly dependent on plate thickness, cooling rate, and initial soak temperature. For the cases studied, the maximum residual surface compressive stress was 26.4 MPa. Furthermore, Dehof et al., (1992) found that calculated values of residual compressive stress for tempered specimens were considerably higher than those for specimens that were slowly cooled and those that were cooled by free convection.

Strengthening and toughening regimes as earlier commented, often cause loss of aesthetics due to increased opacity, (Brukl et al., 1987, Southan 1987)

Kase et al., (1985) listed some of the factors affecting residual stresses in dental ceramic systems due to the manufacturing processes. They include:

- Thermal expansion
- $T_g$ , the glass transition temperature.
- Trapped residual stress (strain)
- Temperature dependence of the elastic modulus
- Crystallinity
- Thermal conductivity
- Heat capacity
- Number of layered components
- Relative layer thicknesses
- Ability to relieve stress

### **2.6.3 Ceramic Structure.**

Alteration of the ceramic structure to enhance toughness has been recognized in industry as a major mechanism to improve toughness and strength. In terms of dental ceramic systems, several attempts have been made to use different ceramic systems with different compositions to overcome the low fracture toughness and strength properties of dental ceramics. These different ceramic systems have been described above in sections 2.2.1 to 2.2.4. Lampe et al. (1996) demonstrated significant differences for fracture loads between dental ceramic systems that had dissimilar structures.

### **2.7 The effect of adhesive luting systems.**

The use of all-ceramic systems for the restoration of teeth often involves an adhesive luting agent. These agents are usually derivatives of dental composite resin restorative systems. They are often used in conjunction with acid etched and /or "silane" surface treatments for the ceramic and etching techniques to increase adhesion to the tooth. This technique is subject to contamination, however, where used with a strict clinical protocol good adhesion to the ceramic can be achieved. In the short term, heating the silane treated ceramic at 100° C for 30 minutes can increase the bond strength by up to 33 percent, (Roulet et al., 1991)

There have been some reports that resin bonding can increase the fracture resistance of ceramic inlays (Dérand 1991). A more recent article by Lampe et al., (1996) demonstrated

a significant influence from luting systems when fracture loads for five different all-ceramic crown types was measured.

## 2.8 Mechanical Testing.

There is considerable discussion about the type of mechanical testing systems that should be applied to ceramics. In a publication, (Anusavice K.J. 1988), it was stated that the axial or tangential tensile stresses are more likely to contribute to crack initiation or propagation under intra-oral loads. Marquis (1985) pointed out that the likely cause of failure for porcelain-jacket-crowns (an all-ceramic restoration) was tensile force at the inner fitting surface, particularly at the lingual and gingival margins. A more recent article, Geiger 1990, reviews the trends from mechanical testing of advanced ceramic systems and feels that tensile testing is becoming the more favoured method. He quotes from K. C. Liu of the Metals and Ceramics Division, Oak Ridge National Laboratory that "ceramics must be tested in the tensile mode so that a well-established database can be produced." There are a number of methods available for testing ceramics in tension. They include:

1. Three Point Bend test, using a beam supported at both ends and loaded at a single point in the centre.
2. A Four Point Bend test, again using a beam supported at both ends but loaded at two positions near the centre.
3. A Bi-axial test method. This method has several variants, but it essentially uses a disc that is supported underneath, either by a continuous ring or three points, and loaded in the centre by a piston or a sphere.

The suitability of the three point bend test using a simple beam was assessed by Edwards et al. (1983) who regarded it as a simple test, but subject to surface conditions; however an earlier paper by Southan et al. (1973) had already proposed the bi-axial method as a better alternative to the three point bend test.

Another approach entirely is to consider fatigue. Rather than simply load a material until it fails, there is also a need to understand why it can fail following repeated loads that are much lower than the maximum failure load found by one of the above tests. McCabe et al., (1990) described two ways that can be used to look at fatigue. One is to test a material at various stress levels and see how many cycles that it survives, or determine the level of stress that a material can withstand given a preset number of cycles.

Finally, indentation fracture has been used extensively for the testing of brittle solids. There are four basic styles used: point load, sharp indenter, flat punch and sphere (Lawn and Wilshaw, 1975). A comment by Lawn et al., (1979) from their paper demonstrates this type of testing to be a simple way for the classification of brittle materials.

## 2.9 Analysis of the ceramic fractures.

### 2.9.1 Fractography

The analysis of fractured surfaces has often been accomplished with the use of a scanning electron microscope. This method is particularly useful for the glassy dental ceramics because of their inherent translucent nature. Kelly et al., (1989 and 1990) described three identifiable fracture surface regions which they described as smooth mirror region, the mist region and the hackle region. Each region represents a different energy states with respect to crack development. In the smooth mirror region, added energy from increase in strain and kinetic energy increases the velocity of the crack, to its maximum. Once this occurs the second region is entered and the increase in strain energy and kinetic energy create additional cracks at the crack tip. Then the last region, the hackle region is entered, where additional energy allows stable growth of secondary cracks. These are shown in figure 2.25.

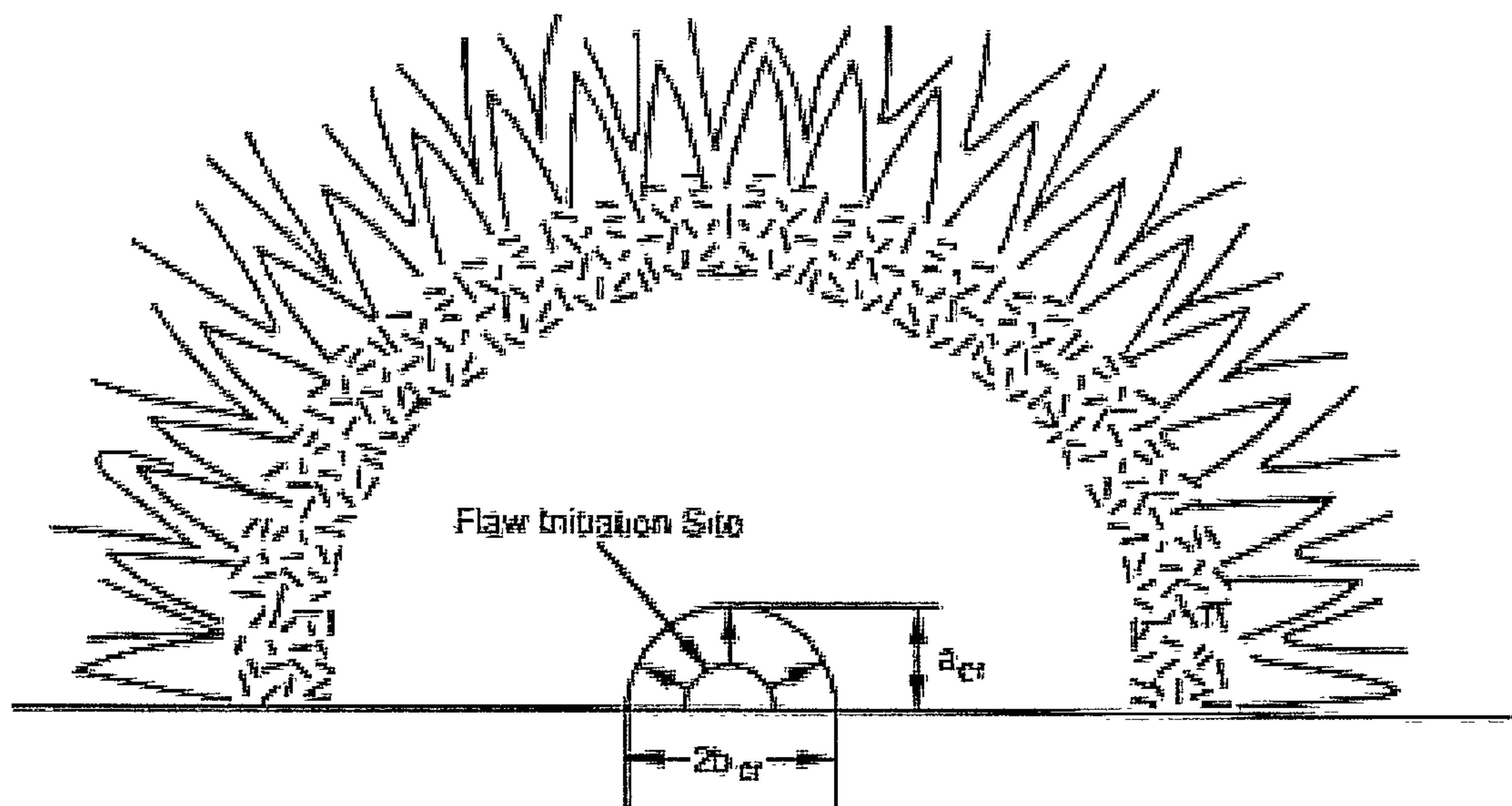


Figure 2.25 This schematic is attributable to Mecholsky (1995) and demonstrates the three regions described above.

### 2.9.2 Finite Element Analysis

Finite Element Analysis (FEA) has been used in dental research to evaluate stress distribution in ceramic restorations, (Anusavice KJ et al., 1988; Papavasiliou G, et al., 1996; Magne P et al., 1999a, b, c & d). The analysis of stress (magnitude and distribution) is useful for indicating the physical response of the system and the most likely place for fracture to occur. This analysis is not useful for indicating at what specific magnitude of applied load that fracture will occur. Such a study, amongst others, needs to include the probability of fracture i.e. Weibull modulus, (Kelly, 1995; Tesk JA et al., 1988; Bendeich P et al., 2001). However, a comparison of the fracture strength of each material with the tensile stresses that exist in each corresponding material will allow a contrast to be made amongst them, indicating when failure is most likely to occur.

The ability of FEA to provide an accurate method to approximate a dental prosthesis is closely coupled with the manner in which the FE model is constructed. This includes caution with the type of finite elements used (2D plane-stress, 2D plane-strain or axisymmetric, 3D brick elements), the particular way that the mesh is constructed and the way that the system is modelled in terms of constraints and loads (Tesk et al., 1988; Koriath et al., 1997; Ridlon et al., 1987).

The relative merits of using the different types of finite elements available are discussed by Koriath and Versluis (1997). The method of mesh construction can vastly affect the accuracy of the FEA solution (Cook 1974; G+D Computing 1999; Ridlon et al., 1987). Generally, the finer the mesh (meaning more elements for a given model), the more accurate the solution of the FEA on the model. As the finite element mesh is refined, results converge towards the exact solution (Ridlon et al., 1987).

In particular, the construction of a fine mesh of elements in regions exhibiting steepest stress slopes (where stresses are rapidly changing), or stress concentrations, is vital (G+D Computing 1999; Ridlon et al., 1987). Such regions are likely to include rapidly changing boundaries (i.e. notches, sharp curves etc) or interfaces between different properties. The generation of transitions from coarse to fine meshes is an important aspect of FEA mesh construction (G+D Computing 1999; Ridlon et al., 1987).

Finite elements also perform best when their aspect ratio, the ratio of the longest edge to the shortest edge, is one (G+D Computing 1999; Ridlon et al., 1987). Thus, for four noded plate elements the preference is to create elements which depict a square, and for three noded plate elements, equilateral triangles (Cook 1974). The model can be improved by choosing higher order elements (Ridlon et al., 1987). The preference exists for the use of four noded elements in contrast to three noded elements.

Accuracy in modelling uniformly distributed loads along an element comes about with an FEA program that equates this load to the nodes of the element in a 'consistent' rather than 'lumped' fashion (G+D Computing 1999; Ridlon et al., 1987). This is where the conversion takes into account bending in each element that results from the uniformly distributed load.

In the post-processing stage, when viewing the resultant stress contours across the model, it is important to have the stresses at nodes averaged only for those elements of the same property type. This allows the true discontinuous contours, where elements of different properties come together, to be generated (G+D Computing 1999).

Paying particular attention to these details, it is obvious that the method of mesh construction plays a key role in obtaining accurate and therefore useful results in the analyses of dental prostheses. It is with these points in mind that this research has been conducted. In areas of high tensile stress, the probability of fracture is greatest. Therefore it is vital to analyse those areas of high tensile stresses and investigate the influences on these peak tensile stresses.

### **2.9.3 Analytical Modelling of Mechanical Contacts.**

The development of an analytical computer program for modelling mechanical contact, (N. Schwarzer et al., 1999a & b; Chudoba et al., 1999; Richter et al., 2000,) crack (Schwarzer et al., 1998b; Schwarzer et al., 2000a) and other deformation problems of inhomogeneous and finite bodies (Jamting et al., 1997; Schwarzer et al., 1998a; Schwarzer et al., 2000b) has enabled the investigation of stresses associated with spherical indented contact on flat elastic surfaces. The program called "Elastica", is described as providing a completely analytical model for load problems on layered materials, (Schwarzer 2000 a & b) and has been confirmed by finite element calculation, (Faulkner et al., 1997), and real nanoindenter experiments, (Chudoba et al., 1999; Chudoba et al., 2000a & b; Wändstrand et al., 2001). It has been further developed to solve contact problems in the vicinity of edges, such as can be found at the periphery of a spherical indenter contact (Schwarzer, 2001a & b)

### **2.9.4 Weibull Statistics**

Weibull statistics will be used to estimate the failure predictability of the crown designs and the various ceramic models used to create them. These statistics are useful as a reliability predictor for objects containing multiaxial stresses. They have been found to provide conservative results which is well suited to designs for clinical application, (Petrovic et al., 1981).

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# CHAPTER 3

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## In vitro Indentation Testing of Anterior All-Ceramic Crowns

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### 3.1 Introduction

In this chapter a group of anterior crowns were constructed and tested to destruction. The crowns were made from the In-Ceram® and Vitadur Alpha® materials, producing typical bilaminar porcelain jacket crowns. These crowns were made as uniformly as possible, however the manual nature of the construction technique lead to considerable variations in specific thickness of the core layer in particular. The testing were made by loading the palatal surface with a spherical indenter. The direction of the load were at approximate right angles to the surface, although it is accepted that clinical contacts may be sliding along the opposing tooth surface in less than ideal occlusal relationships. The crown itself was cemented to a brass die using a proprietary glass ionomer cement. The loads for the approximate initial fracture and final fracture were recorded along with the actual thicknesses of the crowns and their component layers. The fracture patterns were also recorded and compared to the thicknesses and loads.

The use of Weibull statistics as a means of relating the probability of failure to loading condition provides a predictor of reliability. In this manner it also is useful to help ensure some clinical longevity for these restorations. This is a theory suited to multiaxial stresses. Petrovic et al., (1981) found that the Weibull theory generally underestimates strengthening effects of the compressive principal stress, thus providing a conservative failure prediction. A requirement for patient satisfaction is that these restorations will have a considerable life-span as well as being aesthetic. The random presence of flaws in ceramic structures generally means that some will have early initial failure. This was pointed out in chapter one with a small number of early failures reported. The observation was that these failures occurred from an occlusal insult from a spherical like indentation that could be from an opposing tooth cusp. From Lawn (1993) it is learnt that numerous authors have endeavoured to quantify the basis of brittle solids failure using fracture mechanics, with a specific emphasis on Griffith flaws.

Lawn points out that strength reliability involves a probabilistic element and that a pattern emerges where there is a group of early failures, then as time progresses the failure rate settles down for an extended period of time and finally the rate increases again after

prolonged wear and tear. Figure 3.1 below is an adaptation of the “bathtub” curve from Lawn (1993). This demonstrates the early failures during or due to manufacturing events and early screening processes (e.g. trying in an uncemented crown), the period of prolonged use for those items which are manufactured correctly and finally the worn out failures.

### *Strength and reliability*

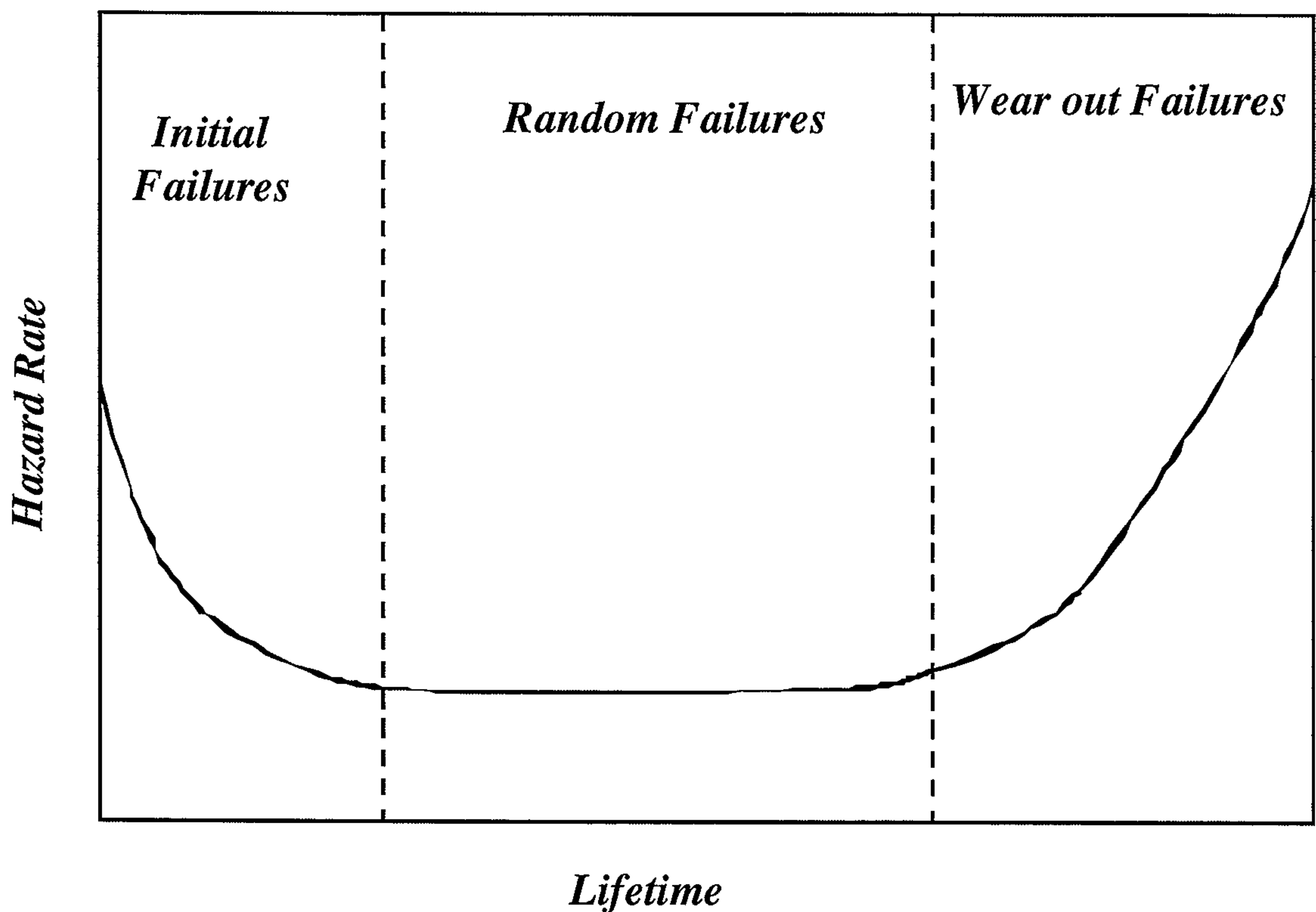


Figure 3.1 “Bathtub” curve, showing hazard rate against lifetime. The “mortality” rate is more probable during and immediately after “birth” (pre-existing defects), or after prolonged wear-and-tear in service, (accumulated defects). Adapted from Lawn (1993).

