DOKUZ EYLÜL UNIVERSITY
GRADUATE SCHOOL OF NATURAL AND APPLIED SCIENCES

OPTIMIZATION OF HUMAN TOOTH CROWN IN TERMS OF STRESSES OCCURED AFTER FIRING PROCESS

by
Yusuf ARMAN

August, 2008
İZMİR

OPTIMIZATION OF HUMAN TOOTH CROWN IN TERMS OF STRESSES OCCURED AFTER FIRING PROCESS

A Thesis Submitted to the
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Ph.D. THESIS EXAMINATION RESULT FORM

We have read the thesis entitled **“OPTIMIZATION OF HUMAN TOOTH CROWN IN TERMS OF STRESSES OCCURED AFTER FIRING PROCESS”** completed by **YUSUF ARMAN** under supervision of **PROF. DR. SAMİ AKSOY** and we certify that in our opinion it is fully adequate, in scope and in quality, as a thesis for the degree of Doctor of Philosophy.

……………………………………………

Prof. Dr. Sami AKSOY

Supervisor

……………………………………………

Assoc. Prof. Mehmet ZOR

Thesis Committee Member

……………………………………………

Prof. Dr. Celal ARTUNÇ

Thesis Committee Member

……………………………………………

Prof. Dr. Mahmut ÖZBAY

Examining Committee Member

……………………………………………

Prof. Dr. Onur SAYMAN

Examining Committee Member

Prof.Dr. Cahit HELVACI
Director
Graduate School of Natural and Applied Sciences

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I am very grateful to my wife and my daughter for their understanding, support and love. They worked very hard to support me all over the past years. For this reason, this dissertation is dedicated to them.

Yusuf ARMAN

OPTIMIZATION OF HUMAN TOOTH CROWN IN TERMS OF STRESSES OCCURED AFTER FIRING PROCESS

ABSTRACT

In this study, it has been purposed to determine the more harmonious material pairs in human tooth crowns which are used for prosthetic treatments and constituted from the different substructure and veneering materials in terms of stresses occurred after firing process. It has been used the most preferred material pairs in dentistry at the present time.

Firstly, elastic and plastic region behaviors in the high temperatures of three of the investigated metal alloys have been determined by the hot compression tests. The other properties of the investigated materials have been obtained from the literature studies or documents of the producing firms. Also, it has been developed the calculations that the viscoelastic behaviors effects of the porcelain in high temperatures have been included into the elastic properties. the temperature variations of the inner and outer surfaces been measured after the standard firing process, preparing a crown specimen relating to central incisor tooth of human and then the cooling curves of the inner and outer surfaces has been determined. With the assistance of these cooling curves obtained, the equivalent convection coefficients include the effects of radiation and convection together, have been calculated.

The three dimensional (3D) model of the test specimen prepared has been constituted, using ABAQUS finite element program. The equivalent convection coefficients calculated from the experiments have been entered into the finite element program, and then the cooling curves obtained from the analyses have been fitted to the experimental cooling curves, making the minor revisions on these coefficients. In this way, the atmosphere in the furnace has been determined exactly in the finite element program.

Afterwards, the analyses of thermal stresses which have been constituted during the cooling process have been done for the different material couples. The results have been evaluated in terms of the maximum stresses and the residual stresses.

Keywords: Dental crown, equivalent convection coefficient, residual stress.

FIRINLAMA İŞLEMİ SONRASI OLUŞAN GERİLMELER AÇISINDAN İNSAN DİŞİ KAPLAMALARININ OPTİMİZASYONU

ÖZ

Bu çalışmada, protetik tedavi amacıyla kullanılan ve farklı alt ve üst yapı malzemelerinden oluşturulan insan dişi kaplamalarında, fırınlama işlemi sonrası meydana gelen ısıl gerilmeler açısından daha uyumlu malzeme çiftlerinin belirlenmesi amaçlanmıştır. Günümüzde, diş hekimliğinde en çok tercih edilen malzeme çiftleri kullanılmıştır.

Çalışmada öncelikle, incelenen metal alaşımlarının üçüne ait yüksek sıcaklıklardaki elastik ve plastik bölge davranışları sıcak basma testleriyle belirlenmiştir. İncelenen malzemelerin diğer özellikleri literatür çalışmalarından veya üretici firmalardan elde edilmiştir. Ayrıca porselenin yüksek sıcaklıklardaki viskoelastik davranışlarının etkilerinin, elastik özellikler içine dahil edildiği hesaplamalar geliştirilmiştir. İnsan üst ön dişine (santral diş) ait bir kaplama numunesi hazırlanarak standart pişirme işlemi sonrasındaki iç ve dış yüzeylerin sıcaklık değişimleri ölçülmüş ve soğuma eğrileri çıkarılmıştır. Elde edilen bu soğuma eğrileri yardımıyla ışınım ve taşınım etkilerini birlikte içeren eşdeğer taşınım katsayıları hesaplanmıştır.

ABAQUS sonlu elamanlar paket programı kullanılarak deney numunesinin üç boyutlu modeli kurulmuştur. Deneysel çalışmalar yardımıyla hesaplanan eşdeğer taşınım katsayıları modele uygulanmış, bu katsayılar üzerinde yapılan küçük seviyelerdeki düzeltmelerle analizlerden elde edilen soğuma eğrileri, deneysel soğuma eğrileriyle uygun hale getirilmiş ve bu şekilde ortam tam olarak tanımlanmıştır.

Daha sonra tanımlanan bu ortam şartlarında farklı malzeme çiftleri için soğuma sırasında oluşan ısıl gerilmelerin analizleri yapılmıştır. Sonuçlar maksimum gerilmeler ile artık gerilmeler açısından değerlendirmiştir.

Anahtar Kelimeler: Dental kaplama, eşdeğer taşınım katsayısı, artık gerilme.

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CHAPTER ONE
INTRODUCTION

1.1 Introduction

It is no doubt that teeth are very considerable in terms of human health. Nevertheless, generally, teeth get deformed in parallel to the aging of the body. Under the circumstances, the teeth restore in order to continue its functions. First and foremost, it must be guaranteed that the materials used in the tooth are conformable both biologically and chemically. Besides, the main object is to achieve a tooth crown that will be used without corrosion and failure, as far as possible; since it is a highly priced procedure to replace these crowns. Additionally, some factors such as strength, aesthetics, ease of manufacture and cost are reasons for predilection.

The dental hygienists are principally concerned with the materials used for restoring to carious lesion, by either assisting the dentist in their manipulation or observing their behavior in the oral cavity. Materials used for an enduring restoration are purposed to satisfy the objectives of the restoration for periods of 20 to 30 years. An ideal restoration would be one that would last as long as the tooth. The metallic restorative materials e.g. amalgam, gold inlays, and crown, and direct filling gold, have the physical properties with greatest potential to meet the demands that are placed upon the restoration.

It is generally accepted that ceramic restorations exhibit a perfect aesthetic qualities. However, the mechanical insufficiencies of such materials include their inherent brittleness and potential to abrade the opposing dentition. Recent developments have attempted to overcome such disadvantages by either the use of increasingly complex technology or by the simplification of existing techniques and/or materials. The diversity of dental ceramics continues to stimulate laboratory and clinical research (Qualtrough & Piddock, 1997).

In general, restorative dental materials are developed by the producer, and selected by the dentist, on the basis of characteristics mechanical and physical properties of the material as well as its biological and chemical reactivity and the tissue response in function. The tissue response to the restoration often results from a combination of the physical, chemical, and mechanical properties of the completed structure. No single property can be used as a measure of quality in all structures of materials. Often several combined properties, determined from standardized laboratory and service tests are employed to give a measure of quality of the material.

A complete evaluation of a new or improved restorative material may require the combined efforts of the dentists, material scientist and biologist. As a result, orderly study of the properties of restorative materials has developed into a special branch or bioengineering. The application of bioengineering principles to restorative materials will expend in the forthcoming years.

1.2 Dental Crowns in Dentistry

Tooth crown or dental crown is the permanent dental restoration of the teeth using materials that are fabricated by indirect methods. This dental crown is cemented to your tooth to cap or completely cover it.

Traditionally, the teeth to be crowned are prepared by a dentist and records are given to a dental technician to fabricate the crown or bridge, which can then be inserted at another dental appointment. The main advantages of the indirect method of tooth restoration include:

- Fabrication of the restoration without the need for having the patient in the chair
- The utilization of materials that require special fabrication methods, such as casting

- The use of materials that require intense heat to be processed into a restoration, such as gold and porcelain.

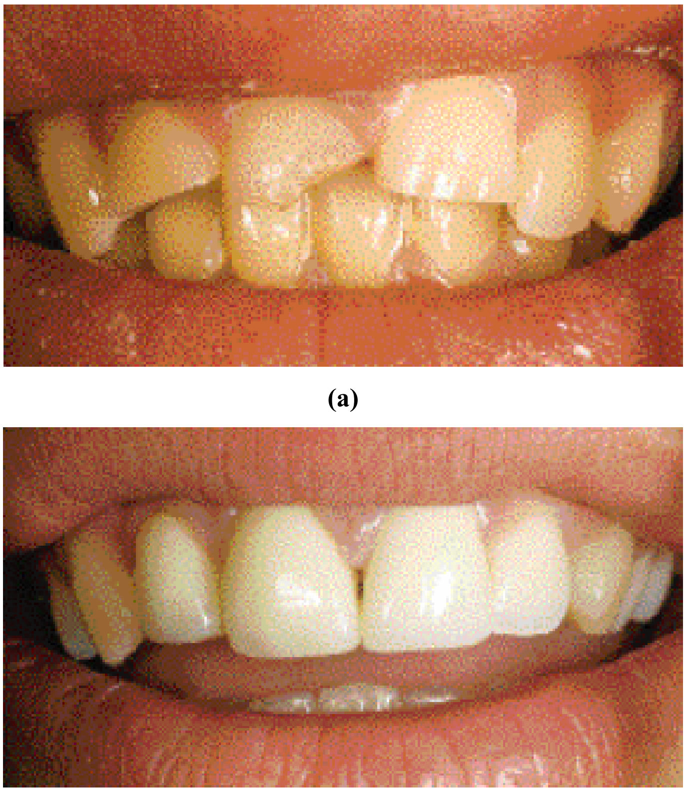
The restorative materials used in indirect restorations possess superior mechanical properties than do the materials used for direct methods of tooth restoration, and thus produce a restoration of much higher quality. As new technology and material chemistry has evolved, computers are increasingly becoming a part of crown and bridge fabrication, such as in CAD/CAM technology.

1.2.1 Why do teeth need dental crowns?

A dentist might recommend placing a dental crown for a variety of reasons but, in general, most of these reasons will usually fall within one of the following basic categories:

- *Dental crowns are used to restore a tooth's shape and strength*

Since a dental crown that has been cemented into place essentially becomes the new outer surface for the tooth, it is easy to imagine how the placement of a crown can restore a tooth to its original shape. It's also easy to see how a dental crown can help to strengthen a tooth by way of being a hard outer shell that encases the tooth structure that lies within it. For both of these reasons, dental crowns are routinely made for teeth that have broken, worn excessively, or else have had large portions destroyed by tooth decay (Figure 1.1 a, b).



(a)

(b)

Figure 1.1 A view of a crowned tooth; **(a)** before crown, **(b)** after crown.

It is conceivable that a dental filling, as an alternative, could be used as a means to restore a tooth’s shape. Dental crowns however offer your dentist a big advantage over dental fillings by way of the fact that they are fabricated “away from your mouth”. By this we simply mean that dental crowns are fabricated in a dental laboratory. Dental fillings, in comparison, are created “in your mouth” by way of your dentist placing the filling material directly upon your tooth.