#### **3.1.1 Original Observations**

The results of the initial investigation of the material retrieved from early clinical failures are now to be reinvestigated in vitro in an effort to understand the original crack problem. One of the empirical observations was that the majority of failures occurred in regions where the core was thinner than recommended or the Vitadur Alpha veneer was thicker than recommended. An example appears in figure 3.2. The other observation was that the direction of loading and point of contact was in such an arrangement that the Vitadur Alpha was less supported than ideal. This chapter will deal with ideal loading on crowns constructed according to the manufacturer’s directions. The directions call for firing the

crowns at 960°C and then the furnace opens to allow them to cool to room temperature straight away. There is no slow cooling process nor is there a force or air tempering cooling process. The latter has been shown by Anusavice et al., (1989, 1991a & b) to inhibit growth of existing cracks.

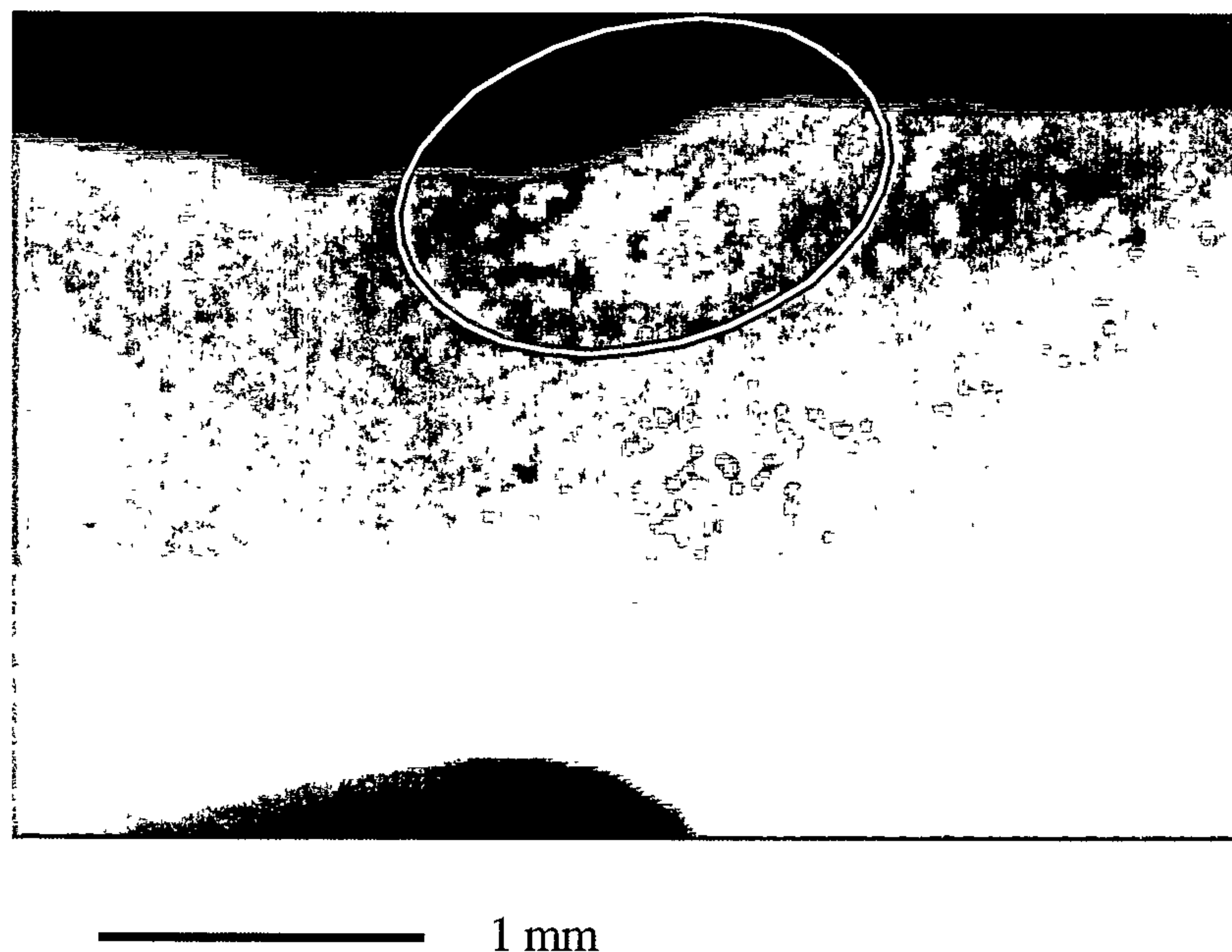


Figure 3.2. This photograph demonstrates the initial occlusal injury (circled) and the thin In-Ceram® core beneath a thicker Vitadur Alpha® veneer.

### 3.1.2 Nature of the Cracking Problem.

According to Mencík (1996), the main modes of failure for coatings or veneering porcelains, as are the case in this study, are delamination and cracking. Figure 3.3 contains a diagram from his work demonstrating this type of failure. Mencík also points out that there are marked differences in the response of the materials dependent upon the type of indenter used, principally the difference between a sharp and a blunt indenter. As this study is aimed at clinical problems, a spherical indenter is a reasonable method of choice because contacts from opposing teeth are often this shape. The load will be applied in a compressive mode, although it is recognised that the distribution of the resultant stresses will result in tensile and shear forces acting in other parts of the crown.

Recognition needs to be made of the complexity of the stress fields generated. This has a consequence on the resultant cracking pattern although for brittle materials crack extension in Mode 1 perpendicular to the principal tensile stresses, (sect. 2.4).

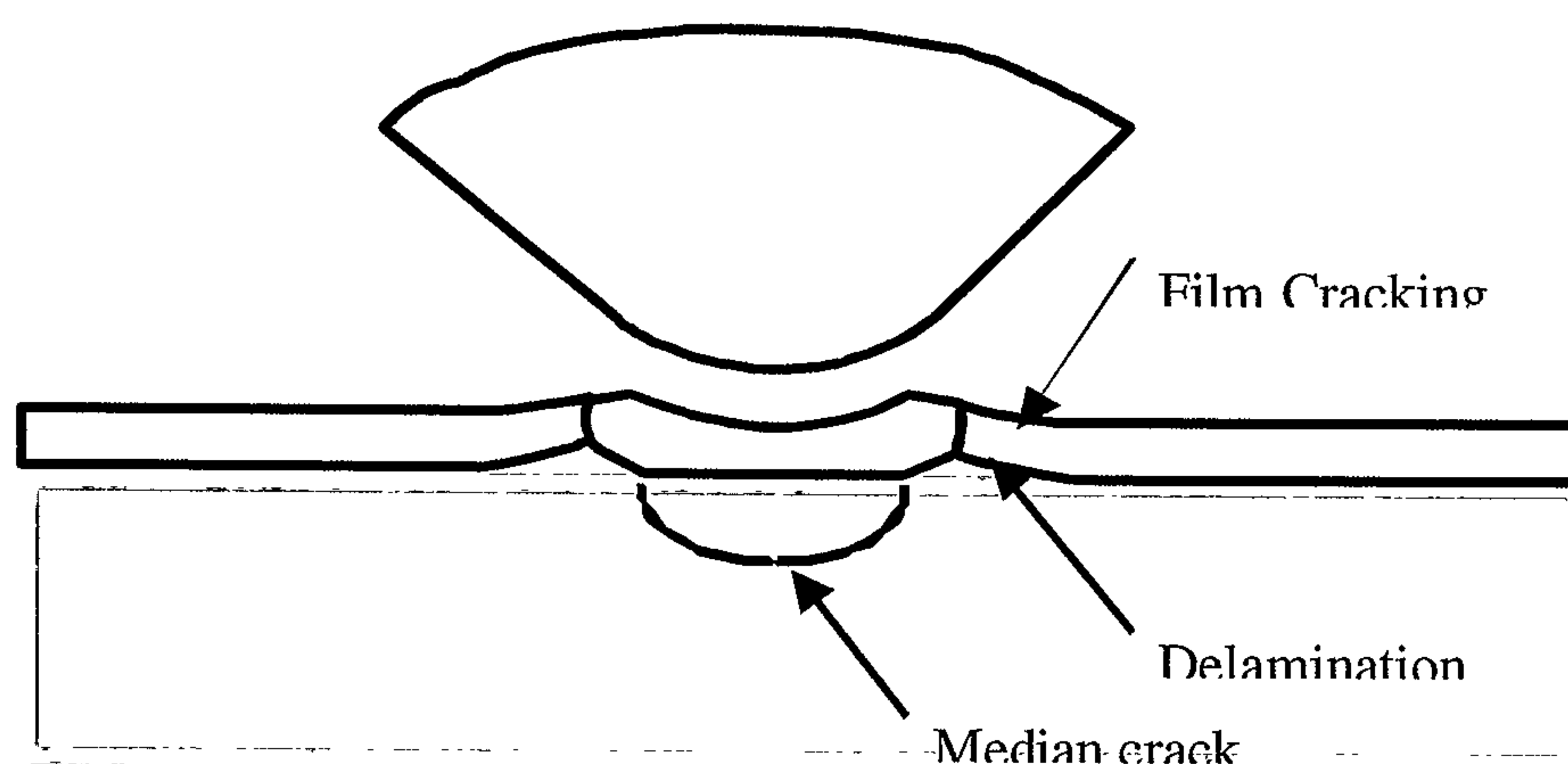


Figure 3.3 Diagram of fracture pattern that forms with a spherical indenter when applied to a hard coating on a hard substrate. From Mencík (1996).

### 3.1.3 Differences between Laminar and Monolithic Constructions.

The differences in mechanical properties between different components of a bilaminar construction presents problems not encountered in a monolithic layer. For example an IPS Empress® crown has essentially one layer with perhaps a thin glaze on the outside, whereas an In-Ceram® crown has two layers with very different mechanical properties. When an elastic monolithic material undergoes strain the material experiences a distribution of stresses according to its given set of mechanical properties. However in a bilayered composite material with differences in Poisson's ratio, Elastic modulus and tensile strength as examples, both layers undergo different levels of strain for a given load. For example if Poisson's ratio or the coefficient of lateral contraction is considered, which is defined as:

$$\nu = \epsilon_y / \epsilon_x \quad (3.1)$$

then the corresponding strains parallel to a load ( $\epsilon_x$ ) and the corresponding strains perpendicular to the load ( $\epsilon_y$ ) will be different where the two layers have different values for  $\nu$ . Mencík (1996) points out the more practical aspect of Poisson's ratio because while stretching a plate in one direction bending occurs in the perpendicular direction. Relating this problem to elastic modulus makes it easier to see the significance. If, for example, there is a modulus mismatch across an interface, then for a bilaminar material in flexure there is a continuity of strain across the interface, which, because of the modulus

difference results in a stress discontinuity. The concern where two layers are used in crown construction comes from the very small critical strain limit of porcelain which is given as 0.05 to 0.2%, Jones (1983). Clearly only a very small amount of dimensional change is required for fracture to occur.

The bond between these two layers then becomes extremely important. It can do one of three things. The bond can either:

1. Remain intact and transfer load from the loaded material to the supporting material, constraining the loaded material to resist fracture.
2. It can break causing a delamination between the two materials, possibly causing the loaded material to fail prematurely if it is the weaker of the two layers.
3. It can help transfer sufficient stress through the join to cause the supporting layer to develop enough tensile stress on the side furthest from the load and fail presumably from radial tension.

#### **3.1.4. Experimental Proposal.**

The experimental proposal was to investigate anterior in vitro crowns that are made from the In-Ceram® and Vitadur Alpha® materials. These were loaded until fracture starts for an extreme case scenario. The crowns were loaded in the middle of the palatal surface so the veneering porcelain has maximum support from the In-Ceram® core. The crowns were luted to a brass die, which is similar to a crown being luted to a cast gold core in a clinical situation. There are a number of potential failure mechanisms associated with such an in-vitro model. These include:

1. Crack initiation from the contact site because of the high local contact stresses resulting in cone crack formation followed by crushing of the porcelain below the contact.
2. Contact damage resulting in contact induced delamination of the layer as shown in figures 1.4 and 3.19a.
3. Crack initiation from the In-Ceram core at the inner surface because of flexure induced by the contact loading

This may help determine whether the cracking can start in the veneering layer and if it seems to proceed to radial cracks through the core. It is noted that a number of the clinical failures only failed through the veneering porcelain, usually where it was quite thick (>2 mm) or unsupported by the tooth preparation, such as over a proximal area.

3.2. Materials and Methods.

3.2.1 Ceramic crown preparation

The crowns used in this experiment were made according to the manufacturer's original instructions. The preparation design was based on Claus (1990) with minor modifications according to Ironside (1993). Rounded axio-gingival line angles are an essential feature to prevent stress concentrations at this point, (Abbate et al., 1989, Sorensen et al., 1990). Figure 3.4 below gives the outline of the restoration and the relative core design. The manufacturer's instructions call for the alumina slip to be laid down on the plaster die with a small paintbrush. This produces a core of approximate thickness, which can be adjusted after this first sinter fire. The thickness is usually recommended to be 0.5 mm but can be quite variable as there is little opportunity to gauge the depth of slip over the plaster die. The thickness of the core was measured at the indentation test point after the crowns had been fractured. These measurements are recorded in table 3.1.

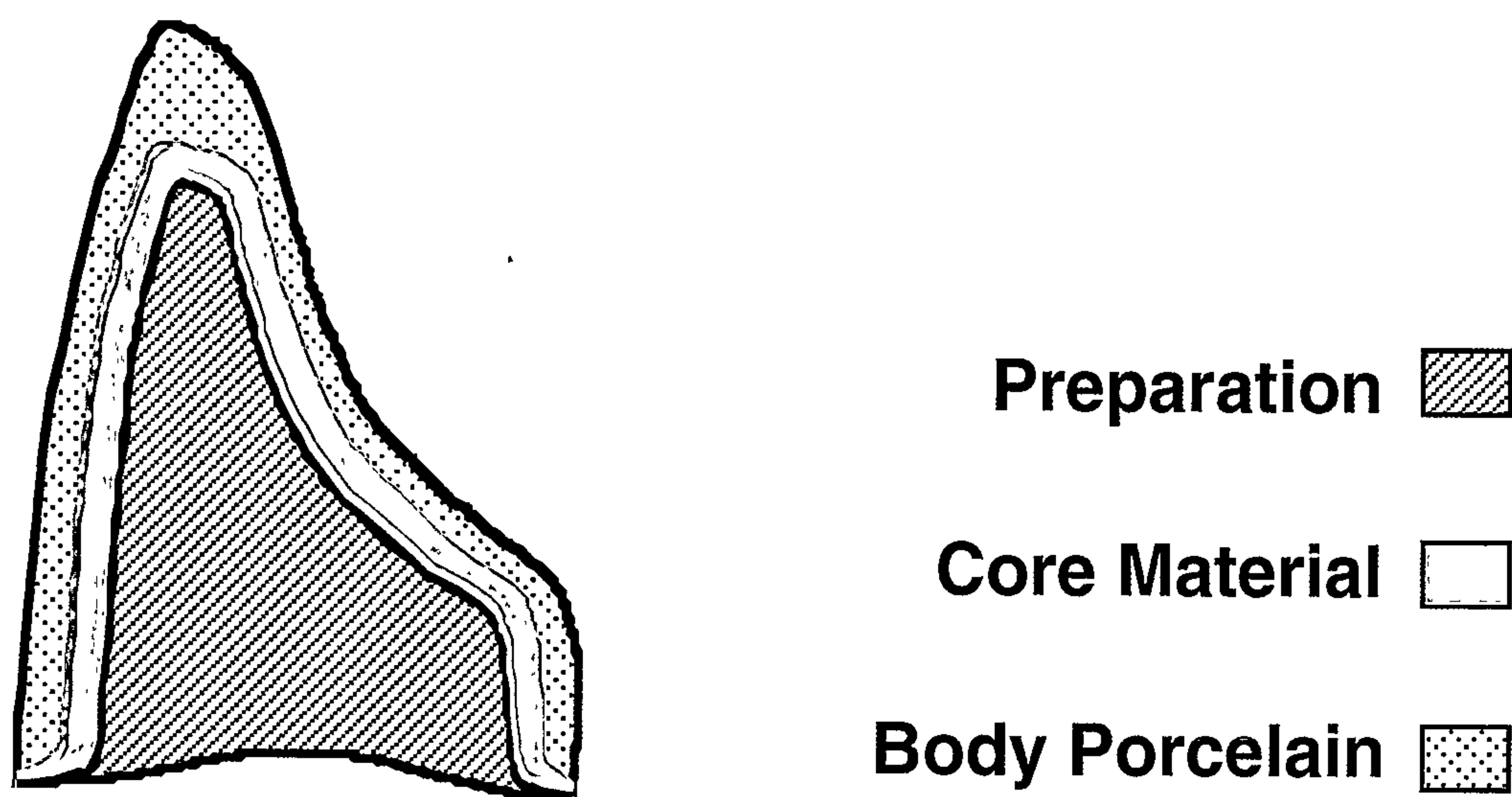


Figure 3.4. The standard design for an In-Ceram® core covered by the veneering porcelain, Vitadur Alpha®.

Each crown was made for a specific brass die. The dies were made from a master die. This master brass die was constructed from an ideal preparation cut on a Columbia dentoform central incisor according to Ironside (1993).

### 3.2.1.1. Construction of the brass dies.

The master preparation was made according to an ideal preparation for an upper central incisor, and described by Ironside (1993). The tooth was then removed from the dentoform and a base added, using Dicor spruing wax, to provide a cylindrical base. This allowed a uniform base of known diameter, which would provide support when the test dies were mounted in the testing apparatus.

Duplicates of the preparation were made using polyvinylsiloxane impression material. Wax patterns were constructed by slowly adding small amounts of wax to the impressions, allowing cooling and contraction to occur before the next increments were placed. These were left overnight to allow thorough cooling to occur.

The wax patterns were then individually invested in "Cristoballite" investment, and allowed to set. The wax was then burnt out and the investment rings left to heat soak at 620 °C for at least 40 minutes. The castings were then made using Degussa Phantom metal, a brass derivative technique metal used to simulate dental gold castings. Defective castings were discarded and a final set of 20 dies was produced.

The finished castings are designed to be mounted in a brass holder supported by a stage. The stage is adjustable in both horizontal axes to allow accurate positioning of the specimen below the indenter head. A photograph of the master brass die and a duplicate test die appear below in figure 3.5.

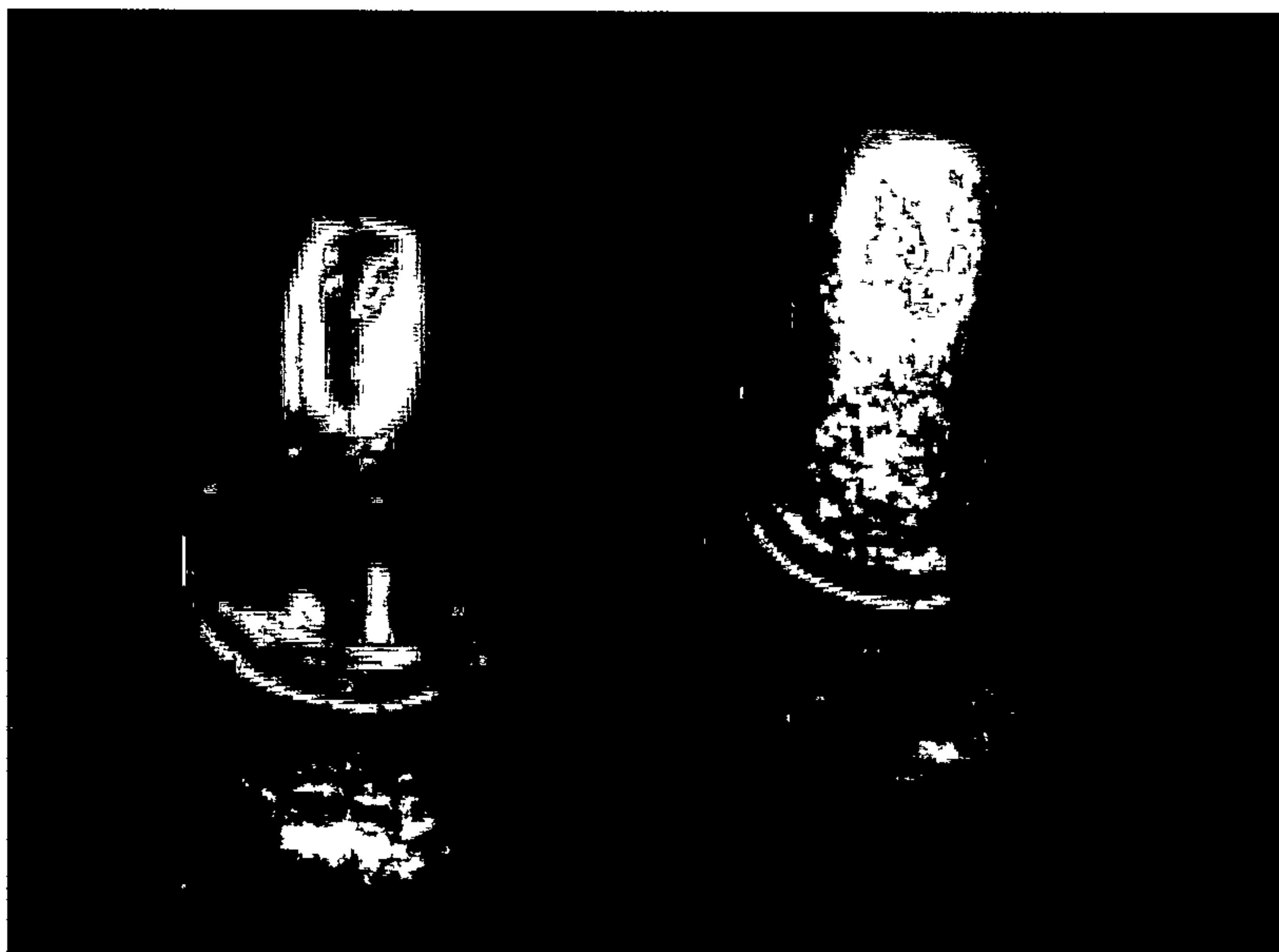


Figure 3.5. Master Brass Die on the left, with the polished surface and a duplicate brass die on the right. The duplicate die has a sandblasted surface to enhance the retention of the glass ionomer luting cement.

### 3.2.1.2 Construction of the In-Ceram® Crowns

The construction and firing schedules for the In-Ceram® core material and the Vitadur Alpha veneering porcelain are listed in tables 1 and 2 of Appendix II. Below are a series of photographs, which demonstrate how the alumina slip (In-Ceram®) is painted onto the special plaster die that represents the prepared tooth. The photographs then demonstrate the glass infiltration process and the final cleaned core. The Vitadur Alpha® is then built up onto the core using a water-based slurry of porcelain powder. This is called building the body of the crown and is a handcraft operation.

The first build is generally made approximately 10 percent oversize to accommodate the shrinkage that occurs during the vacuum firing process. The process involved constant measurement of 27 individual grid points to produce a series of crowns with the same approximate dimensions. After the first main body build, the porcelain is shaped using diamond burs as described in part 2.1.4.6. The crown is refired after porcelain additions are made to correct underbuilt areas and finally a glaze coating is applied and the crown fired once more to seal the surface. All crowns were finished with a high glaze (Vitachrom Delta® 740). All porcelain additions were made using the proven porcelain slurry-on-brush technique with hand vibration to reduced the incidence of air bubbles, (Aranda and Barghi 1988).

The consideration of thermal coefficient of expansion and its effect during firing onto a core substrate is explained by Asoaka et al., (1991). Basically dental porcelains behave like glasses and have very low viscosity at high temperatures which allows movement of the porcelain over the core as it cools. Where the core ceramic and the veneering porcelain have similar coefficients of thermal expansion, residual stresses between the two are then minimized. Stress release as measured by acoustic emission can be detected when the porcelain is hot enough to act as an elastic solid which occurs as it passes its glass transition temperature (Asoaka et al., 1992)

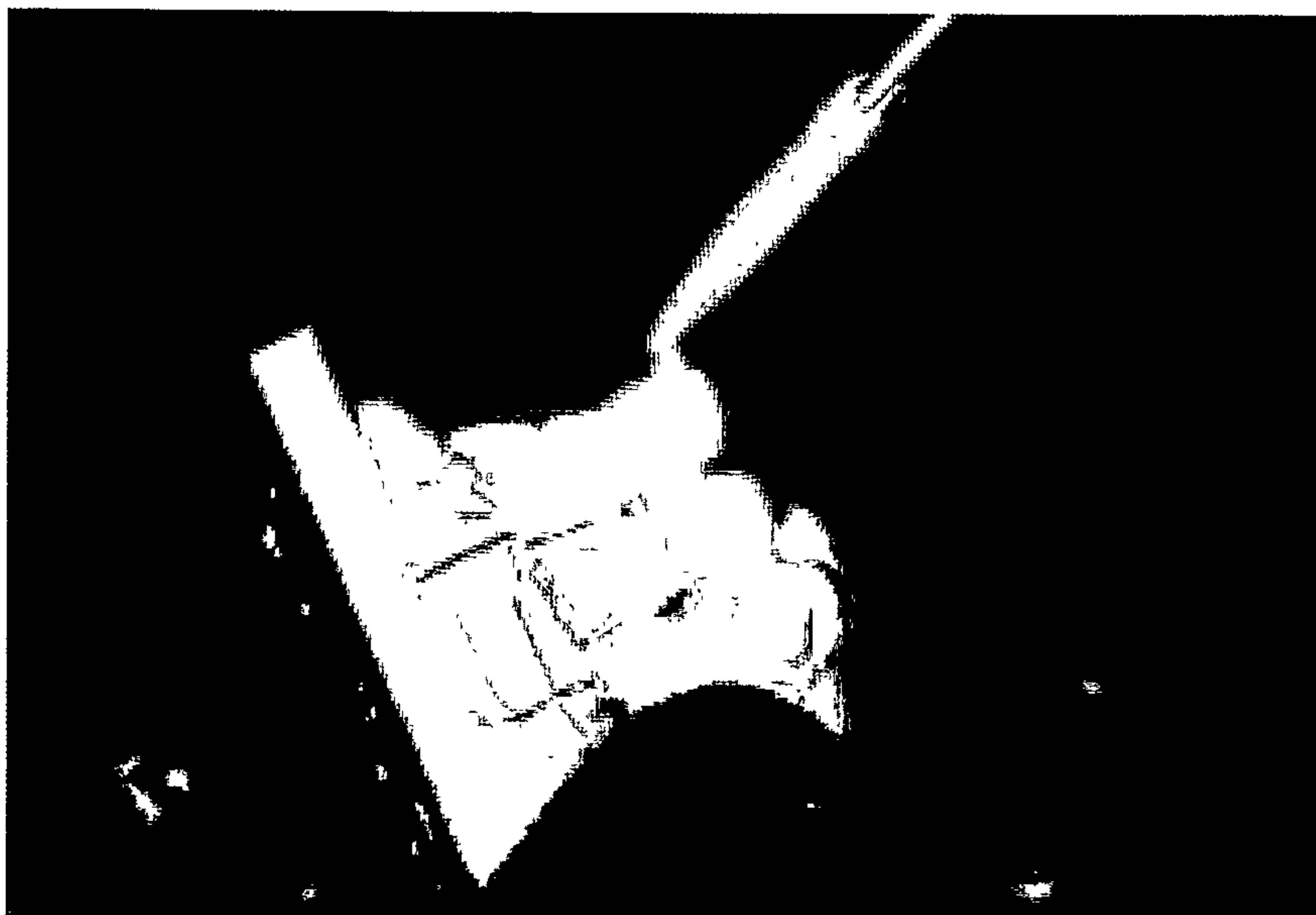


Figure 3.6 Painting the alumina slip onto the plaster die.



Figure 3.7 A plaster die prior to slip application on the left and after slip application on the right



Figure 3.8. The trimmed alumina coping with a slurry of lanthana glass powder placed on the outside. The coping and slurry are sitting on the platinum alloy foil.

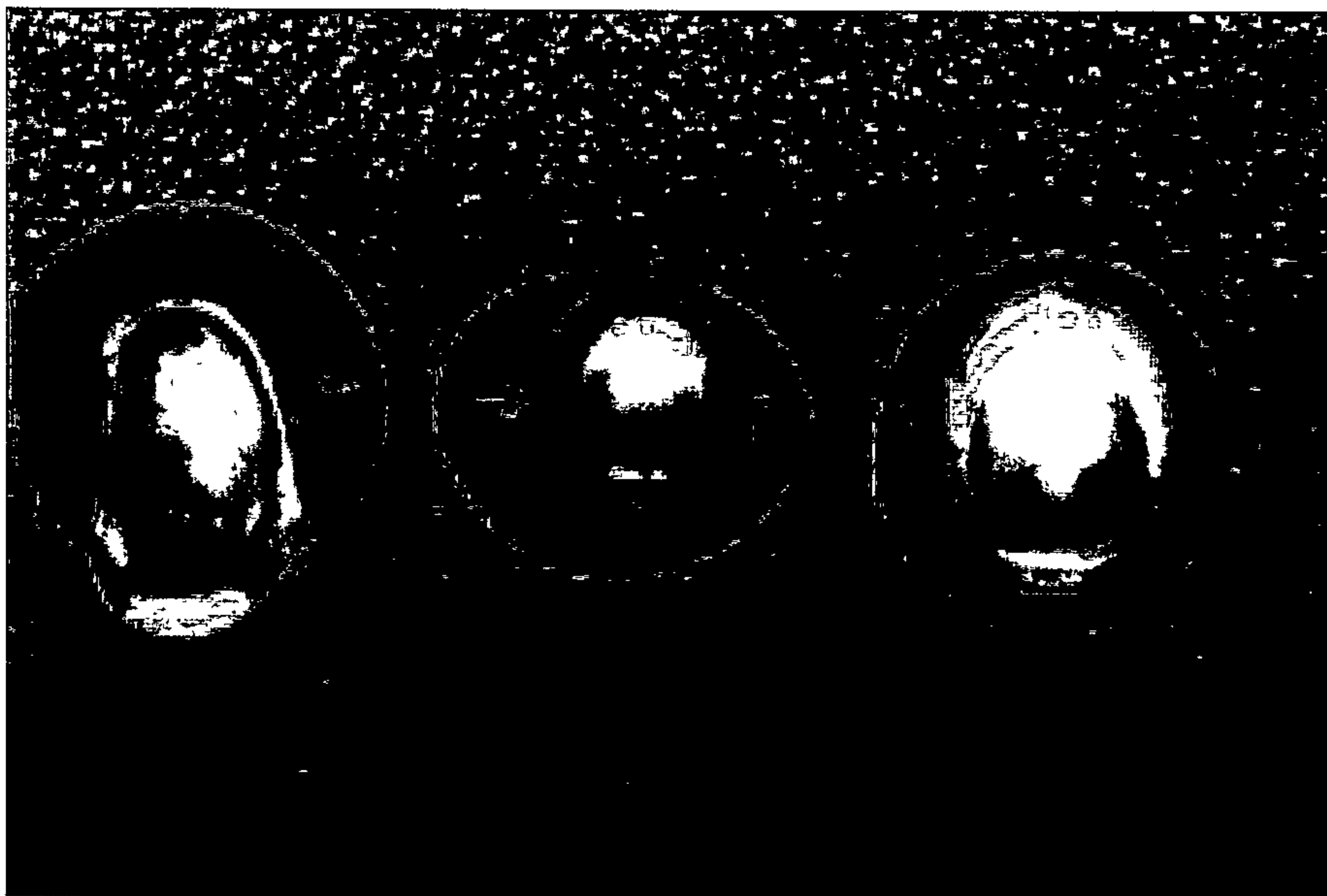


Figure 3.9 Examples of three In-Ceram® copings for single crowns after glass infiltration.