When a dental crown is made the dental laboratory technician can visualize and examine all aspects of your bite and jaw movements, from a variety of angles, and then sculpt your dental crown so it has the perfect anatomy. In comparison, when a dentist places a dental filling they have far less control over the final outcome of the shape of your tooth because it is often difficult for them to visualize, evaluate, and access to the tooth on which they are working.

From a standpoint of strength considerations, there are some types of filling materials that can bond to tooth structure. For the most part, however, dental fillings are not considered to substantially strengthen a tooth in the same way that a dental crown, with its rigid encapsulation a tooth, can.

- *Porcelain dental crowns are used to improve the cosmetic appearance of teeth*

Since a dental crown serves to cup over and encase the visible portion of a tooth, any dental crown that has a porcelain surface can be used as a means to idealize the cosmetic appearance of a tooth. Possibly you have heard it rumored (especially in past decades) that certain movie stars have had their teeth “capped”. This simply means that the person has obtained their good smile by way of having dental crowns placed.

Actually, getting your teeth “capped” just to improve their cosmetic appearance can at times be a very poor choice. Dental crowns are best utilized as a way to improve the cosmetic appearance of a tooth when the crown simultaneously serves other purposes also, such as restoring a tooth to its original shape (repairing a broken tooth) or strengthening a tooth (covering over a tooth that has a very large filling).

In general, a dental crown probably should not be used as a means to improve the appearance of a tooth if there is any other alternative dental treatment that could equally satisfactorily achieve the same cosmetic results. This is because a dentist must grind a significant portion of a tooth away when a dental crown is made. If a more conservative dental procedure could equally well improve the tooth’s appearance, such as a porcelain veneer, dental bonding, or even just teeth whitening, then it is usually best to consider that treatment option first.

1.2.2 Types of dental crowns

Dental crowns (also known as “dental caps” or “tooth caps”) can be made from metal (gold or other metal alloys), ceramic materials (such as porcelain), or a combination of both which is known as porcelain-fused-to-metal (PFM).

- ***All metal dental crowns/Gold dental crowns***

Some dental crowns are made entirely of metal. The classic metal dental crown is one made of gold (Figure 1.2), or more precisely a gold alloy. Over the decades a variety of different metal alloys have been used in making dental crowns. Some of these metals are silver in color rather than yellow like gold.

Figure 1.2 An example of all metal dental crowns: Full gold crown.

Having a gold dental crown made can be an excellent choice. Here are some reasons why:

a. Because of its physical properties, dentists find gold to be a very workable metal. This characteristic helps a dentist to be able to achieve a very precise fit with the crown.

b. Since they are metal through and through, gold crowns withstand biting and chewing forces well. They will not chip. It would be uncharacteristic for a gold crown to break. Of all of the types of dental crowns, gold crowns probably have the greatest potential for lasting the longest.

c. Although they are very strong, the wear rate of a gold crown is about the same as tooth enamel. This means that a gold dental crown won't create excessive wear on the teeth it opposes (the teeth it bites against).

Metal dental crowns are usually placed on those teeth that are not overly visible when a person smiles (i.e., molars). If you are considering a gold crown take our advice on this point, before you give your dentist the go ahead on making the crown check with your spouse first. They're the one who will be looking at your smile and your new dental crown the most.

- ***Full-porcelain dental crowns***

Some dental crowns are fabricated in a manner where their full thickness is porcelain (dental ceramic). These crowns can possess a translucency that makes them the most cosmetically pleasing of all of the different types of dental crowns.

Although they can be very life like in appearance, the overall strength of all-porcelain dental crowns is less than other types of crowns. While they can be a good choice for front teeth, due to the hefty chewing and biting forces that humans can generate, all-porcelain dental crowns may not be the best choice for back teeth.

- ***Porcelain-fused-to-metal dental crowns***

Porcelain-fused-to-metal (PFM) dental crowns are somewhat of a hybrid between metal crowns and porcelain crowns. When they are made the dental technician first makes a shell of metal that fits over the tooth. A veneering of porcelain is then fused over this metal (in a high heat oven), giving the crown a white tooth-like appearance. The correct porcelain coverage design is depicted in Figure 1.3. Depending on the requirements of your situation, these crowns are sometimes made where the porcelain veneer only covers those aspects of the crown that is readily visible (meaning the other portions of the crown have a metal surface). In other cases these crowns are pretty much fully surfaced with porcelain.

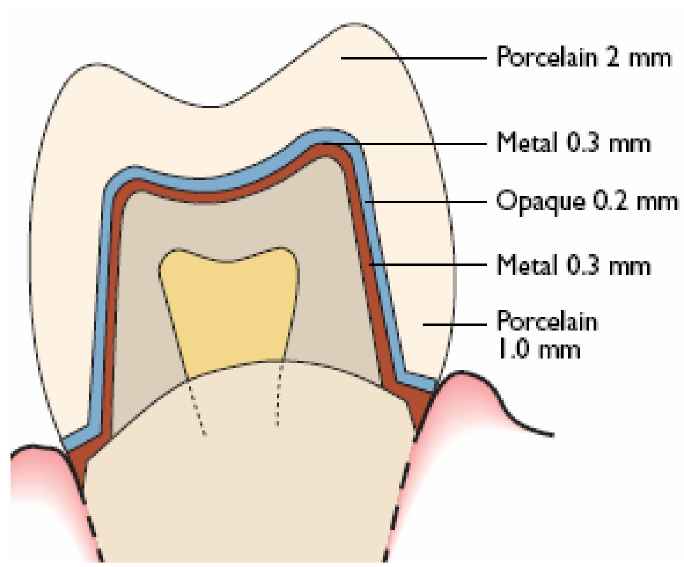


Figure 1.3 Correct porcelain coverage design for PFM crowns.

PFM dental crowns can be a good choice for either front or back teeth. These crowns are strong enough to withstand heavy biting pressures and at the same time can have an excellent cosmetic appearance. There are some disadvantages associated with PFM crowns however (which no doubt your dentist will try to minimize as much as is possible). They are:

a. While the cosmetic appearance of these crowns can be excellent, they often are not as pleasing aesthetically as all-porcelain dental crowns.

b. The crown’s porcelain can chip or break off.

c. The porcelain surface of the crown can create wear (sometimes this wear is significant) on those teeth that it bites against.

d. The metal that lies underneath a crown’s porcelain layer can sometimes be visualized as a dark line found right at the crown’s edge. A dentist will usually try to position this dark edge just underneath the tooth's gum line but if a person's gums recedes this dark line can show, thus spoiling the crown's appearance.

1.3 Preparing and Applying of Dental Crowns

A dental crown is a cap. Crowns can give support to misshapen or badly broken teeth and permanently replace missing teeth to complete a smile or improve a bite pattern. They may be molded from metal, ceramic, or combinations of all of them. They are cemented in place and coated to make them more natural looking. With regard to history, a variety of materials have been used as tooth replacements. The ancient Egyptians used animal teeth and pieces of bone as primitive replacement materials. More recently, artificial teeth have been fabricated from substances such as ivory, porcelain, and even platinum. With modern technology, high quality tooth replacements can be made from synthetic plastic resins, ceramic composites, and lightweight metal alloys.

1.3.1 Main parameters in design of dental crowns

There are several key factors to consider in the design of dental crowns. First, appropriate raw materials with which to make the crown must be identified. These materials must be suitable for use in the oral cavity, which means they must be acceptable for long term contact with oral tissues and fluids. Crown components must have a good safety profile and must be non-allergenic and non-carcinogenic. The American Dental Association/ANSI specification 41 (Biological Evaluation of Dental Materials) lists materials which have been deemed safe for use. In addition to safety considerations, these materials must be able to withstand the conditions of high moisture and mechanical pressure, which are found in the mouth. They must be resistant to shrinkage and cracking, particularly in the presence of water. Metal is preferred for strength but acrylic resins and porcelain have a more natural appearance. Therefore the selection of crown material is, in part, dependent on the location of the tooth being covered. Acrylic and porcelain are preferred for front teeth, which have higher visibility. Gold and metal amalgams are most often used for back teeth where strength and durability are required for chewing but appearance is less critical.

The second factor to consider when designing a crown is the shape of the patient's mouth. Dental restorations must be designed to mimic the bite properties of the original tooth surface so the wearer does not feel discomfort. Since every individual's mouth is different each crown must be custom designed to fit perfectly. Successful crown design involves preparation of an accurate mold of the oral cavity (Ellison, 1980; Geering et al., 1993; Goldstein, 1997; Woodforde, 1968).

1.3.2 Basic materials of crown construction

There are four main types of materials used in crown construction: The plasters used to create the mold, the materials from which the crown itself is made (e.g., metal, ceramic, plastic), the adhesives used to cement the crown in place, and the coatings used to cover the crown and make it more aesthetically appealing.

- ***Molding plasters***

Plaster molds are made from a mixture of water and gypsum powder. Different types of plasters are used depending on application: impression plaster is used to record the shape of the teeth, model plaster is used to make durable models of the oral cavity, and investment plaster is used to make molds for shaping metal, ceramics and plastics. Waxes are also sometimes used in this regard.

- ***Crown fabrication materials***

Because of good hardness, strength, stiffness, durability, corrosion resistance, and bio-compatibility, metals are frequently used in crown fabrication. Common alloys used in crowns are based on mixtures of mercury with silver, chromium, titanium, and gold. These mixtures form a blend than can be easily shaped and molded, but which hardens in a few minutes.

Ceramics are well suited for use in crowns because they have good tissue compatibility, strength, durability and inertness. They can also be made to mimic the appearance of real teeth fairly closely. However, the tensile strength of ceramic is low enough to make it susceptible to stress cracking, especially in the presence of water. For this reason, ceramic is most often used as a coating for metal-structured crowns. The two primary types of ceramics used in crowns are made from potassium feldspar and glass-ceramic.

The first resin used in denture materials was vulcanized rubber in 1839. Since then, a number of other resins have been developed which are more suitable for dental applications. Today, acrylic polymer resins are commonly used in dentures and crowns. Specifically, polymethyl methacrylate is most often used. Some of these resins harden at room temperature as this reaction progresses. Others require heat or ultraviolet light to catalyze the change.

Special dental adhesives, or dental cements, are used to hold the crown in place. These can be classified as either aqueous or nonaqueous. The aqueous types include zinc phosphates, polycarboxylate cements, glassionomer cements, and calcium phosphate cements. The nonaqueous types include zinc oxide-eugenol, calcium chelates, and acrylic resins such as polymethyl methyacrylate.

Coatings are used to make the crown appear more natural. Porcelain is used in this regard, but it is difficult to work with and hard to match to the tooth’s natural color. Resins similar to the ones used in tooth construction are also used to create tooth-colored veneers on crowns. These resins have an advantage over other veneers in that they are inexpensive, easy to fabricate, and can be matched to the color of tooth structure. However, acrylic coatings may not adhere to the crown’s surface as well as porcelain or other materials. Therefore, the prosethedontist may design the crown’s surface with mechanical undercuts to give the coating a better grip. Resin coatings also have relatively low mechanical strength and color stability and poor abrasion and stain resistance as compared to porcelain veneers (Ellison, 1980; Geering et al., 1993; Goldstein, 1997; Woodforde, 1968).

1.3.3 Fabrication process of dental crowns

- ***Designing the mold***

a. Before beginning, the dentist may need to prepare the area where the crown is to be installed. This may require the removal of 2-3 millimeters of tooth structure from the four sides and the biting edge, as seen in Figure 1.4.

Figure 1.4 Removing of tooth structure from the four sides of tooth.

Then, an impression of the tooth is taken to record its shape. This step uses impression plaster which is the softest and fastest setting type of dental plaster. The impression plaster is mixed with a small amount of water until it is fluid. This slurry is placed in a tray that is fitted over the teeth. The tray is held still in place until the plaster hardens. When the tray is removed from the mouth, it retains a three dimensional impression of the tooth that is to be covered. This impression is a negative, or reverse, image of the tooth.

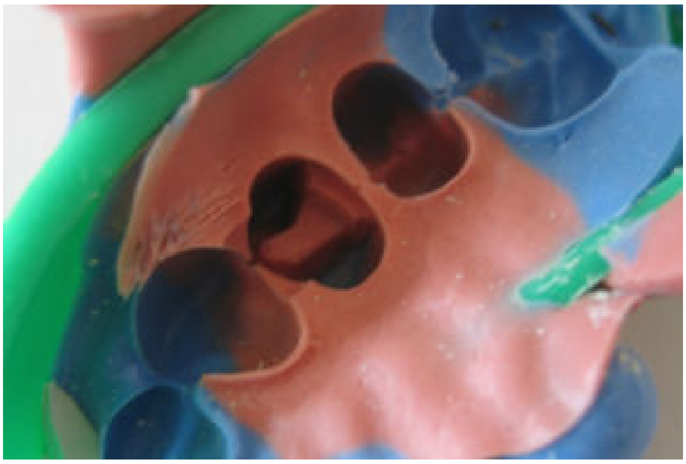


Figure 1.5 A view from impression mold of a tooth prepared for restoration.

b. The next step is to prepare another type of plaster, known as model plaster. This type of plaster is harder than the impression plaster. Once again the plaster is mixed with the appropriate quantity of water. Then the slurry is poured into the impression mold. In this way, a positive model of the tooth can be made. This positive model made from the negative impression mold is called a cast. The cast is used by the dentist for study purposes.

Figure 1.6 Occlusal view of a plaster model of the lower arch.

c. The impression is also used to make a mold, called an investment, which is capable of withstanding high temperatures. This is an important consideration because some metals and ceramics require temperatures higher than 1300°C for molding. These investments are made from calcium phosphate mixed with silica and other modifying agents.

- ***Casting of the crown***

Manufacturing of the crown is done by filling the investment with the appropriate material. In the case of metals, this is done at a high temperature so the metal is molten. For ceramics and plastics, the mixture is initially fluid but may require the addition of heat to cause the materials to cure and harden. A vertical vise may be used to help pack the casting investment tightly. The process also requires the mold first be treated with a release agent to ensure the crown can be easily removed after it has hardened. Some acrylic resins must be heated for up to eight hours to make sure they are fully cured. After the processing is done and the investment has cooled, the mold is broken apart and the crown is removed.

Figure 1.7 A view from end of casting process of dental metal.

- ***Applying the porcelain on the crown***

After the crown has been successfully completed from the mold, it is ready for porcelain installation. The prosethedontist applies cement to the inside of the crown surface and then fits it into place over the tooth. Because of the number of processing steps there may be a slight discrepancy in the fit and the crown may require minor grinding and smoothing of its surface to ensure it fits correctly.

- ***Finishing process***

Some crowns may require a finishing coat to seal it and improve its natural appearance. Such coatings are typically acrylic polymers. The polymer can be painted on as a thin film, which hardens to a durable finish. Some polymers require a dose of ultraviolet light to properly cure (Ellison, 1980; Geering et al., 1993; Goldstein, 1997; Woodforde, 1968).

1.4 Future of Dental Restorations

Dental technology is constantly advancing and these improvements are already finding application in dental crown manufacturing. State of the art crowns can be made with an industrially produced core made of densesintered ceramic, and an outer layer of porcelain is added by hand. This futuristic crown material is made by an advanced Computer Aided Design (CAD) process, known as Procera process, which was introduced in the mid-1990s in Switzerland. This process results in crowns with improved strength and optimal fit. Unlike other crown materials, crowns made by the Procera process can be used anywhere in the mouth due to the strength of its core material and its more natural appearance. Another advance in crown technology involves pre-made and pre-sized stainless steel crowns, which are designed as generic tooth replacements. Usage of this new type of crown is very simple: first, the tooth surface is prepared then the selected crown is cemented in place with a standard stainless steel crown adhesive. The crown can be crimped or cut to fit and the epoxy finish will not chip or peel. While this new technology offers increased simplicity, it does not give the same appearance as a custom made crown. Other future advancements are likely to come from new resins, which have improved adhesion in the high moisture environment of the oral cavity.

CHAPTER TWO
DENTAL MATERIALS USED TODAY

2.1 Introduction

Today, most patients opt for crowns that match the color of the rest of their teeth. Some people, however, still prefer to have gold or metal crowns placed over their teeth instead of the more natural looking porcelain varieties. Generally, most patients prefer the most esthetic and strongest restorations possible. That generally means porcelain fused to a metal, alumina or zirconium substructure for back teeth, and all-porcelain crowns for the front teeth. The three types of substructures mentioned here lend greater strength to the porcelain so that chewing and bruxing pressures are not as likely to break the porcelain on the back teeth. On front teeth where chewing and bruxing forces tend to be less severe, all-porcelain crowns are generally of sufficient strength and are somewhat more esthetic than porcelain fused to metal. The newer alumina and zirconium substructures are relatively opaque, but can be fabricated in the same colors as the porcelain overlying them. They can provide nearly the strength of an all metal substructure without the esthetic liability. All-metal crowns are the strongest configuration anywhere in the mouth since they can’t break under pressure. Porcelain crowns with a zirconium or a glass infused alumina substructure are almost as strong as porcelain fused to metal and are almost as esthetic as all porcelain crowns. All-gold crowns have the advantage of being hypoallergenic while base metal crowns have the advantage of being somewhat less expensive than either porcelain or gold crowns. The various types of materials used to fabricate crowns are given below.

2.2 Substructure Materials

Substructure materials are made by fabricating a hollow mold, pouring a molten metal into it, allowing the metal to solidify and separating the now solid metal casting from the mold. Ultimately, all metallic objects originate from castings. In dentistry, substructure materials are used to restore teeth, replace teeth, and as frameworks for removable partial dentures. Today, substructure materials are also used as metal frameworks to support porcelain crowns or fixed partial dentures in order to produce strong and yet very esthetic restorations (A course in dental alloys, nd).

2.2.1 Gold alloys used for all-gold restorations

Gold based alloys are virtually the only castable alloys used in dentistry. There are four types:

Table 2.1 Types of gold based alloys

Type	Hardness	Yield Strength (MPa)	Percent Elongation
I	Soft	<140	18
II	Medium	140-200	18
III	Hard	201	12
IV	Extra-Hard	>340	10

Type I is hard enough to stand up to biting forces, but soft enough to burnish against the margins of a cavity preparation. It is used mostly for one surface inlays.

Type II is less burnishable, but hard enough to stand up in small multiple surface inlays that does not include buccal or lingual surfaces.

Type III is the most commonly used type of gold for all-metal crowns. It is still used whenever a patient requests an all gold restoration such as an all gold crown, inlay or onlay. A typical Type III gold alloy has approximately 75% gold, 10% silver, 10% copper, 3% palladium and 2% zinc. The purpose of each component is as follows:

- Gold is a noble metal. In other words, it resists tarnish and corrosion and will participate in very few chemical reactions, which means that it is non toxic and hypoallergenic. It is also highly ductile and malleable and has a relatively low melting point, which major factors are accounting for its use by people in early historical periods. Gold's long civilizational lineage and incorruptibility made it a natural first choice for use in dentistry. It forms the bulk of the composition of the alloy.

- Copper is the principal hardener. It is necessary for heat treatment and is usually added in concentrations of greater than 10%.
- Silver lowers the melting temperature and also modifies the red color produced by the combination of gold and copper. It also increases ductility and malleability.
- Palladium (another noble metal) raises the melting temperature, increases hardness and whitens the gold, even in very small concentrations. It also prevents tarnish and corrosion and acts to absorb hydrogen gas which may be released during casting causing porosity.
- Zinc acts as an oxygen scavenger and prevents the formation of porosity in the finished alloy. It also increases fluidity and reduces the surface tension in the molten state improving the casting characteristics of the alloy.

Type IV was used for partial denture frameworks, and was not used in fixed prosthetics.

2.2.2 Porcelain Fused to Metal (PFM) restoration alloys

PFM alloys are classified according to the proportion and types of noble metals they contain.

2.2.2.1 High noble alloys

They have a minimum of 60% noble metals (any combination of gold, palladium and silver) with a minimum of 40% by weight of gold. They usually contain a small amount of tin, indium and/or iron which provides for oxide layer formation which in turn provides a chemical bond for the porcelain. High noble alloys have low rigidity and poor sag resistance. They may be yellow or white in color. There are three general types of High noble alloys:

a. Gold-Platinum alloy developed as a yellow alternative to otherwise white palladium alloys, these can be used for full cast as well as metal-ceramic restorations. More prone to sagging, they should be limited to short span bridges. A typical formula is Gold 85%; Platinum 12%; Zinc 1%; silver to adjust the expansion properties (in some brands).

b. Gold-Palladium alloy can also be used for full cast or metal-ceramic restorations. Palladium has a high melting temperature, and even fairly small amounts of it will impart a white or gray color to the finished alloy. The palladium content reduces the tendency of the casting to sag during porcelain firing. These alloys usually contain indium, tin or gallium to promote an oxide layer. A typical formula is Gold 52%; Palladium 38%; indium 8.5%; Silver to adjust the expansion properties (in some brands).

c. Gold-copper-silver-palladium alloys: These have a low melting temperature and are not used for metal-ceramic applications. They contain silver which can cause a green appearance in the porcelain, and copper which tends to cause sagging during porcelain processing. A typical composition is Gold 72%; Copper 10%; Silver 14%; Palladium 3%.

2.2.2.2 Noble alloys

They contain at least 25% by weight of noble metal. This can mean gold, palladium or silver. Any combination of these metals totaling at least 25% places the alloy in this category. They are the most diverse group of alloys. They have relatively high strength, durability, hardness and ductility. They may be yellow or white in color. Palladium imparts a white color, even in small amounts. Palladium also imparts a high melting temperature.

a. Gold-copper-silver-palladium alloys: Note that this classification is also included under the high noble category. The difference here is that the proportion of gold and palladium is a great deal less than its high noble cousin. More copper and silver are in the mix in its place. These alloys have a fairly low melting temperature, and are more prone to sagging during application of porcelain. Thus they are used mostly for full cast restorations rather than PFM applications. A typical formula is: gold 45%; Copper 15%; Silver 25%; Palladium 5%.

Palladium based alloys offer a less expensive alternative to high noble alloys since they can cost between one half and one quarter as much as the high gold alternative.

b. Palladium-copper-gallium alloys: These are very rigid and make excellent full cast or PFM restorations. They do contain copper and sometimes are prone to sagging during porcelain firing. The gallium is added to reduce the melting temperature of the alloy as a whole. A typical formula is Palladium 79%; Copper 7%; Gallium 6%.

c. Palladium-Silver and Silver-Palladium alloys: As the name(s) imply, the recipes for these alloys vary depending on the relative content of palladium and silver. These were popular in the early 1970’s as a noble alternative to the base metal alloys with which they were designed to compete. Higher palladium alloys are popular for PFM frameworks. Higher silver alloys are more susceptible to corrosion and the silver may lead to greening of the porcelain unless precautions are taken. On the other hand, they have high resistance to sagging during porcelain firing and are very rigid, so they are good for long spans. They are also more castable (more fluid in the molten state), easier to solder and easier to work with than the base metal alloys. Typical recipes include: Palladium 61%; silver 24%; Tin (in some formulas). Another is: Silver 66%; Palladium 23%; Gold (in some formulas, a low percentage of gold was included to satisfy insurance requirements regarding the definition of nobility in the alloy).