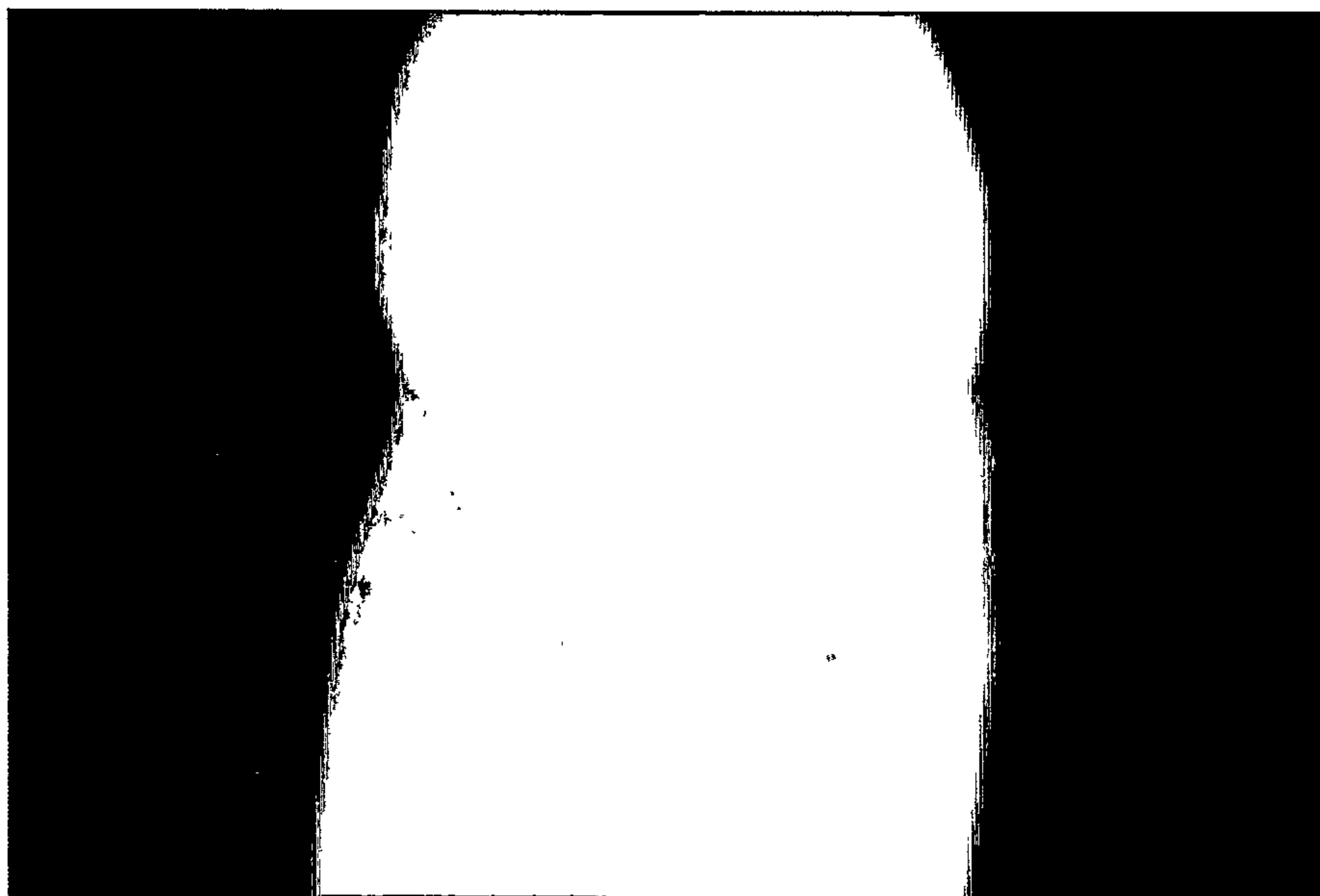
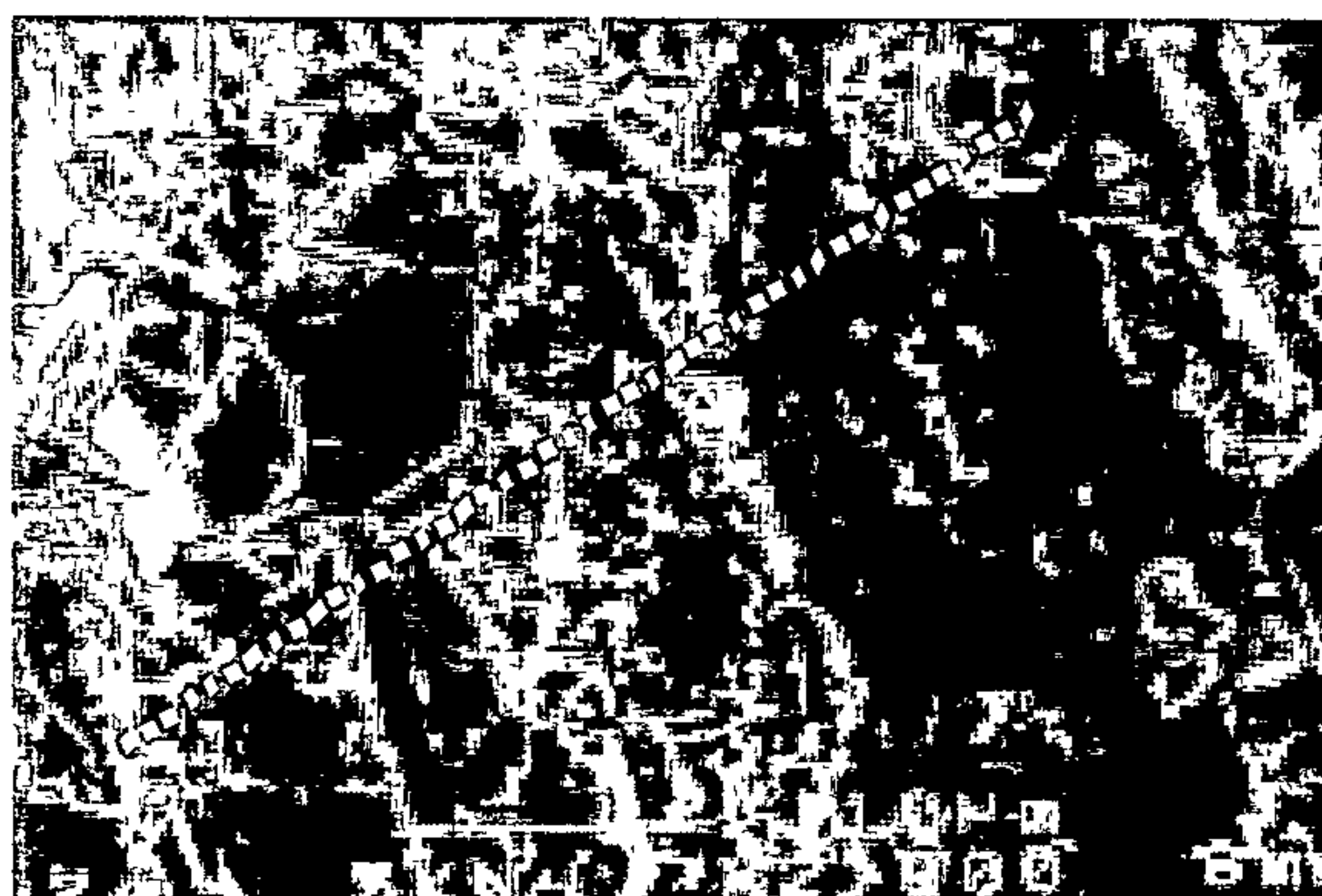
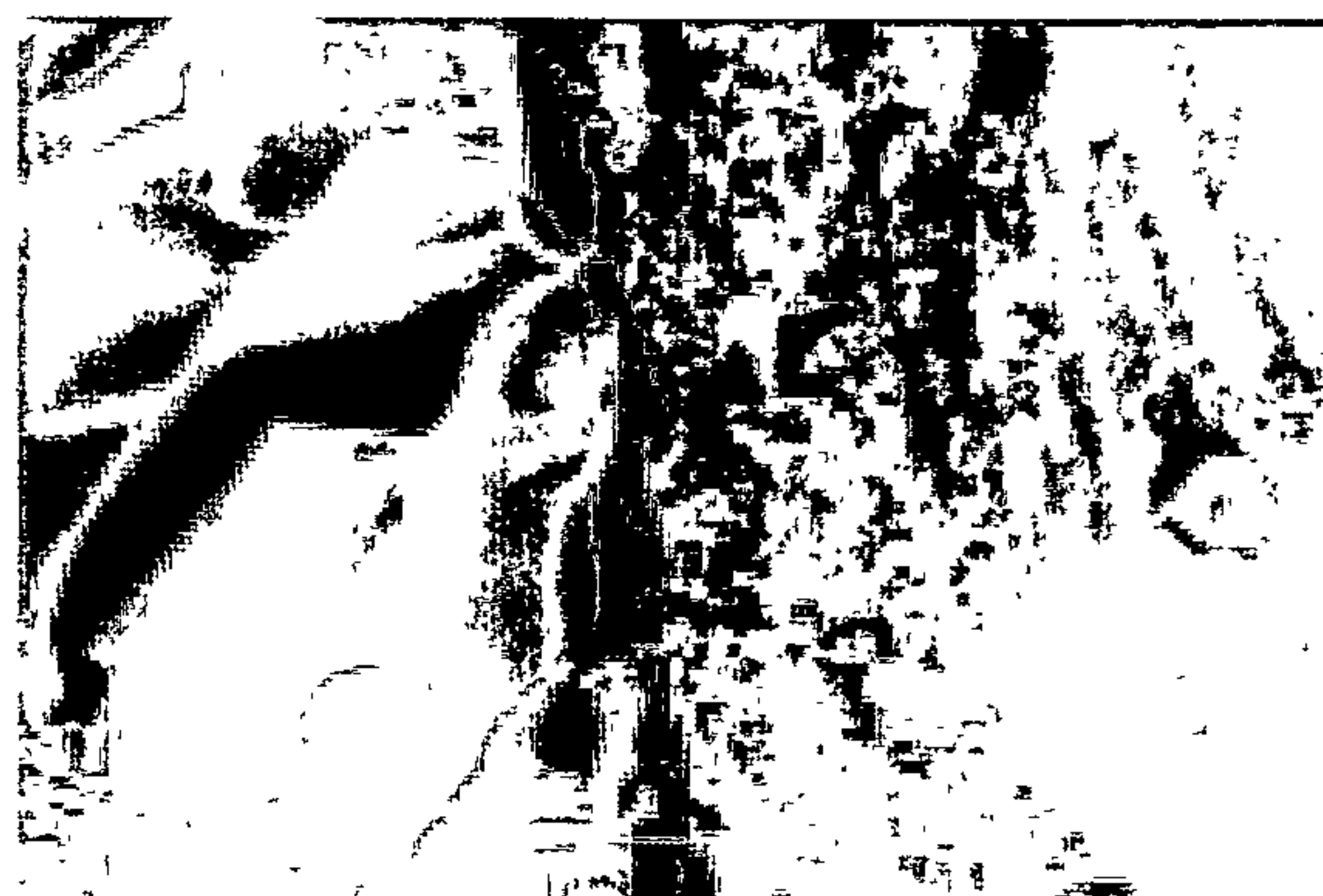


Figure 3.10. The completed coping after sandblasting, placed back on the master die.



———— 10  $\mu$ m

Figure 3.11a



———— 10  $\mu$ m

Figure 3.11b

Figures 3.11 a and 3.11 b are used to demonstrate the union between the In-Ceram® core material and the Vitadur Alpha® veneering material. The line of white dots in 3.11a is an approximate indication of the boundary between the two layers. The specimen in figure 3.11 b has been etched with 10% hydrofluoric acid to define the boundary more clearly. Both electron micrographs were taken at an original magnification of 2000x.

Carrier and Kelly, (1994), noted that it is important to ensure that excess glass from the infiltration process is removed to ensure adequate fusion of the Vitadur Alpha to the In-Ceram.

### 3.2.2 Testing Method

Following completion of crown construction they were cemented to the brass dies using a resin modified glass ionomer luting agent (3M Vitremer). The thickness of the luting agent was on average, 30  $\mu\text{m}$ . The brass dies were lightly grit blasted with 110  $\mu\text{m}$  alumina blasting powder to provide some mechanical retention for the luting agent. The glass ionomer was allowed to cure for 24 hours prior to testing. Luting the crowns to the dies was performed to prevent any untoward movement on the die. A study by Aboush et al., (1989) demonstrated the effect adhesive systems can have on tensile strength of the crown material, good adhesion resulting in an increase in loads for fracture. This is in agreement with Dérand (1974 and 1991). Davidson et al., (1991) have explored the high stresses that can be developed between the crown and tooth surfaces when they are bonded with a thin layer of composite resin and it was for this reason that resin was not used as it may have unduly influenced the result.

A separate series of tests was conducted to assess the difference between grit blasted and polished dies. The results appear in table 3.5. There was a significant difference between the two groups indicating that constraining the luting agent had an effect on the failure load. This part of the procedure was required because the testing of experimental In-Ceram discs would be carried out without luting agent. It is not practicable to lute the discs to the three mounting balls used in the bi-axial flexure test. It is assumed that the geometry of the test rig and the disc specimens combined with the direction of load would lend themselves to prevent slippage. However the crowns would need to be luted to the brass dies to prevent slippage during the test because they are a very different shape.

The crowns and their individual dies were then mounted in a jig and tested. A diagram of the loading direction appears in figure 3.12. The position and orientation of the load were arranged so that the effect would be principally on the palatal portion of the crown that would be supported by either a metal or dentine core. This would then enable consistence for the later development of an experimental model using discs to examine design considerations for the use of the ceramic. This was influenced by the report of Anusavice and Hojjatie (1988) who pointed out that it was the orientation of the applied load that influenced the distribution of stresses in ceramic crowns. All tests were conducted with a crosshead speed of 0.5 mm per minute

A photograph of a crown after testing in the test rig appears below in figure 3.13. and a photograph of the set-up in the Shimadzu Autograph universal testing machine appears in figure 3.14

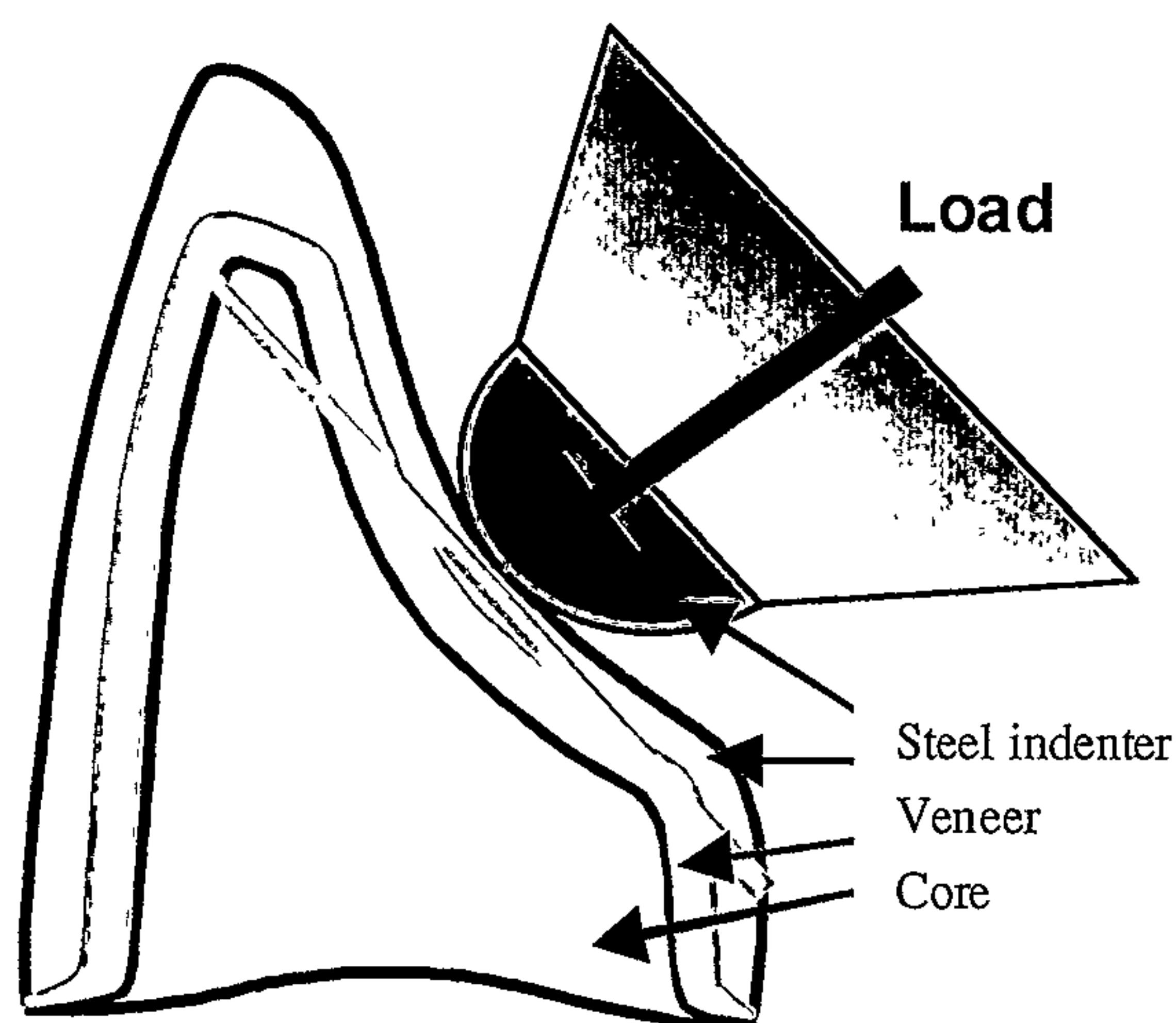


Figure 3.12. Diagram of direction of the load application to the in vitro crown. The load (red arrow) was aligned at right angles to the plane of the palatal surface (green line) at the contact point.



Figure 3.13 The hard steel indenter can be seen at its contact point in the middle of the palatal surface of the test crown. The palatal surface is at right angles to the loading axis and the crosshead speed was 0.5 mm per minute. The die can be seen loaded into the shaft of the mounting block where it is held firmly in place by a steel screw threaded from the top of the mounting block.

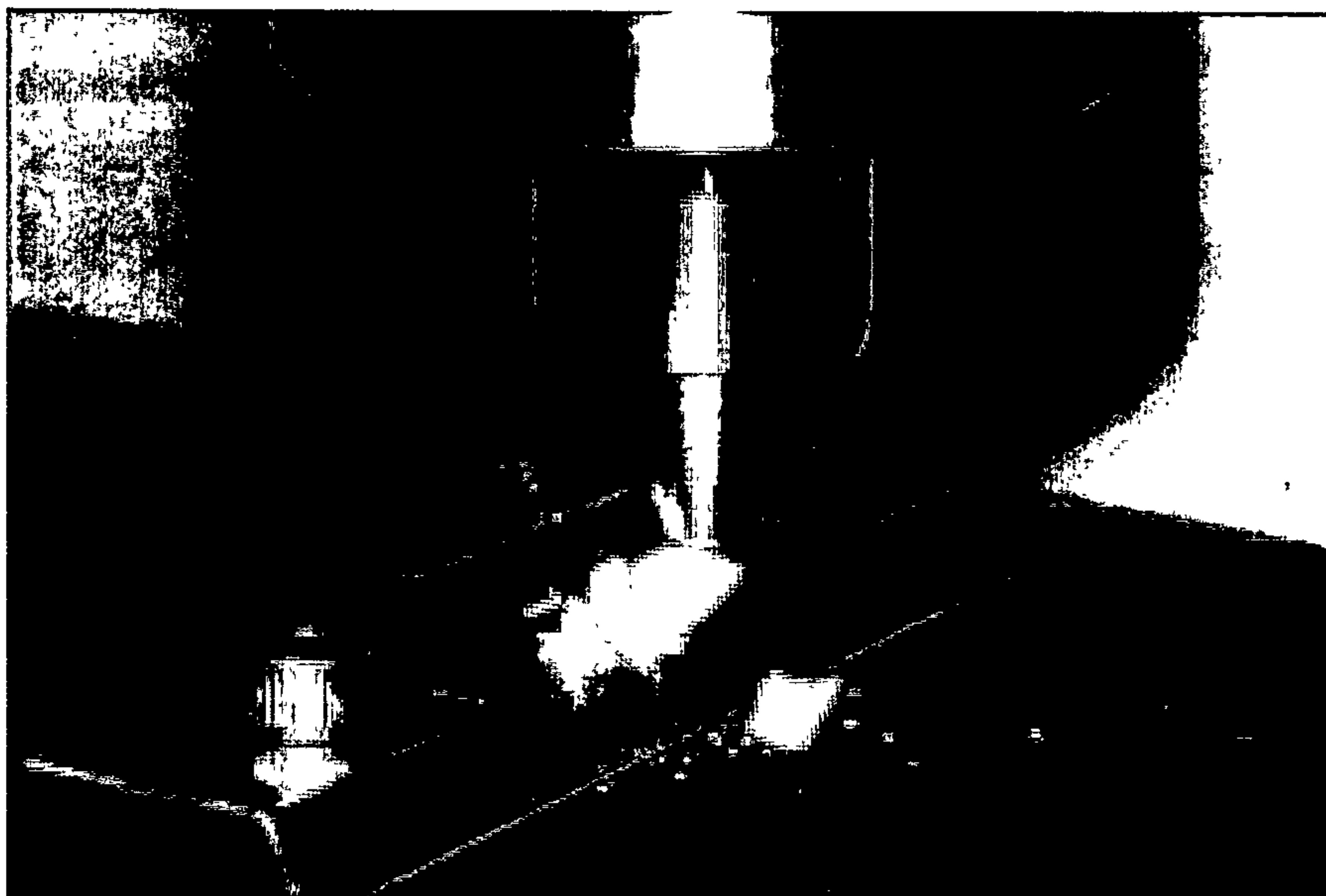


Figure 3.14 Extended view of the testing rig with a fractured crown following loading.

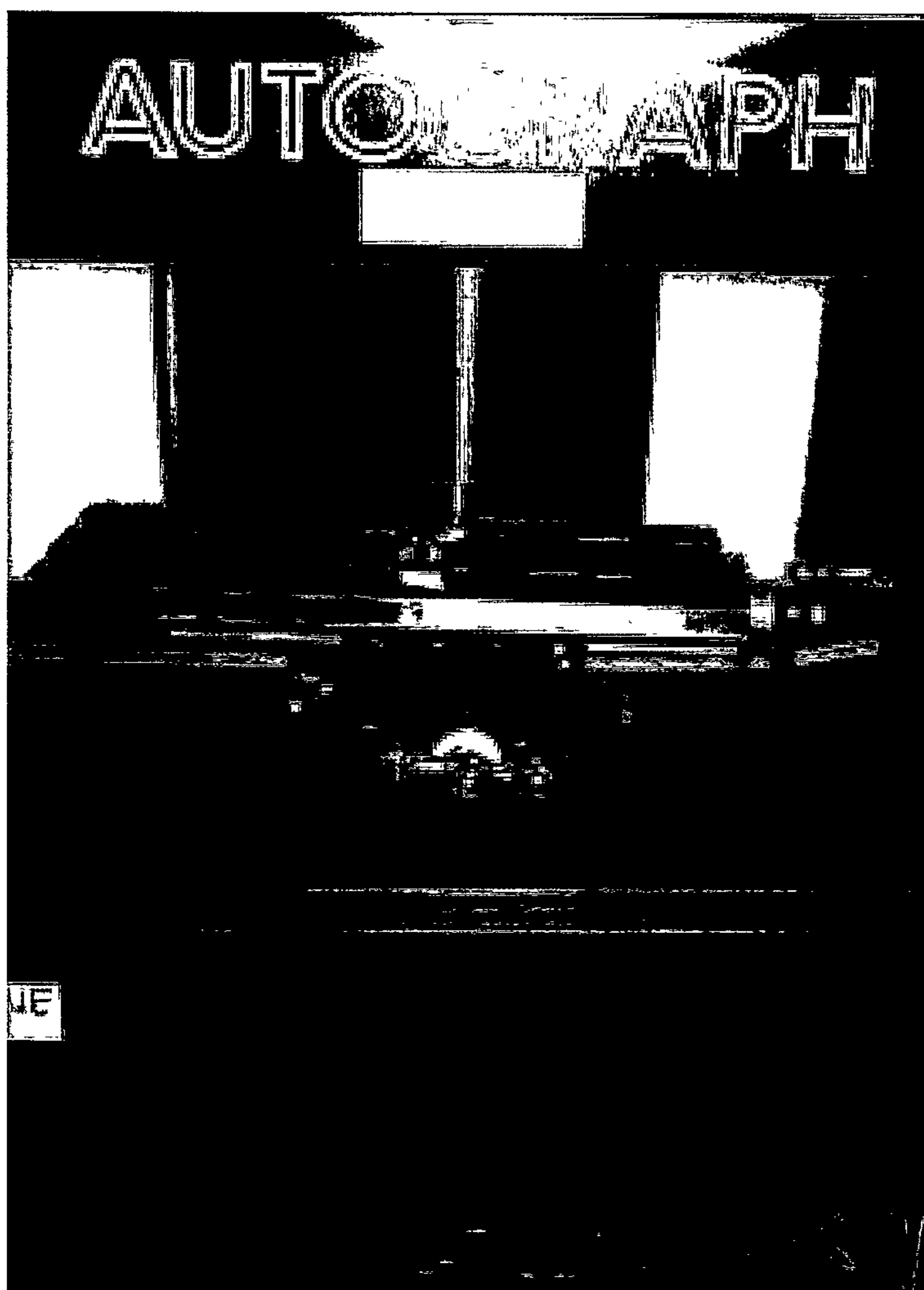


Figure 3.15 View of the test mounting stage with adjustment capability for both horizontal axes.

3.3 Results.

The single outstanding feature was that all crowns were distinguished by a circular crack at the indenter contact point. The crowns all appeared to have two parts to their failure. The first was the development of the classical Hertzian spherical indentation induced cone crack. This was accompanied by an audible click from the specimen and a recorded step in the simultaneous graphic recording of the load versus displacement. The force display was linear until the click, which was recorded as the initial failure load. The display then stepped down while the veneering porcelain crushed beneath the indenter and recovered to climb again as the load was applied to the underlying core material. A scanned copy of the recorded trace for specimen 56N is displayed below in figure 3.16.

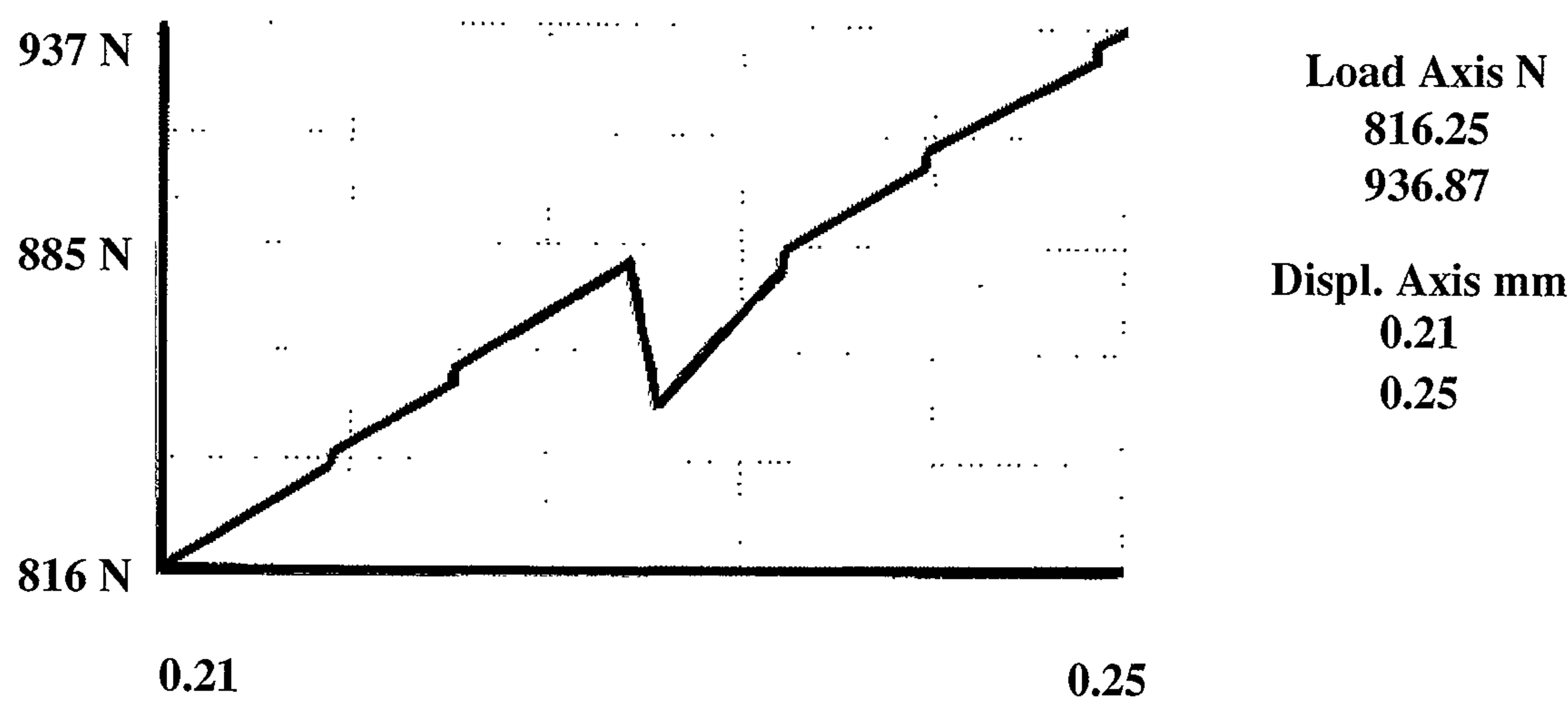


Figure 3.16 Scanned copy of the graphical recording of the load versus displacement during the test of crown number “56N”. The initial crack was recorded at a load of 885 Newtons and the final failure occurred at 937 Newtons.

The results from the test series are listed in tables 3.1 and 3.2. In conjunction with the numerical results are the observations about the crack propagation directions. The crack patterns are several and it is almost impossible when viewing the fractured crown to determine which was first. Diagrams and photographs of the different examples can be seen below in figures 3.17 to 3.20. The crack patterns did fall into two basic groups. The first, designated type "I", has a circular crack at the point of contact, with superimposed vertical and horizontal radial cracks both of which extend around the crown and invariably led to the complete failure of the crown on the die. The second, designated type "II" also has a circular crack at the contact point, but the extended cracks are haphazard and did not necessarily lead to the complete failure of the crown. In both instances there was evidence of delamination from the reflected light that developed from the separation of the porcelain

and the core material. (See the region just below the contact site in figure 3.17b as an example)

**Table 3.1      Comparison of load, core and veneer thicknesses, and fracture pattern**

| Initial Failure Load<br>N | Final Failure Load<br>N | Core Thickness<br>(mm) | Veneer Thickness<br>(mm) | Fracture Type |
|---------------------------|-------------------------|------------------------|--------------------------|---------------|
| 795                       | 804                     | .700                   | .430                     | I             |
| 855                       | 762.5                   | .710                   | .500                     | I             |
| 551                       | 885                     | .430                   | .520                     | I             |
| 1275                      | 1346                    | .720                   | .800                     | I             |
| 1337                      | 1345                    | .520                   | .580                     | II            |
| 1424                      | 1423                    | .540                   | .900                     | I             |
| 1191                      | 1196                    | .540                   | .550                     | II            |
| 1466                      | 1470                    | .400                   | .800                     | I             |
| 1198                      | 1300                    | .640                   | .730                     | I             |
| 1078                      | 1078                    | .680                   | .910                     | I             |
| 998                       | 998                     | .620                   | .640                     | I             |
| 1201                      | 1201                    | .420                   | .820                     | I             |
| 1145                      | 1146                    | .500                   | .780                     | I             |
| 1465                      | 1476                    | .820                   | .370                     | I             |
| 1378                      | 1420                    | .700                   | .360                     | II            |
| 1322                      | 1323                    | .620                   | .670                     | I             |
| 629                       | 709                     | .340                   | .880                     | I             |
| 1187                      | 1465                    | .900                   | .400                     | II            |
| 885                       | 937                     | .540                   | .710                     | I             |
| 844                       | 876                     | .500                   | .450                     | I             |
| 1055                      | 1710                    | .740                   | .500                     | I             |
| 802                       | 813                     | .420                   | .410                     | I             |
| 1149                      | 1168                    | .400                   | .430                     | I             |
| 1061                      | 1061                    | .580                   | .610                     | I             |
| 1192                      | 1192                    | .730                   | .350                     | I             |

**Table 3.2      In vitro failure loads and thicknesses for In-Ceram crowns of a standard design.**

| Standard In-Ceram® Crowns | Mean load<br>MPa | Std Dev. | Mean Core<br>mm | Std Dev. | Mean Veneer<br>mm | Std Dev. |
|---------------------------|------------------|----------|-----------------|----------|-------------------|----------|
| Non-Collared n = 25       | 1099.3           | 252.1    | 0.613           | 0.133    | 0.604             | 0.185    |
| Type I, n = 21            | 1078.5           | 260.2    | 0.571           | 0.136    | 0.627             | 0.183    |
| Type II, n = 4            | 1252.0           | 109.1    | 0.713           | 0.180    | 0.437             | 0.100    |

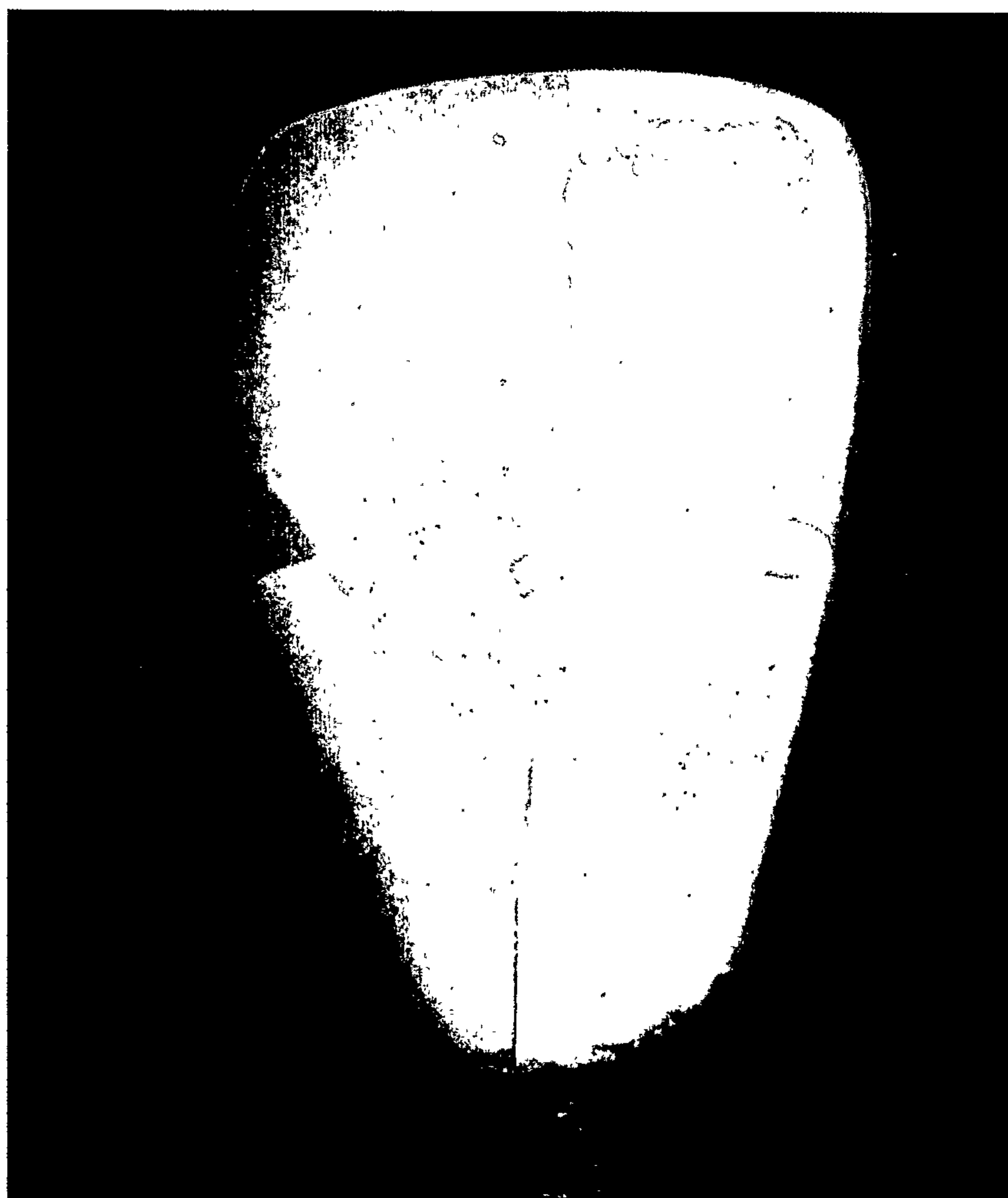


Figure 3.17a Photograph of the crack pattern through an in vitro crown specimen, which has been designated as Type I.



Figure 3.17b Photograph of the crack pattern through an in vitro crown specimen, which has been designated as Type II. The light areas indicated by the arrows are areas of delamination.

The crown specimen in figure 3.18 was from the standard design group. The initial fracture produced the cone crack, part of which has been defined by the arrow. The test then continued and the area below the contact area crushed and shattered, allowing pressure to build up against the core material and inducing the median and radial cracks that caused the crown to finally split and allow a cross sectional view of the insult. The picture shows one view of the clinical reality of variation in veneer porcelain and core ceramic thicknesses. The example below is relatively uniform.

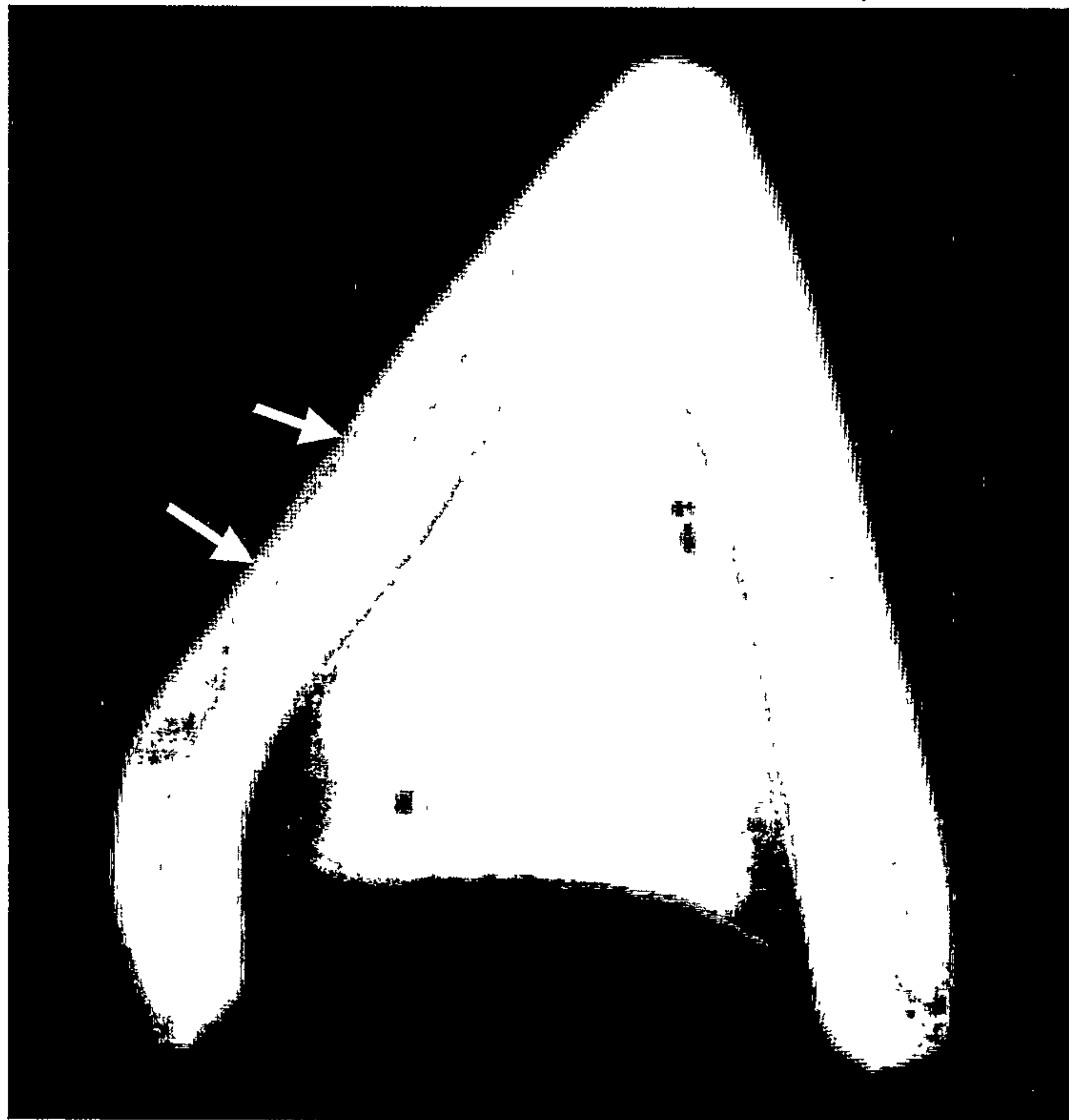


Figure 3.18

Figure 3.18      Photograph of the Hertzian cone crack from a fractured crown. The origins of the cone crack outline are arrowed. The crown was from the group designated Type I.