2.2.2.3 Base metal alloys

Base metal alloys have been around since the 1970's. They contain less than 25% noble metal, but in actuality, most contain no noble metal at all. They can be used for full cast or PFM restorations, as well as for partial denture frameworks. As a group, they are much harder, stronger and have twice the elasticity of the high-noble and noble metal alloys. Thus castings can be made thinner and still retain the rigidity to support porcelain. They have excellent sag resistance and are great for long span porcelain bridges. They appear at first glance to be the ideal metal for cast dental restorations, and for a while, they were heavily used for PFM frameworks due to their low cost and high strength characteristics.

Unfortunately, Nickel and Beryllium, two of the most commonly used constituents used to make base metal alloys, can cause allergic reactions when in intimate contact with the gingiva. Since many women (and now men) have been sensitized to these metals by wearing inexpensive skin piercing jewelry, crowns and bridges made from these alloys have been known to cause gingival discoloration, swelling and redness in susceptible individuals. Note that the allergic reaction is limited to contact gingivitis and affects the gingiva (gums) alone. There are no known systemic (whole body) allergic reactions reported as a result of exposure to oral appliances made from base metal alloys. Allergic reactions appear to be limited to fixed appliances (crowns and bridges). Nickel containing metals rarely cause allergic dermatitis when used for removable partial denture frameworks.

Base metal alloys also have other disadvantages for the lab technicians and dentists that work with them. They have a very high melting temperature which makes them more difficult to cast. They exhibit a high casting shrinkage (about 2.3%) which must be compensated for. Their hardness makes them difficult to burnish and polish and their high melting temperature makes them difficult to solder. They are also more prone to corrosion under acidic conditions, and finally, the fumes of nickel and beryllium may be carcinogenic during casting and finishing, so lab techs must take special precautions when using these alloys.

Today relatively few labs fabricate fixed restorations made from base metal alloys. The companies that sell dental alloys still carry a small number of these alloys specifically for making crowns and bridges, but not many labs carry them and not too many Western dentists order them. Base metal alloys are heavily used today in the manufacture of removable partial denture frameworks. There are two subcategories of base metal alloy:

a. Nickel-chromium alloys: These contain at least 60% nickel, and may contain a small amount of carbon (about 0.1%) as a hardener. They also can contain either >20% chromium or <20% chromium with or without beryllium. These are used now mostly for removable partial denture frameworks.

b. Cobalt -chromium alloys: These are a nickel free alternative to the nickel-chromium alloys. They seem to have become the most commonly ordered type of base metal for removable partial dentures. They can also be used for PFM framework fabrication as well. The major problem with this formulation is that it is more difficult to work with than the nickel-chromium alloy due primarily to its high melting temperature. This necessitates the use of specialized casting equipment. This alloy's high hardness and low ductility also make it difficult to finish and polish.

2.2.3 Properties of the metals used in dental alloys

a. Gold (Au): Soft, malleable and yellow colored with a low melting point. Looks great, but by itself it lacks sufficient strength to stand up to the forces generated in the oral cavity.

Gold is a noble metal and does not corrode or tarnish in the mouth. The softer alloys are “burnishable”, meaning that the margins can be rubbed with a blunt instrument to seal them and increase marginal adaptation. It is also very kind to the opposing dentition, and will not wear down opposing teeth.

Its native thermal expansion is too high to be used by itself as a base upon which to build a porcelain superstructure. If porcelain were bonded directly to a gold understructure, it would "shiver" and break off the substructure during cooling. This characteristic, however, can be modified by alloying it with other metals.

Finally, since Gold is so inert, it cannot bond to porcelain by itself chemically.

b. Palladium (Pd): Palladium is also a noble metal which means that it resists corrosion and tarnish, and almost the opposite of gold. It is hard, very strong, and white and has a high melting point. It has a very high modulus of elasticity which means it is not very ductile.

Its native thermal expansion is very low and by itself cannot be used with porcelain because porcelain would “craze” (the opposite of shivering) and break off the substructure during cooling.

Even relatively small amounts of palladium will whiten gold dramatically. When added to a gold alloy, it will raise the melting range, raise the modulus of elasticity, and improve strength and hardness.

Small amounts of palladium dramatically improve the tarnish and corrosion resistance of gold-silver-copper crown and bridge alloys. It is an essential component for preventing tarnish and corrosion in Au-Ag-Cu alloys with gold content below 68% by weight.

Palladium and gold are completely soluble in one another, both as liquids in the molten state, and as solids in the finished alloy.

Palladium and gold are found together in so many dental alloys because they compliment each other. Unfortunately, the correct combination of gold and palladium sufficient to produce, say, the correct coefficient of expansion will not necessarily produce an alloy that meets the other necessary characteristics such as modulus, color or stiffness that a lab or manufacturer may need to produce a correct product. Hence, it is necessary to balance the formula with other metals as well.

c. Platinum (Pt): Platinum is used as an alternative to palladium in order to maintain a yellow color in the final alloy. It raises the melting range, increases the hardness, strength, and modulus, and lowers the thermal expansion of the alloy. It is less effective than palladium in producing these effects, but it is able to alter these characteristics with less impact on the golden color of the finished product.

d. Silver (Ag): In PFM alloys, silver is used principally to raise the thermal expansion of the alloy in order to balance the low thermal expansion of Palladium. Silver also lowers the melting range of both gold and palladium and adds fluidity to the melt improving its casting properties.

In gold-silver-copper alloys used for all-gold restorations, the silver compensates for the reddish color imparted by the copper. It also acts along with the copper to increase the strength and hardness of the alloy.

The major problem with silver in PFM formulations is that the silver can impart a greenish tint to the finished porcelain. This danger is offset by the very dramatic effect the silver can have on the modulus of expansion, and by the fact that modern porcelains are now formulated to resist this greening effect.

e. Copper (Cu): In crown and bridge alloys (all-gold), copper's major job is hardening and strengthening the alloy. It also imparts a reddish color, which may be an advantage, but can be offset by adding silver.

In PFM alloys, it is used mostly to increase the modulus of thermal expansion, and is responsible for the dark oxide layer characteristic of palladium-copper-gallium alloys.

Unfortunately, copper, like silver can cause discoloration of the overlying porcelain, however this effect is seldom seen when there is a very high percentage of palladium in the mix. Thus copper is seldom used in high noble PFM alloys (these alloys have lots of gold and little palladium.

f. Zinc (Zn): Zinc is used in crown and bridge alloys primarily as an oxygen scavenger. Zinc readily combines with oxygen that may have dissolved in alloy when it was in a molten state. This prevents the oxygen from forming gas porosity in the casting. In PFM formulations, zinc also lowers the melting range, increases strength and hardness, and raises the thermal expansion.

g. Indium (In): In crown and bridge alloys such as gold-silver-copper, indium is added to improve the fluidity of the melt thus improving castability. In PFM alloys, it strengthens and hardens both gold and palladium, and raises the thermal expansion of both. Indium also lowers the melting range of both gold and palladium. Indium contributes to the formation of the porcelain bonding oxide layer.

h. Tin (Sn): Tin is added to an alloy to increase the strength and hardness of both palladium and gold. It also lowers the melting range and raises the thermal expansion. Like indium, tin also contributes to the formation of the porcelain bonding oxide layer.

i. Gallium (Ga): Gallium is used almost exclusively in palladium based PFM alloys. Gallium can be a potent strengthener, and it lowers the melting range of palladium.

j. Iron (Fe): Iron is used almost exclusively in gold-platinum based PFM alloys. It is used as a strengthener. Iron also contributes to the formation of the porcelain bonding oxide layer.

k. Cobalt (Co): Copper is sometimes used as a substitute for copper in palladium based PFM alloys. Mostly, it is used along with nickel to formulate alloys for partial denture frameworks.

l. Ruthenium (Ru), Iridium (Ir) and Rhenium (Re): These three elements are used in very small concentrations as grain refiners. Alloys have better characteristics if the grain structures are small. The addition of small amounts of any of these three elements helps to produce small grain size when the alloy is cooling from the molten state.

The theory behind this is as follows: These elements are having a fairly high melting point and tend to be the first to form crystals in the molten matrix. Their low concentration allows their atoms to "atomize" and distribute themselves more or less evenly throughout the melt. As the grains of these elements form, they remain very small due to their low concentration throughout the solution. Since they crystallize first, these tiny grains form the nucleus around which the other elements begin to form their larger grains. The even distribution of grain formation throughout the solution limits the size of the larger grains as well.

2.3 Veneering Materials

It could be said that the ceramic material known as porcelain holds a special place in dentistry because, notwithstanding the many advances made in composites and glass-ionomers, it is still considered to produce aesthetically the most pleasing result. Its color, translucency and vitality cannot as yet be matched by any material except other ceramics.

Traditionally, its use is in the construction of artificial teeth for dentures, crowns and bridges. From the 1980s onwards the use of ceramics has been extended to include veneers, inlays/onlays, crowns and shortspan anterior bridges. The construction of such restorations is usually undertaken in dental laboratories by technicians skilled in the art of fusing ceramics.

As people retain their teeth for much longer than in the past, the need for aesthetically acceptable restorations is continuing to increase. This is reflected in the growing use by dentists of restorative procedures using ceramics. In below, it is given a series of differing ceramic structures available for dentistry (Ironside & Swain, 1998).

a. The Feldspathic Porcelains

Several studies 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. 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 now substantially glassy and Binns (1983) describes their classification as a porcelain as “somewhat of a misnomer”. An approximate location of dental porcelain in the classical tri-axial whiteware formulations is shown in Figure 2.1.

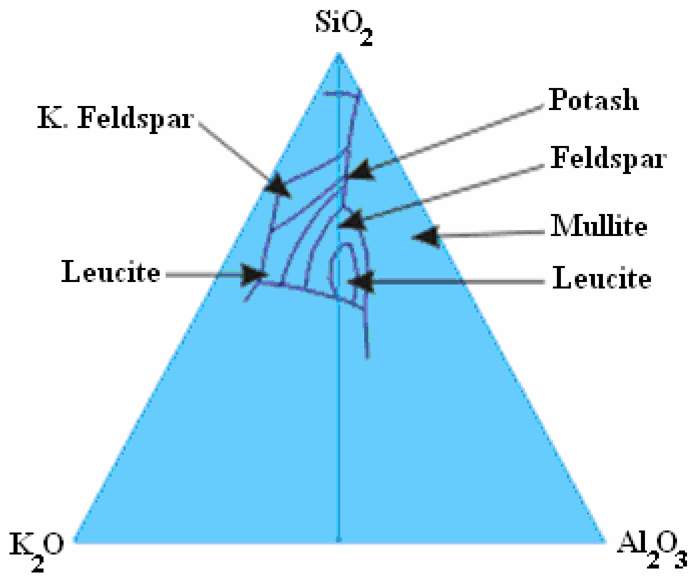


Figure 2.1 Location of dental porcelains in the classical triaxial whiteware formulations system

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 a greater proportion of leucite crystals (coefficient of thermal expansion $27x10^{-6}/°C$) which led to the overall coefficient of thermal expansion rising to something in the order of $13.5\text{-}15.5x10^{-6}/°C$.

The feldspathic porcelains used in all-ceramic systems have coefficients of thermal expansion ranging from $5.5\text{-}7.5x10^{-6}/°C$ when used over castable glass and alumina based core materials, to $16x10^{-6}/°C$ when used over the newer pressed leucite systems.

b. The Leucite Systems

Leucite has been widely used as a constituent of dental ceramics to modify the coefficient of thermal expansion. This is most important where the ceramic is to be fused or baked onto metal.

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.

Similar versions using finely dispersed leucite grains to increase toughness, strength and modify wear patterns and rates to make them similar to enamel wear rates are now available for metal-ceramic restorations.

c. 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 of glass with various forms of mica. The Dicor crown system 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 fluromica 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. In this latter technique an optical scan of a prepared tooth is loaded into a computer and a milling system is used to produce the restoration. The restoration is then “bonded” to the remaining tooth structure using a dental BisGMA based composite resin.

d. The Alumina Based Systems

- **The Aluminous Jacket Crown:** The modern Aluminous Jacket Crown, probably more commonly known as the Porcelain Jacket Crown (PJC) was popularised in the mid 1960’s by McLean. 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 aluminous porcelains reported by McLean are also very prone to strength degradation when they contain porosity.

- **Pure Alumina Core-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. Unlike the other dental ceramic materials, there is no glassy phase present between the particles. Feldspathic veneering porcelains such as Vitadur Alpha and Duceram are then fired onto this alumina core to provide the colour and form for the restoration.

- **The Glass-Infiltrated Alumina System for Cores:** During the 1980’s, Dr. Michael Sadoun and Vita Zahnfabrik, developed a slip casting system using fine grained alumina. The cast alumina was sintered and then infiltrated with a Lanthana based glass. This provided a glass infiltrated alumina core (In-Ceram) on which a felspathic ceramic could be baked to provide the functional form and aesthetic component of the restoration. In-Ceram has the highest flexural strength and fracture toughness of all the currently available dental ceramic systems available to most commercial dental laboratories. The system also has the greatest versatility for dental use of any metal free ceramic restorative.

The driving force for these developments 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. Typical mechanical properties of dental ceramics and tooth structures are listed in Table 2.2.

Table 2.2 Strength of tooth structures and dental ceramics

Material		**Flexural Strength (MPa)**	**Fracture Toughness (MPa/m^2)**
Porcelains	Feldspathic	60-110	1.1
	Leucite	120-180	1.2
Glass Ceramics	Lab cast/Cerammed	115-125	1.9
	Premade/HIP	140-220	2.0
Alumina	Alumina/Glass Infiltrated	400-600	3.8-5.0
	Spinel/Glass Infiltrated	325-410	2.4
Tooth Structures	Dentine	16-20	2.5
	Enamel	65-75	1.0

These are examples of the different directions that have been chosen to improve mechanical properties while maintaining aesthetic and economic considerations.

2.4 Opaque Porcelains

An opaque dental ceramic (porcelain) paste for applying to dental substructures, to be used in the preparation of ceramic dental appliances, such as crowns and bridges, and a method of using the same are provided.

When crowns, bridges and other metal dental substructures (copings) are to be veneered, the dental technician must first apply and fire a layer of opaque ceramic paste. This layer ensures a good bond to the veneering ceramic and, in addition, masks the unfavorable metal color of the substructure which is necessary to obtain a good aesthetic appearance.

Before the application of the opaque paste, an opaque slurry may optionally be fired on. For this purpose a thin suspension of ceramic opaque is mixed, applied and fired.

Conventional opaque ceramic paste consists of a pulverized opaque ceramic powder which is mixed with water, or a special modeling liquid, by the dental technician. The opaque powder itself consists of ground glass frits, which, owing to their chemical compositions can be melted at temperatures below 1000°C and opacifying agents. During firing, leucite is partially crystallized out, thus the thermal expansion of the glass is adapted to the thermal expansion of the metal. A high portion of opacifiers ensures a good masking of the dark metal. The opaque materials are sold in several shades since they form the color basis for the desired tooth shade.

In preparing the opaque ceramic paste, the dental technician must adjust the consistency of the paste to obtain a paste which is sufficiently viscous so as to adhere to the metal substructure without sagging, but not so viscous that it cannot be easily applied. This procedure involves much trial and error and is very time consuming.

Conventional opaque application requires much time and skill because a very even layer thickness has to be achieved. If the layer of ceramic opaque is too thin, the metal substructure shines through, when the opaque is too thick, there is not enough space left for the full application of the subsequent body and enamel layers, which results in a diminished aesthetic affect.

Accordingly, there is a need in the art for an opaque paste, and a method, which makes it possible to avoid the time consuming procedure of mixing the paste in the laboratory, and makes it possible to apply opaque ceramic paste more easily and more consistently, using fewer steps, to dental substructures and whereby an even layer thickness is achieved (Polz, July 11, 1989).

CHAPTER THREE

MECHANICAL DEFECTS ENCOUNTERED DURING PREPARING AND APPLYING OF DENTAL CROWNS

3.1 Introduction

Ceramic dental prostheses have become a popular restoration of choice for crown and bridge applications because of their superior aesthetics and biocompatibility. However, the strength of the ceramic remains a problem for a restoration’s longevity. To overcome this problem, most of these systems require the combination of two layers of material such as a metal alloy or a strong ceramic core and a weak veneering porcelain with better optical properties.

Compatibility of the ceramic system is dependent on the harmony of properties of both materials (substructure and veneering material). Acceptable restorations require chemical, thermal, mechanical and aesthetical compatibility between these components. Thermal and mechanical compatibility include a ceramic firing temperature that does not cause distortion of the substructure, in conjunction with the optimal combination of coefficient of thermal expansion. If the passive fit of a substructure changes after ceramic firing, the reason for this may be thermal incompatibility. This is a potential problem since the ceramic of tight-fitting substructure-ceramic restorations can subsequently fail. During the cooling cycles, the different thermal and mechanical behaviour of the materials used could produce transient and residual stresses in the restorations. If there is a significant difference in thermal expansion behaviour between the metal and veneer ceramic, residual stresses at non negligible levels remain on the crown. These stresses are the major factors contributing to the potential for catastrophic failure.

3.2 Effects of Cracks Occurred during Production of Dental Crowns

Because of thermal expansion coefficient mismatch, occurring the micro cracks on ceramic is unavoidable during the production phases (powder mixing and pressing, forming, drying, firing, last forming) of dental crowns. These micro cracks are exposed to cycling loads in the course of chewing. Initially, these micro cracks progress slowly, and then, fracture of ceramic happens with the effect of its stress intensity and moist ambient. The influential factors of the strength of dental ceramic are preparation shape, status of the supporting tooth tissue, shape and conformity of the restoration, thermal and mechanical properties of the materials, stress distribution on the teeth, material thickness and production technique.

During the preparing of the restoration, the veneering material (e.g. porcelain) are fired in high temperatures (950ºC) along with substructure material to bake entirely the porcelain plastered on the substructure material. Owing to temperature difference between the outer media and own, and different mechanical behaviors of two structures, thermal stresses take place markedly in the restoration removed from the own, after the firing process. In the some situations, these stresses cause the fractures even can be seen with eye in the outer surface of dental crowns. These fractures' happening is more probable in the interfaces that the stress intensities are higher strength in. Besides, the residual stresses at a certain level remain in the restoration which reaches in the room temperature. As a matter of fact, the existence of these residual stresses in the outer surfaces of dental crowns and bridges has been proved with the various experimental studies done. Also the cooling velocity of the restoration removed from the own is an important factor affects the stress intensities and fracture occurance alongside the different mechanical behaviors of the substructure and veneering materials. Due to both chewing forces and hot-cold foods which are eaten, dental crowns and bridges applied to patient's mouth subject to the great amount of mechanical and thermal fatigues. Hence, the micro cracks, occurred during preparing of the dental restoration, grow up with the effect of the fatigue loads and cause the fracture of the crown.

This problem met frequently at present affects the health of the patient directly and the necessity of the elimination of this problem or the decreases the lowest levels appears. Overcoming this problem is possible by taking the measures decreased the stresses and fractures, and increased the fatigue life.

3.3 Strength of Dental Restorations

3.3.1 All ceramic restorations

All-ceramic systems are attractive materials in restorative dentistry because they provide excellent biocompatibility and aesthetics. The improvement in aesthetics has been achieved by the application and bonding, through cycle firings, of layers of translucent veneering dentin and enamel porcelains onto a high strength ceramic core. Unfortunately, most all-ceramic systems seem unsatisfactory due to their high failure rate (Kelly et al., 1995; Scherrer et al., 2001). The primary reason might be the thermal incompatibility between ceramic core and veneering porcelains, which can introduce residual thermal stresses resulting in fracture or cracking of the restoration. In the event of moderate residual thermal stresses, a ceramic structure can be permanently distorted. Most ceramics are brittle materials and approximately 0.1% deformation may result in fracture (Jones et al., 1972) due to the propagation of cracks, usually present on the surface, through the bulk of the material. This deformation limit has to be taken into consideration when a ceramic core and veneering porcelain materials are designed to be bonded together. Of course, all-ceramic systems are developed by their manufacturers to be compatible in order to prevent thermal residual stresses. However, incompatibility in commercially available systems sometimes arises. In addition to that, dental technicians tend, for economic reasons, to veneer a ceramic core with porcelains leftover from other all-ceramic systems. From the literature review it is known that some factors such as the magnitude of temperature change and differences in material properties, including the coefficient of thermal expansion (α), the glass transition temperature (T_g), and the viscosity (η), might affect thermal stress in layered ceramic structures (Kingery et al., 1976; Whitlock et al., 1980).

The most important disadvantages of the all-ceramic restorations in comparison with porcelain-fused-to-metal (PFM) and all metal restorations are having lower fracture strength. Although the compressive strengths of the ceramics are high, its tensile strengths are low. In practice, besides shearing forces, hitting forces occurs during chewing. The stresses constituted by these forces cause the fractures in the restoration. Full ceramic restorations have only 1% elastic deformation. Owing to the fact that metals have higher modulus of elasticity, PFM and all metal restorations exhibit higher elastic deformation. Consequently, full ceramic restorations are more brittle.

3.3.2 Porcelain Fused to Metal (PFM) restorations

Porcelain-fused-to-metal crowns look like natural teeth and are stronger than ceramic crowns. PFM crowns can be matched to your natural teeth so they provide an attractive appearance. However, the porcelain portion can be chipped off and the underlying metal can peer through as a dark line. Stronger than all-porcelain crowns, PFM crowns also wear down, and can show more of the dark line as the gum recedes.

While PFM crowns are definitely more beautiful than metal crowns, rarely do they match up to all-porcelain crowns in beauty. And while PFM crowns are more fracture-resistant than all-porcelain crowns, they can still fracture unlike all-metal crowns. Thus, while PFM crowns do incorporate good aspects from both metal and porcelain, they are not the perfect solution in all cases. Also, for individuals who have metal allergy, PFM crowns may cause a reaction.

PFM crowns are most commonly used on the back chewing teeth. Here their strength is an advantage, yet the less natural appearance is not as noticeable because they are not as visible when one smiles. PFM crowns can and are used on front teeth in cases where strength is required, and if created properly, they can be very beautiful.

3.3.3 Full metal restorations

Advantages of metal crowns, such as full gold crowns, include their strength and durability, as well as their excellent fit. Because of its physical properties, dentists find gold to be a very workable metal. This characteristic helps a dentist to be able to achieve a very precise fit with the crown.

Since they are metal through and through, gold crowns withstand biting and chewing forces well. They will not chip. It would be uncharacteristic for a gold crown to break. Of all of the types of dental crowns, gold crowns probably have the greatest potential for lasting the longest.