Figure 3.19a Photomicrograph of the Hertzian cone crack from a fractured crown. The crown was from the group designated Type I. The classic outline of the cone crack can be seen emanating from the contact point on the upper surface.

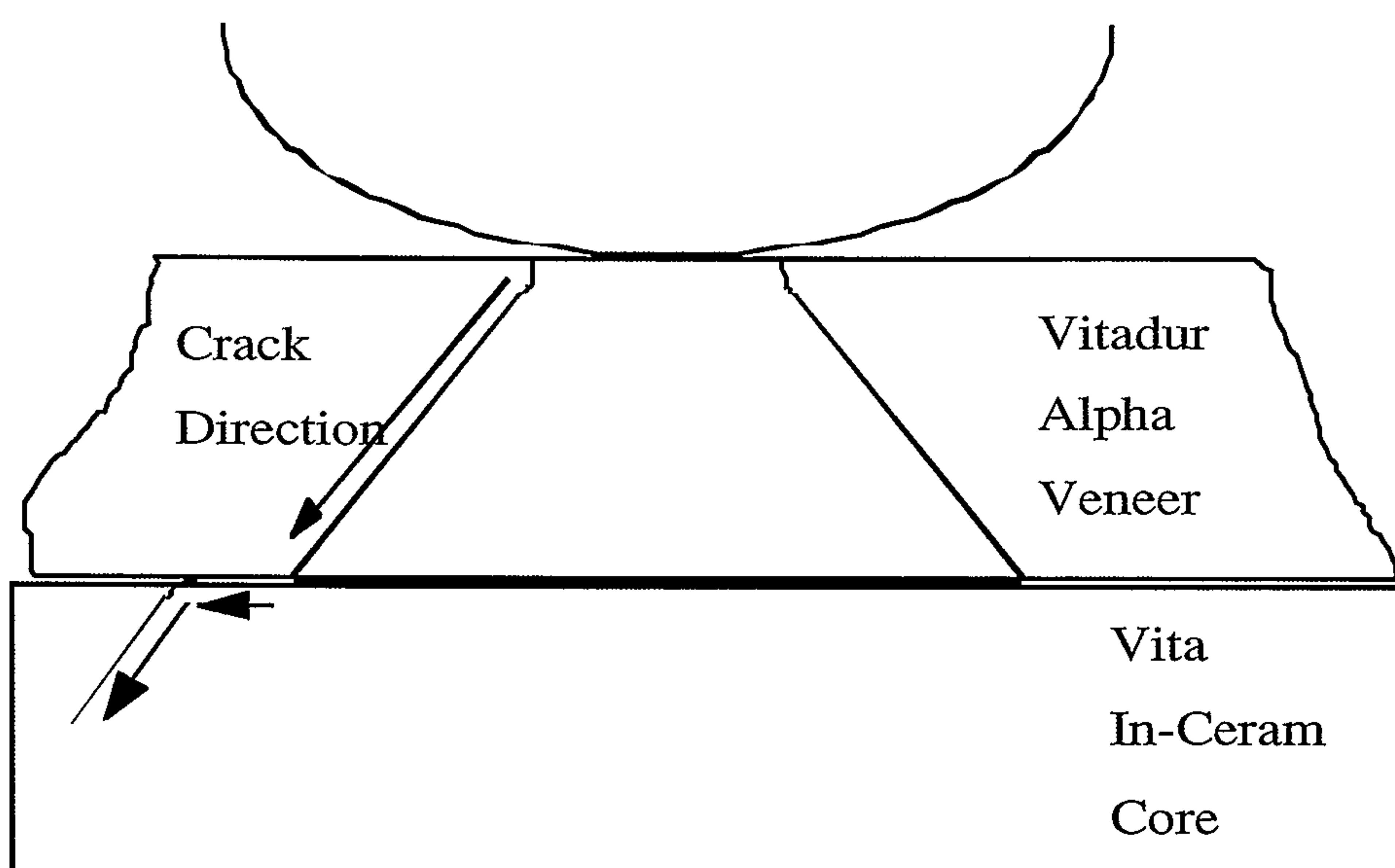


Figure 3.19b

Figure 3.19b Diagram of the Sectional view of Type I fracture Hertzian cone crack through an in vitro Type I crown specimen.

The crack propagation from the cone crack to the interface and then onto a delamination is seen in figure 3.20a. The cone crack started with almost classical angulation, but diverted when it encountered the interface and then in this instance chose to travel along that boundary between the two layers. This was not visibly evident to the naked eye in the type I fractures, but it was almost universal for type II fractures with their haphazard crack directions.

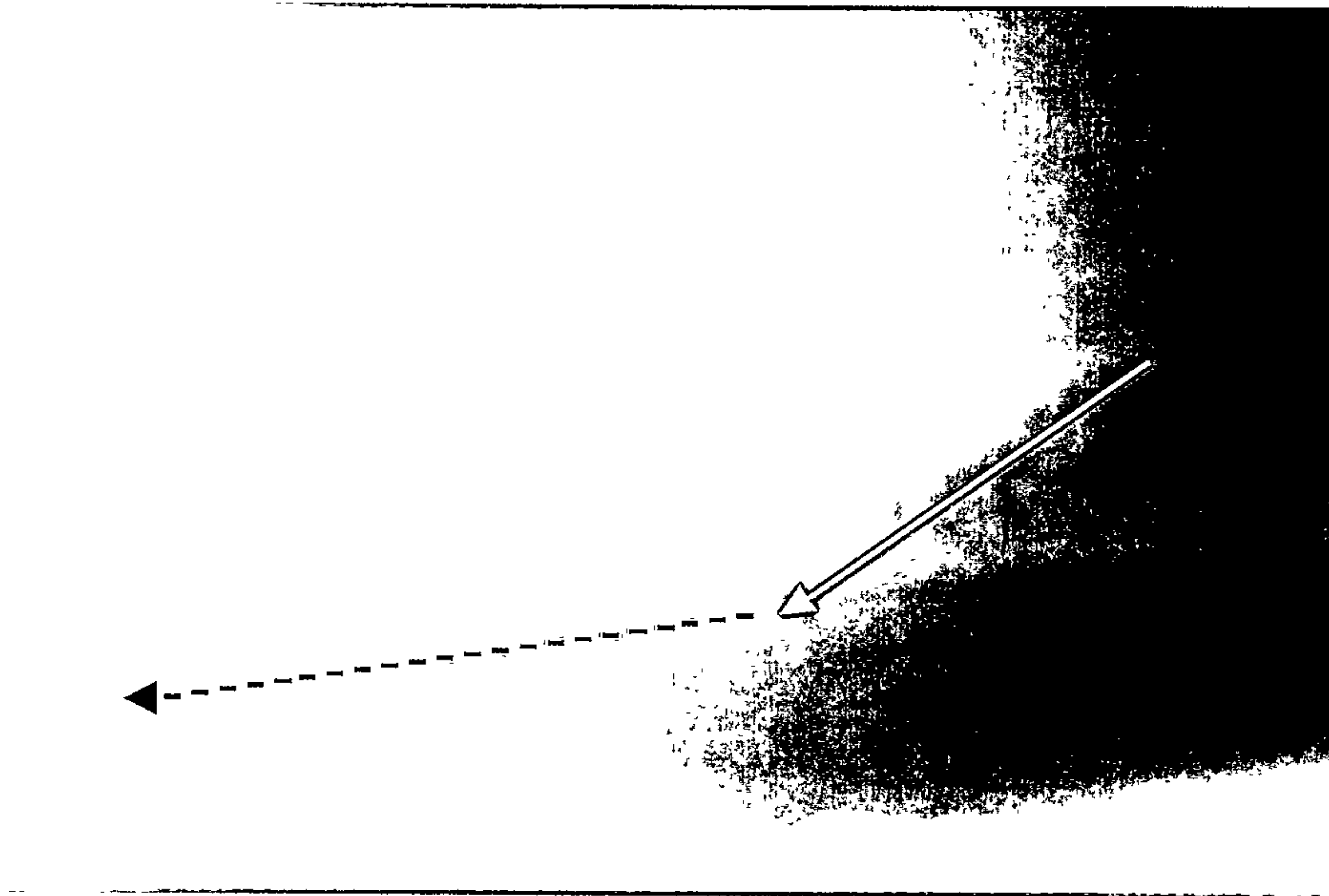


Figure 3.20a Photograph of the cross section of the crack site. The bottom layer is the In-Ceram core and the next layer is the Vitadur Alpha porcelain veneer. The cone crack and its direction of travel is outlined by the solid white arrow. The delamination is outlined by the broken arrow.

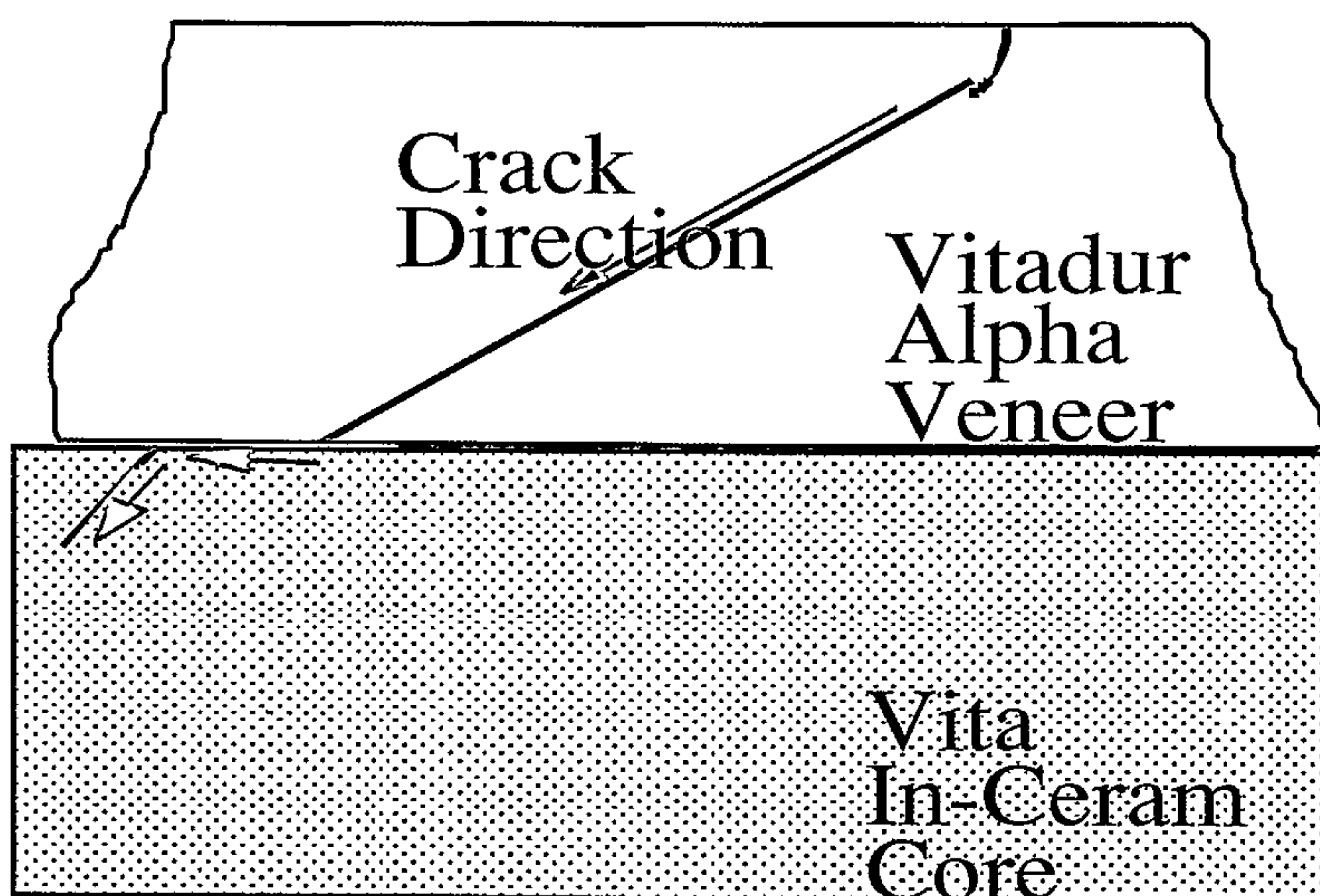


Figure 3.20b Diagram of the crack as it proceeds along the interface and deflects into the core layer.

### 3.3.1 Comparison of Results

The results for the loads were tested to see if there was any correlation between the load at initial failure and either the core thickness or the veneer thickness. The scatter plots for the core thickness vs initial fracture loads and veneer thickness vs initial fracture loads are depicted below. The plots in each case seem to be widely varied but there is a weak correlation between core thickness and initial fracture load and a slightly smaller correlation between veneer thickness and initial fracture load. These correlations are positive. The results were also tested to determine whether there was any significant correlation between the loads for type I and type II fracture patterns. The result was a weak positive correlation for type I fractures and a stronger correlation for type II fractures. However, the number of type II fractures for this group was only four and the results were highly variable. Correlations were tested using Fisher's R – Z test.

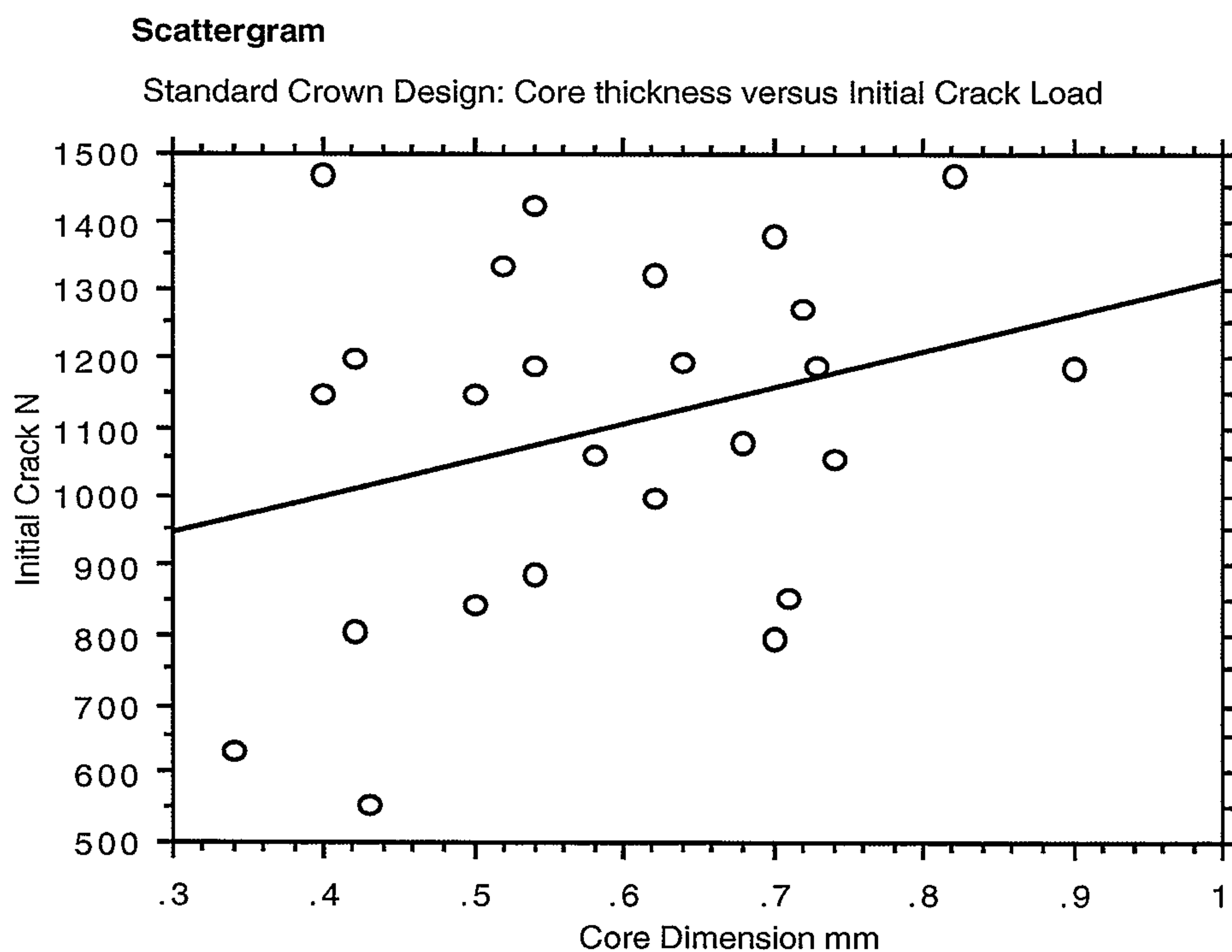


Figure 3.21 Scattergram of the results for core thickness versus load for the initial crack, regardless of crack pattern.

The scattergrams in figures 3.21 and 3.22 have produced a random spread, and the correlation coefficients were weak, hence the flat straight line, but the nature of the cracking pattern was not taken into account. When the fracture pattern was used to select the results into two different groups the correlations between layer thicknesses and loads became more apparent.