Although they are very strong, the wear rate of a gold crown is about the same as tooth enamel. This means that a gold dental crown won't create excessive wear on the teeth it opposes (the teeth it bites against).

Unfortunately, they are not very natural or beautiful, and some alloys contain metals, such as nickel or zinc, that have a relatively high incidence of allergy and can therefore cause irritation to oral tissues.

While all-metal crowns are still used in dentistry, they are much less common now than in the past. They are used primarily in situations where high chewing stresses are expected and may cause premature failure of crowns made with porcelain.

3.4 Objective of This Dissertation

PFM crowns are fabricated by fusing porcelain at high temperatures (~1000ºC) to an alloy frame. As soon as the ceramic's temperature falls below its glass transition temperature (T_g) during the subsequent cooling phase, it changes from a sluggish, plastic into a brittle, elastic state. Thereafter, thermal stresses build up in the crown during further cooling.

The transient thermal stresses (stresses occurring during the cooling phase between glass and ambient temperature) depend on a large number of parameters: mode of cooling, glass transition temperature of the porcelain, temperature dependence of the mechanical and thermal coefficients of alloy and ceramic (Modulus of elasticity, Poisson's ratios, thermal expansion coefficients, thermal conductivities, heat capacities).

The residual thermal stresses (stresses in the final configuration at ambient temperature), however, are only a function of Modulus of elasticity and Poisson's ratios at room temperature, the difference of the thermal expansion coefficients of the two materials between the glass temperature of porcelain and ambient temperature, the glass temperature itself.

The aim of this dissertation is to reveal that it is possible to determine transient and residual stress distribution occurring in crown and bridge restorations with adequate sensitivity by using elastic or elastic-plastic solution methods. To this end, firstly, the mechanical and thermal properties at various temperatures of the materials used in a porcelain fused metal (PFM) system were obtained by experimental and literature studies. The values of modulus of elasticity of the dental porcelain on viscoelastic region were calculated by using its viscoelastic properties. The influences of the viscoelastic and viscoplastic behaviors were reflected on the elastic properties of the porcelain. The cooling curves were measured experimentally after the standard firing process of a central incisor crown restoration. The equivalent convection coefficients, calculated by the help of the cooling curves, were applied in the 3D finite elements model. Numerical and experimental cooling curves were made to be congruent with each other. It has been observed that the maximum stresses occur within a short time period after cooling begins and that stresses decrease during the cooling process and remain at a constant value at the end of process; these are the thermal residual stresses.

3.5 Literature Review

The most common used method to predict residual thermal stresses in a system is to determine the relation of temperature with the expansion of each material separately and compare them. According to ISO 9693 for metal–ceramic dental restorative systems, the thermal expansion coefficient which is determined by heating the specimens at a low rate (+5°C/min) from room temperature up to 500°C or up to the T_g (glass transition temperature) when the T_g is lower than 500°C. This standard defines the coefficient of thermal expansion (CTE) as the percentage increase in length per degree centigrade over the latter temperature range, thereby assuming a linear relationship between temperature and dimensional change. Although the calculated CTE can be used by the manufacturer for batch control, it is questionable whether it is an appropriate tool for the prediction of thermal compatibility. Moreover, during the routine firing process in dental laboratories, cooling rate is in general higher than standard dilatometric measurements. Therefore, during cooling, the values of T_g and T_s (softening temperature) increase a higher level.

Fairhurst et al. (1981) determined the values of T_g for dental porcelains as a function of heating or cooling rates. The method described in their study provided a means of determining the glass transition temperature of porcelains during cooling or heating.

Twiggs et al. (1985) investigated T_g values at high heating rates. T_g values are reported for a number of commercial dental porcelains and other materials. These data were obtained by the bending beam technique, employing a special low thermal-mass furnace to permit rapid heating and cooling rates. Measurements were made at rates as high as 600°C/min. This relation is consistent with previously reported low rate T_g data obtained by a different technique.

An analytical model that considered the effects of the applied load and heating rate on T_g for dental porcelain was developed by DeHoff & Anusavice (1986). In their study, an equation similar to the Moynihan equation relating heating rate to T_g was found for midspan load. The validity of the analytical model was demonstrated by using viscosity data from one published study to predict T_g behavior in an independent study. Values of T_g calculated with the analytical model for a dilatometer specimen were found to be higher than T_g values calculated for a bending beam viscometer.

Asaoka & Tesk (1990) calculated T_g according to the cooling rate in their computer simulation in order to obtain the residual stresses in porcelain-fused-to-metal (PFM) restorations. They conducted a computer simulation of stress developed in a PFM beam.

Lenz et al. (1998) heated rotationally symmetric, premolar porcelain-fused-to-metal (PFM) crowns in a porcelain furnace to barely above the ceramic’s glass transition temperature (~600ºC). After opening the furnace, they simultaneously measured local temperatures at the crown’s surface with thermocouples and recorded during the cooling phase down to ambient temperature. They smoothened and used these data as input to a finite element model to calculate the internal temperature and stress distribution as a function of time. Finally, they found maximum residual stresses at the bond interface in the transition region from the occlusal to the wall section.

Some factors in firing process such as high firing temperatures and irregular geometrical shapes of the restorations make it difficult to determine the distribution and amount of the residual stresses in experimental methods. In addition, the porcelain exhibits linear elastic behaviour between room temperature and T_g, viscoelastic behaviour between T_g and T_s, and viscoplastic behaviour between the

T_s and T_m (melting temperature). Therefore, it is necessary to consider the effects of all the behaviors and utilize computer aided solutions for these types of problem.

A structural model which combined the viscoelastic theory with the structural relaxation theory was developed by Narayanaswamy (1978). Predictions of this structural model are compared with experimental data on tempering and contrasted with predictions of the viscoelastic model. This study revealed that structural relaxation accounts for approximately 24% of the total residual temper stresses.

Anusavice & Dehoff (1989a) calculated the viscosities of body and opaque porcelains as a function of temperature from creep rates measured for temperatures ranging from 550ºC to 625ºC. They have developed analytical models, which incorporate linear viscoelaticity and structural relaxation effects, to calculate transient and residual stresses in dental porcelain plates, semicircular arch with gapped diametric segment (Anusavice & Dehoff, 1989b) and bilayered porcelain discs (Anusavice & Dehoff, 1992).

The correlations between transient and residual stresses, calculated by viscoelastic analyses for two tempering methods and the resistance to the failures of porcelain disks during the tempering treatment were examined by Asaoka et al. (1992). They calculated the residual stresses, as affected by the tempering method and thickness of a porcelain disk, by computer simulation for regions away from the edges. In their study, simulated residual stresses were also in good agreement with those measured by the indentation fracture method.

Dehoff et al. (2006) calculated residual stresses in an all-ceramic, three-unit posterior fixed partial denture for four ceramic–ceramic combinations by using three dimensional model and viscoelastic option of a finite element software program. They constructed a three-dimensional finite element model of the FPD from digitized scanning data and performed calculations for four systems. They concluded that the high residual tensile stresses in all-ceramic fixed partial dentures (FPDs) with a layering ceramic may place these systems in jeopardy of failure under occlusal loading in the oral cavity.

Residual thermal stresses could be also calculated by the fracture mechanics method. Coffey et al. (1988) aimed to determine the influence of cooling rate and contraction mismatch on the flexural failure resistance of metal opaque-porcelain strips. They subjected to specimens to four-point loading until crack initiation occurred at the metal-ceramic interface and determined the total stress induced from the residual stress and the applied flexural load for these specimens.

Anusavice & Hojjatie (1991) tested the hypothesis that tempering stress can retard the growth of surface cracks in layered porcelain discs with variable levels of contraction mismatch. In addition, they performed biaxial flexure tests to determine the influence of mismatch and tempering on flexure strength.

It was hypothesized that the global residual stresses within the surface of those veneered FPDs may be responsible for partial fragmentation of the veneering ceramic by Taskonak et al. (2005). They determined global residual stresses in the veneer layer, calculated using a fracture mechanics equation, to be responsible for the increased strength and observed chipping, i.e., spallation in bilayer ceramic composites.

From the other point of view, some authors assumed that the dental porcelain is a linear elastic material at all firing temperatures. However, in this type of the assumption, the elastic properties should include the effects of the different behaviors at elevated temperatures (Dai & Shaw, 2004; Jager et al., 2006; Lenz et al., 2002).

CHAPTER FOUR

DETERMINATION OF PHYSICAL, THERMAL AND MECHANICAL PROPERTIES OF THE MATERIALS USED IN THIS STUDY

4.1 Materials Used in This Study

4.1.1 Substructure materials

In this study, the substructure materials which are the most preferred in the clinical studies at present have been used. Specially, it has been working on non-precious, semi-precious and precious alloys among these substructure materials. For this reason, Ni-Cr and Co-Cr as non-precious alloys, Pd-Ag-Au as semi-precious alloy and Au-Pt as precious alloy have been chosen in the light of the clinical studies. The chemical compositions of the chosen substructure materials are given in Table 4.1 (Wirobond C, nd; Wirocer Plus, nd; Capricorn 15, nd; Brite Gold, nd).

Table 4.1 Chemical compositions of the substructure materials used.

Type of Alloy	**Commercial Name (Manufacturer)**	**Chemical Composition**
Ni-Cr	Bellabond Plus (Bego/Germany)	65.2% Ni, 22.5% Cr, 9.5% Mo, Fe-Si-Mn-Nb<2%
Co-Cr	Wirobond C (Bego/Germany)	61% Co, 26% Cr, 6% Mo, 5% W, 1% Si, Fe-Ce-C<1%
Pd-Ag-Au	Capricorn 15 (Ivoclar/Liechtenstein)	51.9% Pd, 23% Ag, 15% Au, 8% In, 2% Ga, Re-Ru<1%
Au-Pt	Brite Gold (Ivoclar/Liechtenstein)	96.3% Au, 2.6% Pt, Cu-In-Ir-Sn-Fe-Mn-B<1%

4.1.2 Veneering material

Commercially, lots of various porcelains are produced by a good number of manufacturers. But the differences between the substructure and veneering materials' thermal expansion coefficients are caused to occur the thermal stresses in the crown and consequently, the crown fractures. Hence, it must be paid attention to choose more compatible material pairs in terms of the thermal expansion coefficients, with regard to choose the material.

In the present study, considering this point, it has been worked on the porcelain named VMK 95. The chemical composition of the porcelain under consideration is given in Table 4.2.

Table 4.2 Chemical composition of porcelain used (Vita Metal Ceramics, VMK 95, nd).

Type of Porcelain	**Commercial Name (Manufacturer)**	**Chemical Composition**
Feldspathic	VMK 95 (Vita/Germany)	52.4% SiO_2, 15.15% Al_2O_3, 9.9% K_2O, 6.58% Na_2O, 2.59% TiO_2, 5.16% ZrO_2, 4.9% SnO_2, 0.08% Rb_2O, 3.24% B_2O_3, CO_2, H_2O

4.2 Physical, Thermal and Mechanical Properties of the Materials

To determine the material properties depending on the temperature, it has been profited from the experimental and theoretical methods and literature investigations within this dissertation. In below, it has been given the methods of determining the physical, thermal and mechanical properties depending on the temperature of all of the materials.

4.2.1 Density (d)

For all of the materials, it has been taken the values given by the manufacturers at the room temperature for density. For the high temperature values, it has been assumed that density can not be change. These values are 8.2 gr/cm^3 (Wirocer Plus, nd), 8.5 gr/cm^3 (Wirobond C, nd), 11.5 gr/cm^3 (Capricorn 15, nd), 19.0 gr/cm^3 (Brite Gold, nd) and 2.4 gr/cm^3 (Vita Metal Ceramics, VMK 95, nd) for Ni-Cr, Co-Cr, Pd-

Ag-Au, Au-Pt alloys and porcelain, respectively. Figure 4.1 depicts that changing of density of all of the materials depending on the temperature.

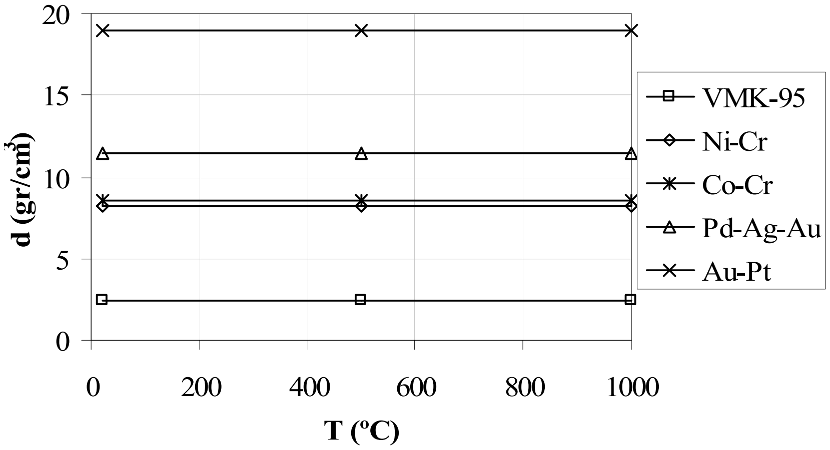


Figure 4.1 Changing of the density of the materials depending on the temperature.

4.2.2 Thermal conductivity (k)

Obtaining the values of thermal conductivity is quite difficult for alloys. Even additions at very little ratios into the alloy may change the thermal conductivity considerably. On the other hand, the more the element diversity increases the less the tendency of variation in the thermal conductivity occurs. Therefore, it is possible to say that thermal conductivity are also at close values in alloys, which involve too many elements and the compositions of which are close to each other.

For this reason, it has been utilized from a material website for finding the values of thermal conductivity of the alloys used in this study. The alloys having the composition which is very close to the composition of the alloys used have been chosen. Also, the values of thermal conductivity at room and high temperatures for the porcelain have been taken as the same with the values in a study given in literature.

The values of the thermal conductivity coefficients depending on the temperature have been given in Figure 4.2 for the alloys and porcelain used in this study.

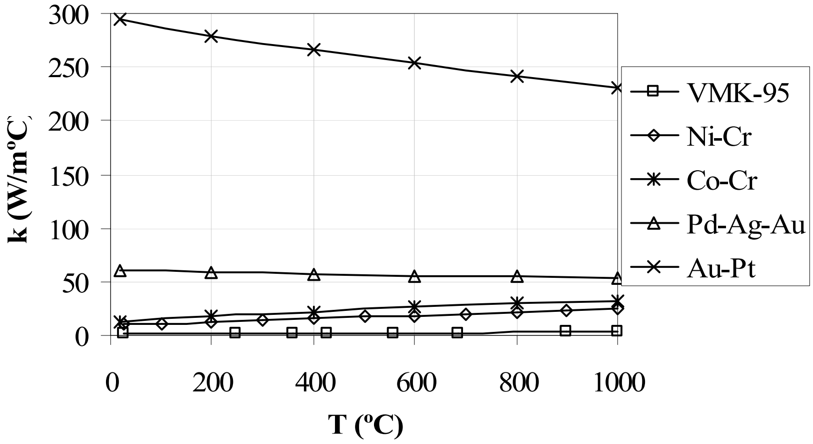


Figure 4.2 Changing of the thermal conductivity of the materials depending on the temperature.

*Ni-Cr (Matweb, Material property data, nd)

*Co-Cr (Matweb, Material property data, nd)

*Pd-Ag-Au (Platinum metals review, Palladium-30.00% Silver, nd)

*Au-Pt (Lenz et al., 2002)

*VMK-95 (Dai & Shaw, 2004)

4.2.3 Specific heat (C_p)

Similar to the calculation of the value of specific heat (C_p) according to the mass mixing ratios in gases, it can be calculated in the same way in the alloys that do not join by a chemical reaction (Çengel & Bones, 2006).

The mass ratio (y_i) of an element in an alloy can determined by taking the volume ratio as v_i, density (d_i), and mass in the unit volume (m_i) of an element in the alloy, its mass ratio (y_i) can be found by;

$$y_i = \frac{m_i}{\sum_{i=1}^{n} m_i} = \frac{v_i d_i}{\sum_{i=1}^{n} v_i d_i} \qquad [4.1]$$

and the specific heat $(C_p)_T$ of the alloy at temperature T can be determined as;

$$\left(C_p\right)_T = \sum_{i=1}^{n}\left[y_i\left(C_{p_i}\right)_T\right] \qquad [4.2]$$

For instance, the C_p values and densities of Nickel, Chrome and Molybdenum which are the fundamental elements constituting the Nickel-Chrome alloy have been found from the literature for certain temperatures (McBride et al., 1993) and given in Table 4.3 and 4.4.

Table 4.3 Specific heats of elements for Ni-Cr alloy (McBride et al., 1993).

Temperature (ºC)	**Specific Heat of Elements (J/kgK)**		
	Ni	**Cr**	**Mo**
27	444,39	451,49	248,09
127	482,38	485,08	260,16
227	530,03	512,23	268,10
327	593,76	531,58	274,45
427	524,60	548,99	279,80
527	529,75	567,35	284,62
627	538,06	588,52	289,34
727	548,37	613,53	294,28
827	560,37	642,73	299,71
1000	573,74	675,92	305,79

Table 4.4 Densities of elements for Ni-Cr alloy (McBride et al., 1993).

Elements	**Density (gr/cm^3)**
Ni	8,9
Cr	7,15
Mo	10,2

From Equation [4.1], the mass ratios of the elements can be written;

$$y_{Ni} = \frac{m_{Ni}}{m} = \frac{v_{Ni} d_{Ni}}{m} = \frac{0.652 v d_{Ni}}{\left(0.652 d_{Ni} + 0.225 d_{Cr} + 0.095 d_{Mo}\right) v}$$

$$y_{Ni} = \frac{0{,}652.8{,}9}{0{,}652.8{,}9 + 0{,}225.7{,}15 + 0{,}095.10{,}2}$$

$y_{Ni} \cong 0{,}692$

Similarly, the mass ratios of Cr and Mo elements has been found as $y_{Cr} \cong 0{,}192$ and $y_{Mo} \cong 0{,}116$.

Then, with the help of Equation [4.2], the specific heat values of the Ni-Cr alloy has been determined for each temperature values and given in Table 4.5.

Table 4.5 Specific heats of Ni-Cr alloy at various temperatures.

Temperature (ºC)	**Specific Heat of Ni-Cr Alloy (J/kgK)**
27	423
127	457
227	496
327	545
427	501
527	509
627	519
727	531
827	546
1000	563

The specific heat values of the other substructure materials (Cr-Co, Pd-Ag-Au and Au-Pt) also have been found by the same method. Again, the C_p values and densities of the fundamental elements constituting the alloys have been obtained from the literature for certain temperatures (McBride et al., 1993).

For the porcelain, the specific heat values at various temperatures have been taken as the same with the values in a study given in literature (Dai & Shaw, 2004).

In Figure 4.3, the changing of the specific heat of the substructure materials and porcelain depending on the temperature has been given.

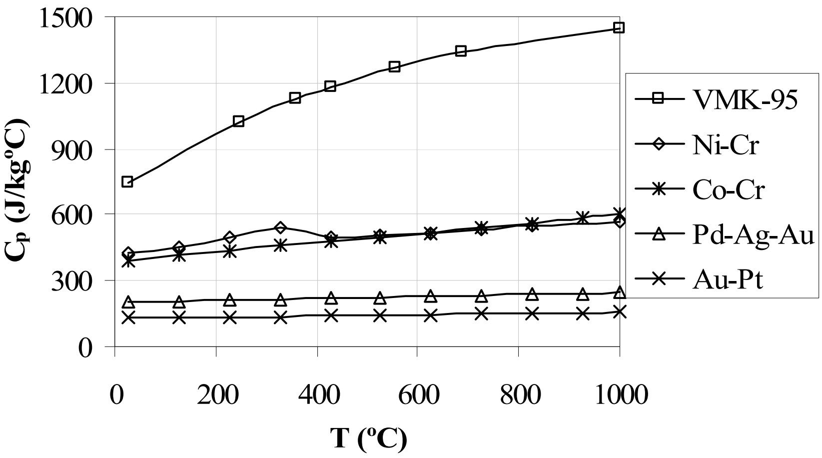


Figure 4.3 Changing of the specific heat of the materials depending on the temperature.

4.2.4 Thermal expansion coefficient (α)

The values of the thermal expansion coefficient between the room temperature and the glass transition temperature (T_g) are given by the manufacturer for the substructure materials. However, due to the fact that the thermal expansion coefficient values for high temperatures are not known, these values have been determined from the thermal behaviours of similar materials.

For example, the thermal expansion coefficient value between the room temperature and T_g of Ni-Cr alloy is given as $13.8\text{x}10^{-6}$-$14.0\ \text{x}10^{-6}$ 1/°C by the manufacturer (Wirocer Plus, nd). These values are not given by the manufacturer for high temperatures. Hence, the variations of the thermal expansion coefficient at high temperatures have been obtained from the thermal behaviour of similar material (Special metals, Inconel alloy 625, nd) at high temperatures.

The variations of the other substructure materials' thermal expansion coefficients also have been determined with the same methods for the room and high temperatures. The references used for these alloys have been given in Figure 4.4.

The thermal expansion coefficient of the porcelain is given the manufacturer from the room temperature to T_g (Vita Metal Ceramics, VMK 95, nd). In the viscoelastik region between T_g and T_s, the thermal expansion coefficient values has been suddenly increased and then decreased to a value above T_s similar the average curve obtained from dilatometric measurements for a feldspathic porcelain by Asaoka and Tesk (1990). After from this region, the variation of the thermal expansion coefficient of the porcelain in viscoplastik region has not been determined exactly. The result of the analyses, done to control the effect of the thermal expansion coefficient, it has been determined that the variation of the thermal expansion coefficient at high temperatures above T_s markedly affects the stresses occurred on the crown. Consequently, for higher temperatures above T_s, investigation of two different acceptances has been approved;

First acceptance: Thermal expansion coefficient (α) value has been decreased a little after from T_s, and then taken a constant value until the maximum temperature of the furnace. For this acceptance, α values of the porcelain have been called as α_1 as seen in Figure 4.4.

Second acceptance: α has been increased to assume that α of the porcelain reaches α value ($35 \mathrm{x} 10^{-6}$ 1/°C) of the liquid phase given in the studies of Asaoka and Tesk, 1990. For this acceptance, α values of the porcelain have been called as α_2 as seen in Figure 4.4.

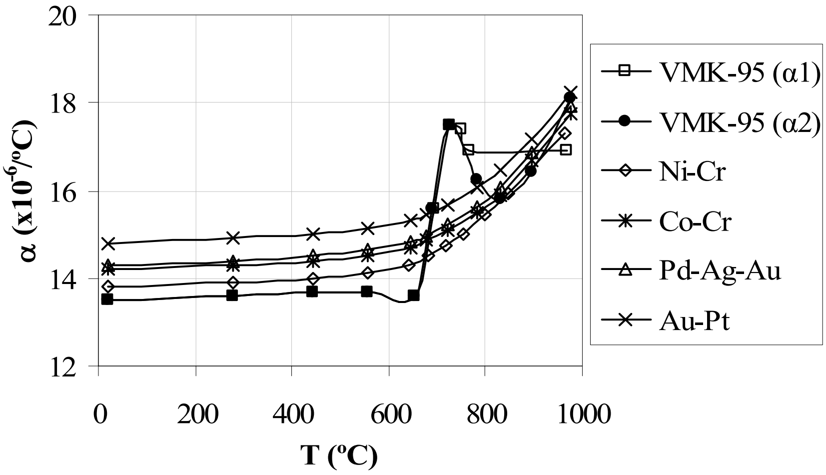


Figure 4.4 Changing of the thermal expansion coefficients depending on the temperature.