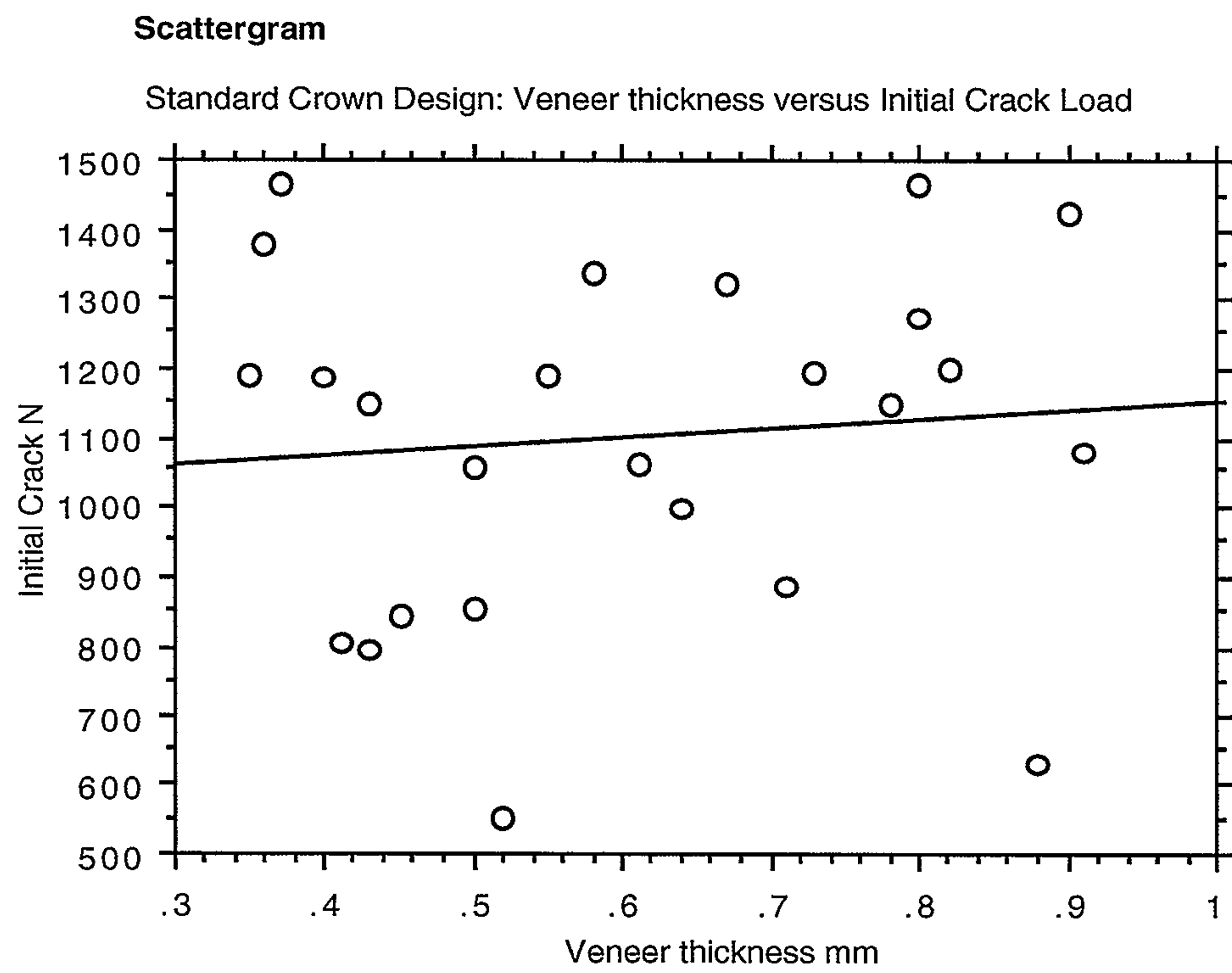


Figure 3.22 Scattergram of the results for veneer thickness versus load for the initial crack, regardless of crack pattern.

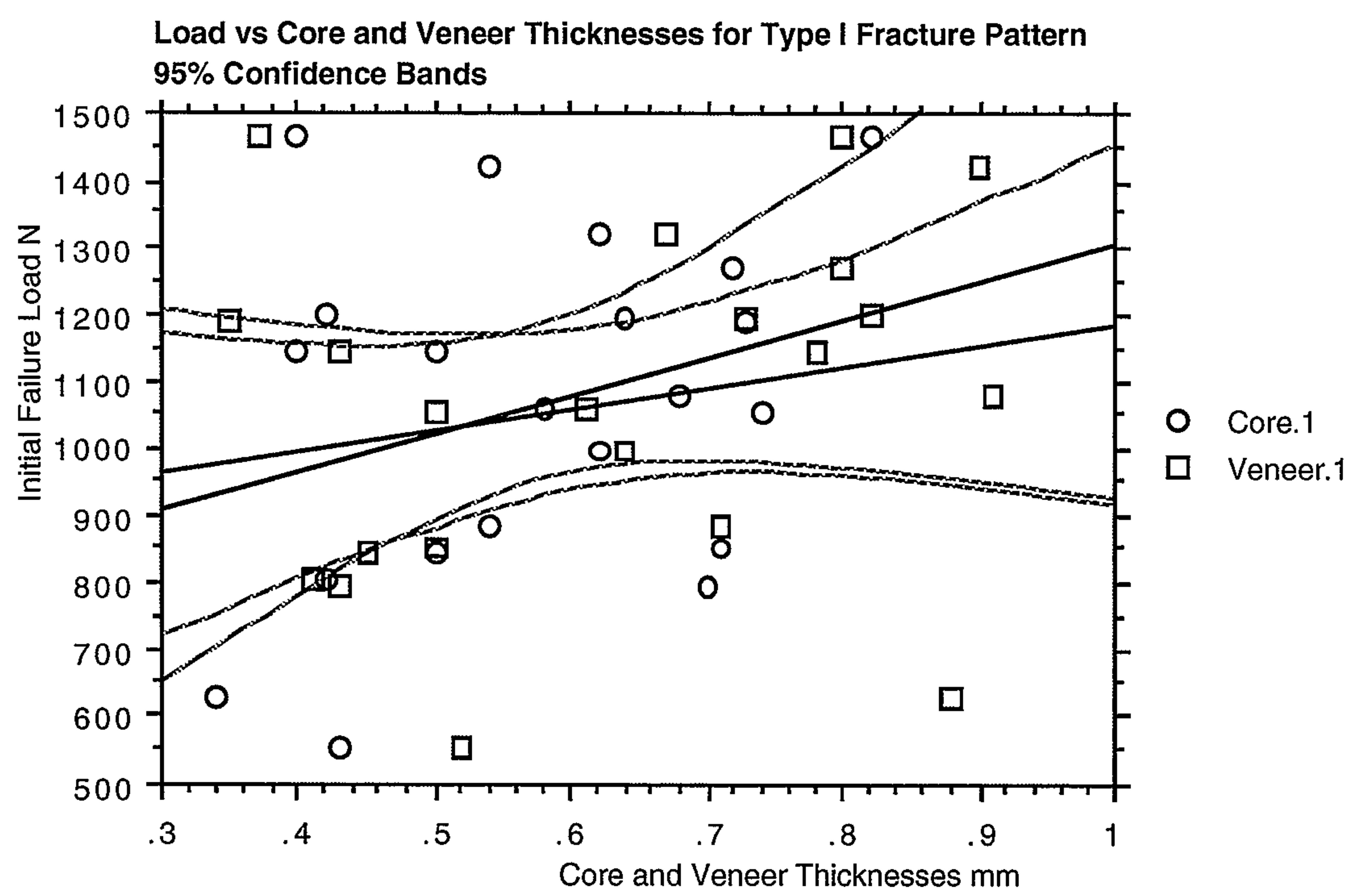


Figure 3.23 Scattergram of the results for core and veneer thicknesses versus load for the initial crack, for *type I* crack pattern. (n = 21)

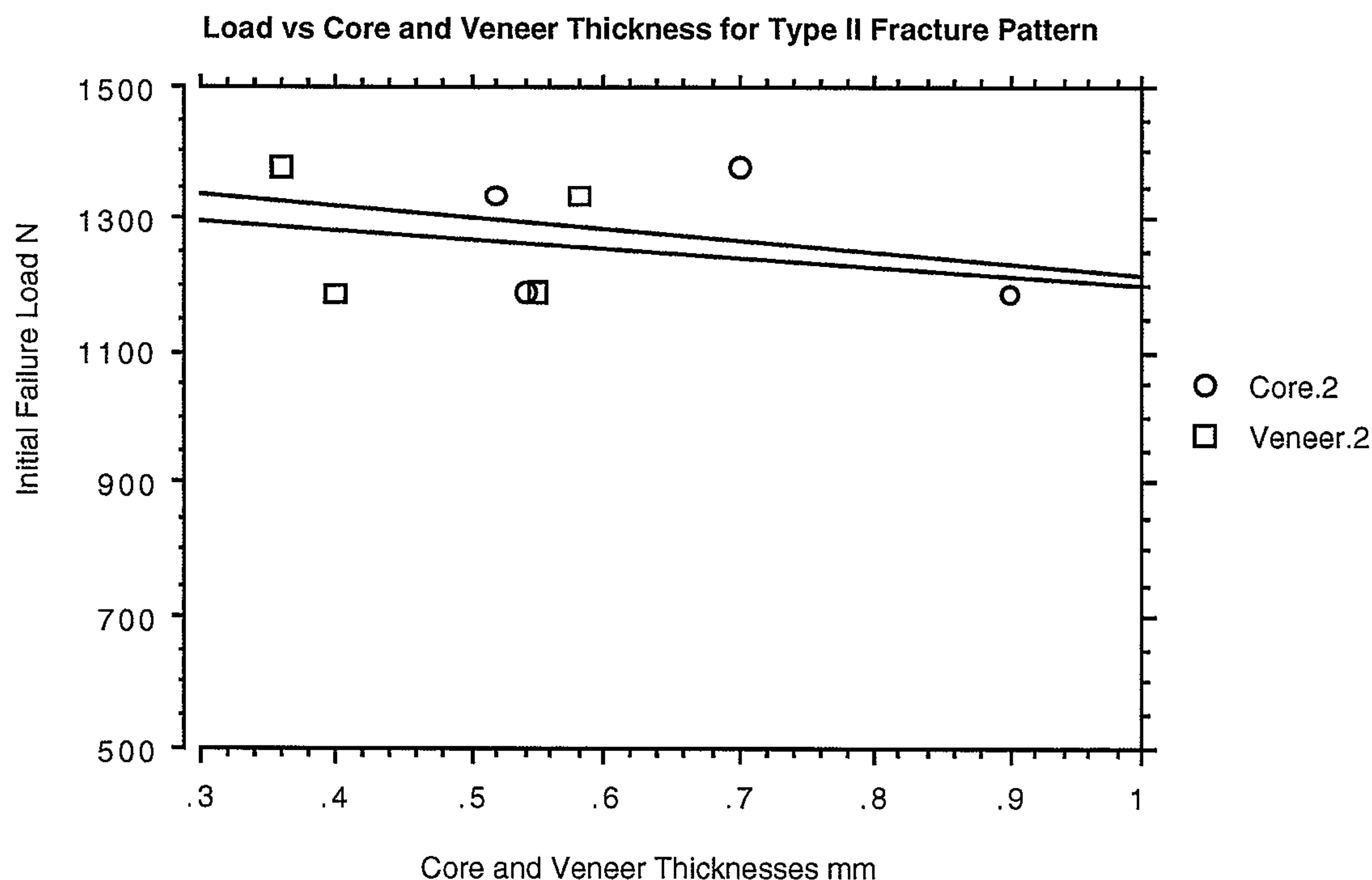


Figure 3.24 Scattergram of the results for core and veneer thicknesses versus load for the initial crack, for *type II* crack pattern. (n = 4)

The pattern for type I fractures in figure 3.23 has a little more order, but it still required testing to give a meaningful result. The result in figure 3.24 is not as reliable because the numbers were small. The correlation coefficients were produced using a proprietary statistical package, (Statview 4.02, Abacus Concepts, Berkeley, Calif. USA) and appear in table 3.3 below.

Table 3.3 Comparison of Correlation Coefficients – Initial Crown Fracture Tests.

| Initial Fracture Load (N) | Core Thickness | Veneer Thickness |
|---------------------------|----------------|------------------|
| All crowns                | 0.304          | 0.228            |
| Type I (1078.5)           | 0.276          | 0.210            |
| Type II (1252.0)          | 0.617          | 0.480            |

From these results it can be seen that a positive correlation exists between both core thickness - load and veneer thickness – load for both type I and II fractures. Given the number of specimens tested it would seem reasonable to assume that the initial fracture will occur at approximately 1 kN under the conditions of this experiment. It also appears

that an increase in core thickness at the expense of the veneer thickness while maintaining the same overall thickness will produce a higher initial and final fracture resistance.

3.3.2 Weibull statistics for the standard crown design.

The load data for the crowns were ranked according to Weibull statistics for reliability. The Weibull modulus for this group was 4.1. In terms of conventional industrial as-finished ceramic products, a Weibull modulus of approximately 10 is considered typical, (Lawn 1993). It needs to be noted that the advantages for the industrial ceramic processes, such as hot pressing and ceramic formulations using non-oxide ceramics for increased strength and reliability are not available to most dental ceramic systems because aesthetics and natural tooth colour are requirements.

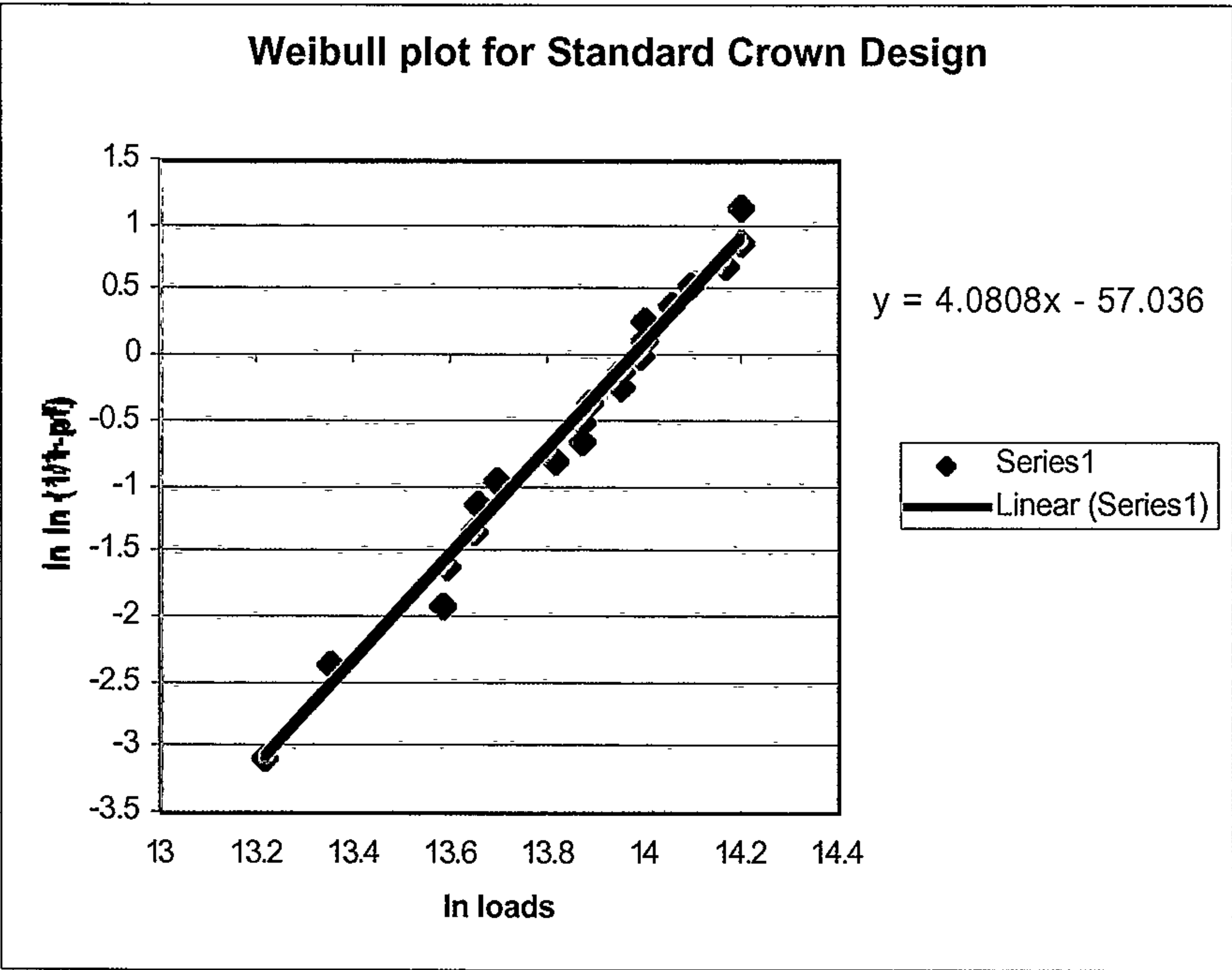


Figure 3.25 Weibull plot for the standard crowns built according to the design from the manufacturer's instructions.

3.4 Discussion.

The loads experienced in the tests were 5 to 10 times higher than would be expected in normal function. The following table (table 3.4) from Fields et al., (1986) serves to place the results in perspective. These loads were recorded for normal chewing function. The interesting aspect of their work was that the power from the masticatory muscles was greater at wider mouth openings, not when the teeth were together. Large vertical openings resulted in a progressive increase in maximum bite force to a maximum at about 20 mm, followed by a decrease then a second increase to near-maximum at about 40 mm for

young adults. However other studies have shown that during subconscious bruxism the biting forces can be very high.

The failures being discussed are those we would expect from the first section of Lawn’s “Bathtub” curve in figure 3.1. To achieve fracture at normal biting forces requires a radius of 3 – 10 times smaller than used. According to Auerbach’s law, (chapter 2 part 2.4.1.4) it was argued that the initiation of the cone crack was due to a critical load which was proportional to the radius of the indenter, i.e.  $P_c \sim r$ . This might occur for unworn teeth, or during abnormal contacts, such as an interference during lateral excursions, where the corner between an axial surface and an occlusal surface of a tooth impinges on the porcelain. In a bruxer these corners can be very sharp and would therefore have extremely small radii creating high stress concentrations in the instance where a lateral interference occurs. Such interferences could be encountered when trying the crown in and testing the evenness of the occlusion.

**Table 3.4 Biting forces (kg) for three age groups.**

| a. Mean vertical occlusal force (kg) for 6-12 year-old children |                  |         |                |         |
|-----------------------------------------------------------------|------------------|---------|----------------|---------|
|                                                                 | Vertical Opening |         |                |         |
|                                                                 | 2.5 mm Opening   |         | 6.0 mm opening |         |
|                                                                 | Mean (kg)        | SD      | Mean(kg)       | SD      |
| Swallowing                                                      | 1.94             | (2.49)  | 2.60           | (2.53)* |
| Chewing                                                         | 7.23             | (6.30)  | 10.18          | (6.02)* |
| Maximum Force                                                   | 14.47            | (12.66) | 15.20          | (10.38) |
| b. Mean vertical occlusal forces (kg) for adolescents           |                  |         |                |         |
| Swallowing                                                      | 1.8              | (1.6)   | 3.0            | (2.3)   |
| Chewing                                                         | 7.3              | (3.9)   | 9.0            | (5.8)   |
| Maximum Force                                                   | 14.5             | (7.5)   | 16.1           | (8.5)   |
| c. Mean vertical occlusal forces (kg) for young adults          |                  |         |                |         |
| Swallowing                                                      | 2.9              | (3.7)   | 4.8            | (4.8)*  |
| Chewing                                                         | 13.5             | (10.4)  | 16.2           | (13.8)  |
| Maximum Force                                                   | 31.0             | (20.0)  | 35.6           | (18.7)  |

\*Differences between openings significant at  $p \leq 0.01$  level

The results from the crown tests are comparable to those of Lampe et al., (1996) where the effect of luting agent in preventing fracture of all-ceramic crowns was compared. There was a significant difference between the crowns with no luting agent and those where

either composite resin or glass ionomer were used to cement the crown to the test die. For this reason a separate group was constructed as mentioned in part 3.2.2 to test the effect of the system used to lute the crown to the die. The design for this group was the same as the reinforced design described in chapter 6. The results appear in table 3.5 below, with the probability that there was a difference being highly significant. The polished dies offered limited mechanical bonding to the glass ionomer luting agent and it was assumed that this instability caused the lowered fracture loads because the crowns were not supported by the adhesion to the die. This gave a reasonably high probability that the crowns were slipping to some degree as the luting agent sheared from the die under load.

**Table 3.5      In vitro comparison of load to failure using surface condition of the specimen die as a variable.**

| Modified In-Ceram® Crowns            | Mean Load<br>Newtons | Std Dev. |
|--------------------------------------|----------------------|----------|
| Polished Die                         | 1159.4               | 227.0    |
| Grit blasted Die                     | 1570.3               | 247.5    |
| Significant difference: $P > 0.0078$ |                      |          |

From the optical observations of the fractured crowns, figures 3.17 a & b it can be seen that the directions of the various cracks are highly variable. This can be explained in terms of the differing physical properties of the two component layers of the crown and the various mechanisms associated with fracture mechanics described in chapter two. In all instances examined it appears that fracture initiated from the loaded surface of the feldspathic outer veneer. They then travel either as radial cracks or, more classically, as Hertzian cone cracks through this layer until the inner glass infiltrated alumina core layer is encountered. At this interface there is a change in the stress field because there is an elastic mismatch between the two layers and also two options for the crack. The crack can extend through the interface into the higher toughness In-Ceram material or along the Vitadur Alpha – In-Ceram interface. The angle of incidence at the cone crack-interface junction appeared to be consistent and did not appear to influence the extent or direction of the crack after it met the interface. The present observations show both responses can occur in the crowns, which is probably a reflection of the diverse nature of the stress distribution around the crown whose shape presents such a complex non-planar structure.

From the optical observations and the chart recordings, the sequence for failure in the crowns designated as type I appeared be:

1. An initial circular crack
2. The circular crack developed into a cone crack
3. The area within the cone crack developed multiple ring/cone cracks
4. The centre of the contact area crushed under inelastic deformation, but the crush zone did not extend into the In-Ceram core material.
5. Radial cracks then developed through the core material and the crown failed completely.

The sequence for the type II failures appeared to be similar although the presence of delaminations and random crack pattern replaced the very distinct radial cracks. The diagram in figure 3.26 demonstrates the relevant features.

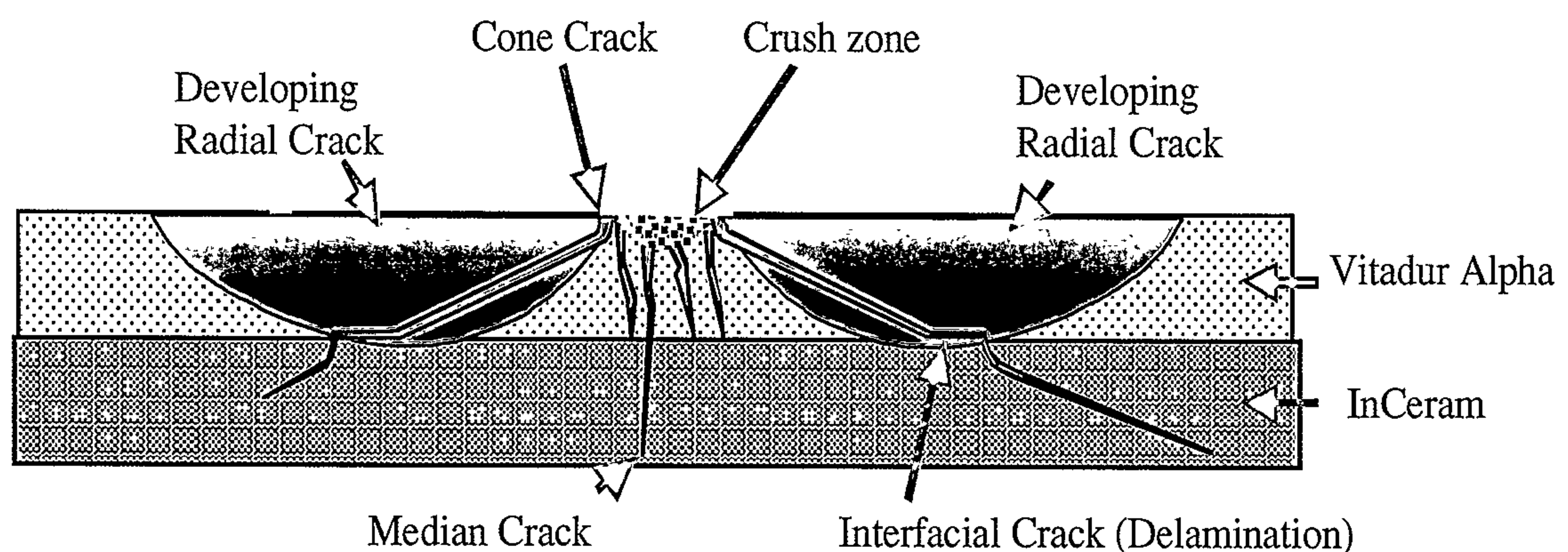
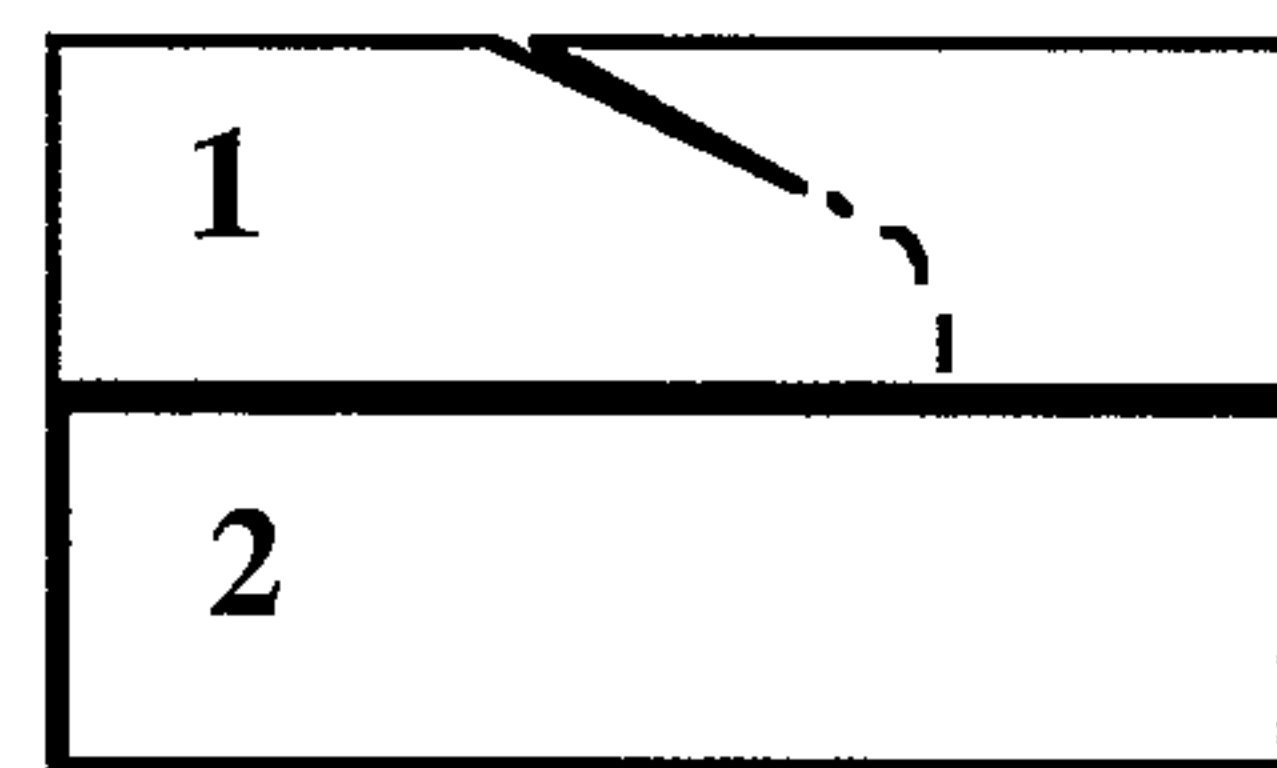


Figure 3.26 Diagram for the different types of damage occurring in the bilaminar crown structure during the tests.

The latter situation is not unlike that described by Davis et al., (1991). Figures 3.18 and 3.19a confirm the cone crack and its deflection at and extension into the interfacial layer as described in section 2.5.4. There was no distinction in this early phase of testing between the onset of the major ring crack and any initial surface ring cracking. The major ring crack was noted as the source of the developed Hertzian cone crack. The radial cracks continue through the interface and the core. This type of failure is described by Mencík (1996) as that which occurs when a hard brittle coating is placed on a hard substrate. The example Mencík used was a glass enamel on alumina, which is not dissimilar to an In-Ceram crown.

The Hertzian cone cracks tended to travel along the interface and occasionally branch down into the core layer. According to Mencík (1996) and He et al. (1989 a&b) the role of fracture toughness for the three possible pathways helps determine the crack direction. The three fracture toughness values are those for the veneer layer, the core material and the toughness of the interface. From Mencík (1996) we also learn that Young's Modulus has a role in determining the direction of crack propagation. There is the possibility that the crack will deflect into the underlying core structure when the first material is stiffer than the underlying material and away from the interface when the underlying material is stiffer than the first. Figure 3.27 demonstrates a diagram of this situation. This particular example has the crack loaded in tension. There was some evidence that this mismatch may have influenced delamination.

$$E1 > E2$$



$$E1 < E2$$

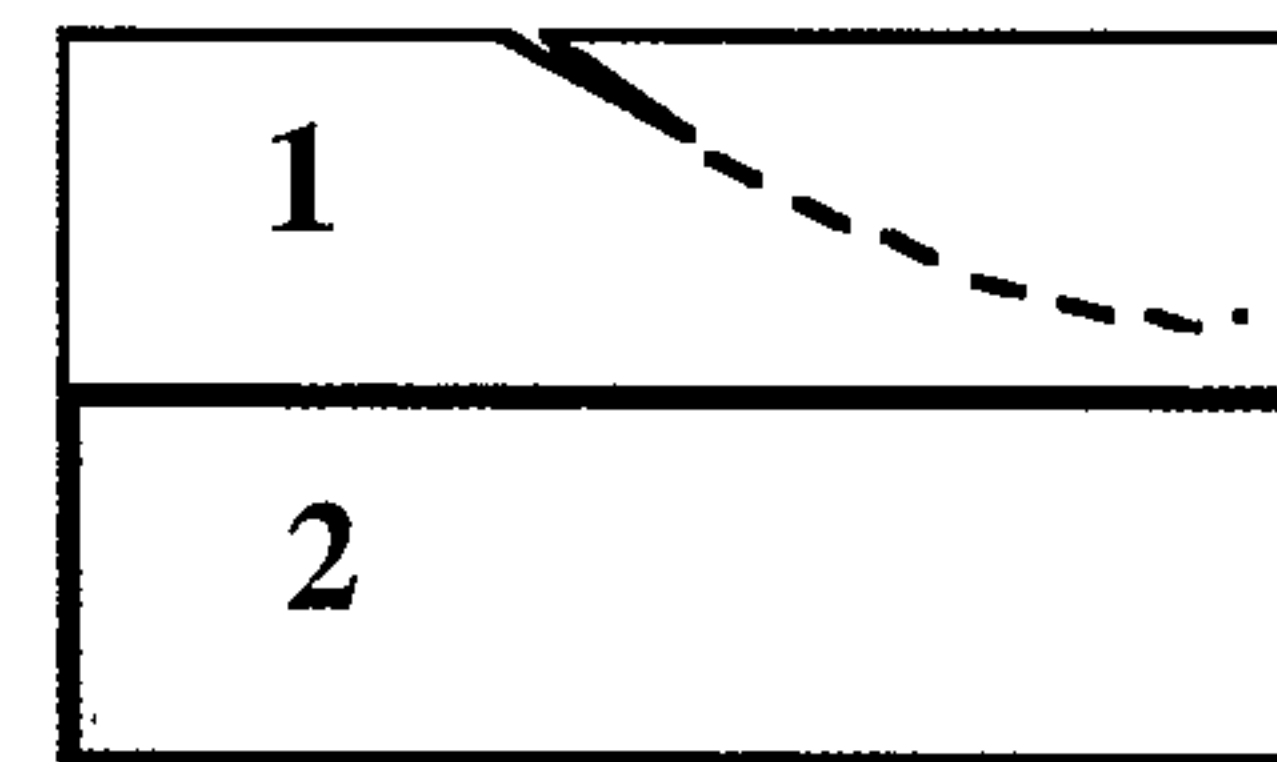


Figure 3.27 Diagram for possible crack deflection based on differences in Young's Modulus (E). From Mencík (1996).

The other factors involved for determining the crack direction are the presence of other stresses such as shear forces and the nature of the initial driving stress e.g. tensile and the direction of application relative to the interface. In the ideal situation, this tensile force would be perpendicular to the interface crack.

The articles by He et al., (1989b) and Davis et al., (1991) also investigate the various possibilities for crack propagation / deflection, particularly where the deflection can be along the interface. This is of limited use in dental crowns, because partial delamination of the veneer layer is invariably accompanied by an increase in reflectance from this point, causing a marked change in the appearance of the crown. The clinical result is therefore usually classed as a failure. In the short term it does provide an energy release system which may prolong the useful life of the crown. They further conclude that the more oblique the angle, the more favourable are the conditions for the crack to deflect along the interface. The clinical crown complicates this further because it presents an interfacial plane that travels through three dimensions, continually changing angle. The crack could then move away from the interface and back through the weaker veneering porcelain

without changing direction. The result would be the shearing of a layer of veneer porcelain as demonstrated by figure 1.5.

The loads for fracture of these crowns were in reasonable agreement with Pröbster (1992) where he recorded the mean fracture load for In-Ceram crowns as 964 N and Yoshinari et al., (1994) where the value was 1060 N. The slightly lower result for Pröbster was probably due to the test method where a 2.5 mm radius rounded chisel, was used to indent the palatal surface rather than a 5 mm diameter spherical indenter. The tests by Yoshinari used a 4.8 mm diameter stainless steel ball. The crowns in his test all failed in the veneer layer. The rounded chisel was placed across the palatal surface and presented the opportunity for a stress concentration to occur on a marginal ridge with little or no core support beneath. Pröbster (1992) gives a value of 1494.1 N as the mean value for failure of the veneer layer on metal-ceramic crowns used in his test series.

Failure loads tended to be higher than the maximum loads given for forces between two antagonistic teeth, which range between 245 N and 540 N (Pröbster 1992). This was possibly due to contact radii, where Auerbach's Law ( $P=AR$ ), provides an explanation for the lower loads. Some failure loads were close to this range which gives some credibility to the model. The difference between compliance of the model and the compliance of a crown on a tooth is not disputed. The model represents a rigid situation, which can be argued to present a worse case scenario. In the paper by Rhee et al., (2001) it was noted that failure due to a more compliant supporting base tended to induce large and destructive radial cracks. Their case used polycarbonate (Young's Modulus 2.3 GPa) as the support base. The Young's Modulus of dentine is approximately 10 times greater at 20 GPa.

Finally, the clinical failure examples from chapter one and the in vitro test crowns did not demonstrate any evidence that the fractured crown bilaminar collapsed down into the underlying layer, i.e. the dentine in the clinical examples. This has been shown to occur in industrial examples where hard surface layers crush into a ductile substrate (Weppelmann et al 1996). The cement layer may have deformed enough that the critical strain limit for the underlying core ceramic was finally exceeded and cracks normal to the surface appeared causing the crown to fail catastrophically. The issues of stresses developed throughout a loaded bilaminar simple model of the more complex crown are discussed in the following chapters.

### 3.5 Conclusions.

The conclusions that can be drawn are of clinical value given that any major crack through the system will result in either an aesthetic or functional failure. The aesthetic failure is where a section of the veneering material shears away from the core and leaves that section

intact, at least protecting the pulp. That is, the underlying core material can provide sufficient resistance to deformation from the load and protect the tooth. The functional failure could also be from loss of the veneer or from complete failure of both layers.

The loads for crown failure are reasonably representative of the clinical situation and are less than those for metal ceramic crowns. The simple model developed has provided a critical load for failure of these bilaminar crowns. There were two distinct types of fracture pattern, designated type I and II. The mean load for type I fractures was 1066 N and for type II fractures the load was 1273 N. This is explained in terms of Auerbach's Law where the load for crack initiation is proportional to the radius of the indenter. Therefore the results can serve as a guide for expected loads on simpler models that could be used to evaluate the initial stress distributions in the ceramic layers. The failure modes are similar to the limited clinical observations mentioned in Chapter 1.

It would be useful to investigate further design modifications to prevent or reduce the risk of the initial crack process. This is indicated from the correlations derived from the data, which indicate a significant difference in crack pattern, where the core to veneer thickness ratio approaches 2:1.

The Weibull modulus is an indicator for a new design approach for this type of crown. The difficulty with this model is how to quantify the observations, relate them to the physical properties of the materials and use that information to modify the design.

In the following chapters a simplified planar bilaminar model is investigated. The model enables the role of the porcelain (Vitadur Alpha) to ceramic core material (In-Ceram) to be addressed. One of the primary considerations for this model is to ensure that the dimensions are reasonably consistent with those found in crowns and that they also comply with the ISO standard for dental ceramic. (ISO 6872: 1995). The model will also be used to develop strengthening concepts based on variation of the core to veneer thickness ratios and how the underlying support for the core affects the crack resistance. The first of these chapters investigates how the bilaminar system behaves with no support and the subsequent chapter looks at the behaviour when the system has full support. These are the extreme situations outside the opposite ends of the clinical situation where the ceramic is supported by dentine with a Young's modulus of 20 GPa to support by a hard cast gold or even cobalt-chrome core with Young's moduli between 100 and 220 GPa.