*Co-Cr (For room temperature, Wirobond C, nd)
(For high temperatures, Matweb, Material property data. nd)
*Pd-Ag-Au (For room temperature, Capricorn 15, nd)
(For high temperatures, Matweb, Material property data. nd)
*Au-Pt (For room temperature, Brite Gold, nd)
(For high temperatures, McBride et al., 1993)

4.2.5 Poisson's ratio (ν)

In this study, the variations of the poisson's ratio of the substructure materials and the porcelain used have been accepted to be the same with the variations used for the same type of alloy and porcelain in a study given in the literature (Lenz et al., 2002). These variations have been depicted in Figure 4.5.

4.2.6 Modulus of elasticity (E)

4.2.6.1 Modulus of elasticity of the substructure materials

To determination of the modulus of elasticity of the substructure materials depending on the temperature, hot compression tests have been done in GKSS Research Center in Germany, as seen in Figure 4.6. Ni-Cr, Co-Cr and Pd-Ag-Au alloys have been used for the tests. Au-Pt alloy has not been used in the tests due to its cost.

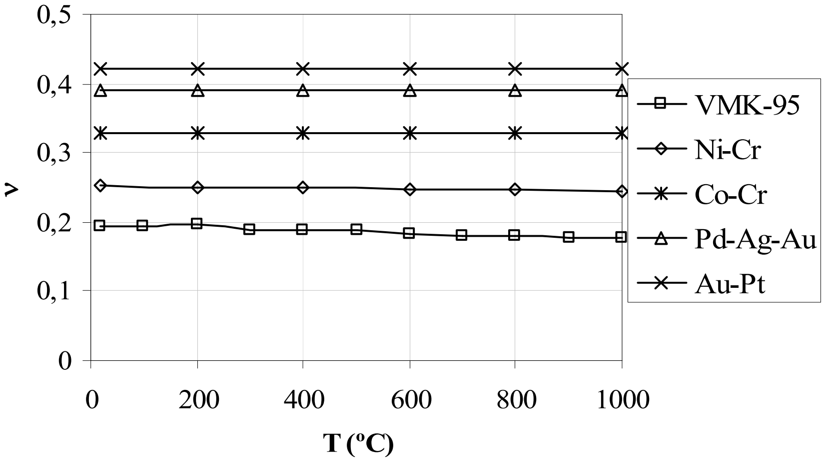


Figure 4.5 Changing of the poisson’s ratio depending on the temperature.

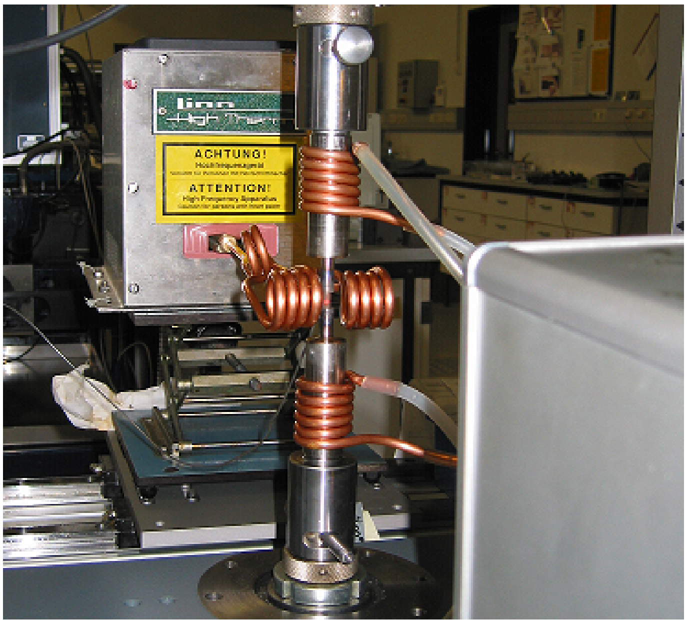


Figure 4.6 A view from the hot compression test mechanism in GKSS.

For the compression tests, firstly, the test samples were prepared from the alloys used. These alloys are produced by the manufacturer in case of ingots. These alloy ingots are obtained in cylindrical shape by the extrusion method. The substructure metals are used in dental laboratories by the casting of these ingots in routine applications. When the internal structures of these ingots are examined, it is encountered very little porosities. Therefore, it was decided to use these ingots, as seen in Figure 4.7, directly for preparation of the specimens used in compression tests.

Figure 4.7 Cylindrical ingots used to prepare the test samples.

The cylindrical samples in diameter of 4 mm and length of 8 mm were prepared from these ingots by turning. The residual stresses that were likely to occur during the turning were eliminated by stress relief process. These samples were subjected to compression tests at the temperature values of 20ºC, 200ºC, 400ºC and 600ºC. Experiments were carried out in a universal compression testing machine and using a laser extensometer. By the help of these tests stress-strain (σ-ε) diagrams of each substructure materials were drawn for each temperature as seen in Figure 4.8. But, in Figure 4.8, σ-ε diagrams at 800ºC and 1000ºC for the specimens used in the tests were drawn with the aid of the mechanical behaviours of alloys that have similar composition, at high temperatures (Special metals, nd). Similarly, σ-ε diagrams of Au-Pt alloy for each temperature were also obtained with the same method above.

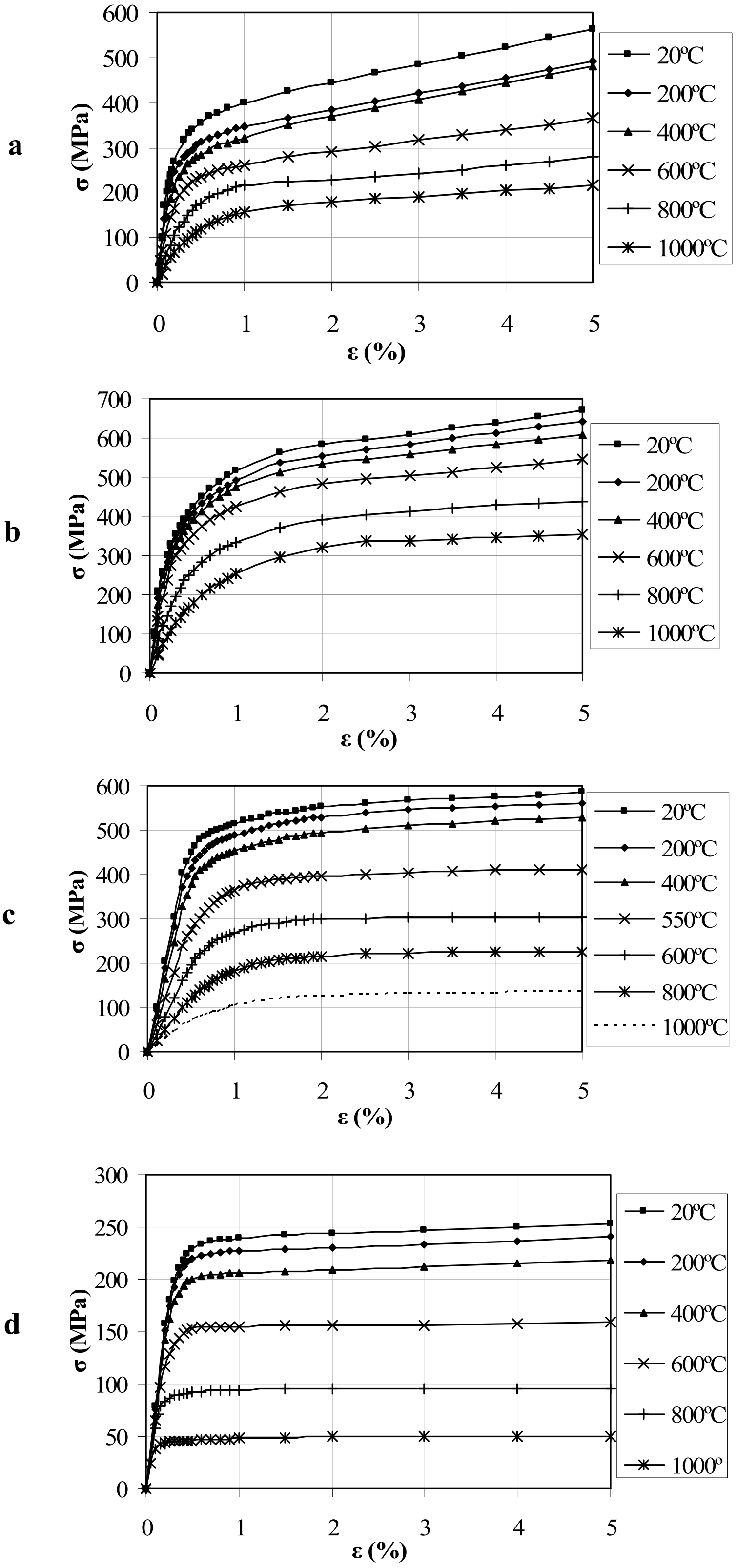


Figure 4.8 Stress-strain (σ-ε) curves of the substructure materials; **a)** Ni-Cr, **b)** Co-Cr, **c)** Pd-Ag-Au and **d)** Au-Pt.

The values of modulus of elasticity of the substructure alloys have been calculated from these σ-ε diagrams for each temperature. In the examination of σ-ε curves at 20ºC, it has been revealed that the values of the modulus of elasticity and the proof stresses are very close to the values given by the manufacturers at room temperature. The variations of the modulus of elasticity depending on the temperature for each substructure materials have been depicted in Figure 4.9.

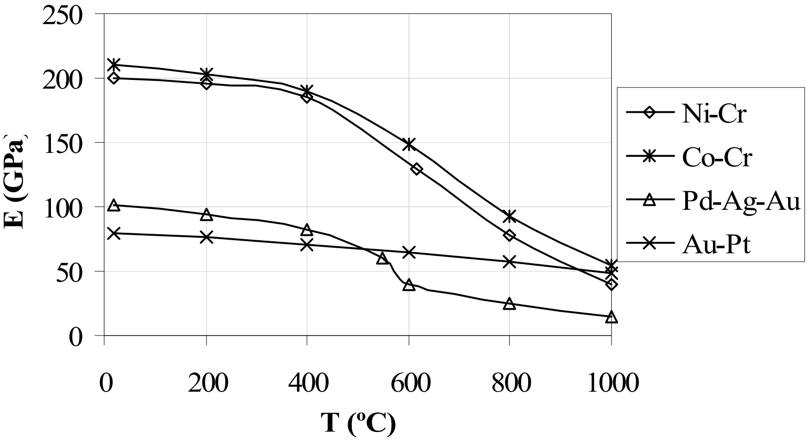


Figure 4.9 Changing of the modulus of elasticity depending on the temperature for substructure materials.

4.2.6.2 *Modulus of elasticity of the porcelain*

The modulus of elasticity (E) of the porcelain between room temperature and glass transition temperature (T_g) has been reported by the manufacturer firm (Vita Metal Ceramics, VMK 95, nd).

The dental porcelains show the viscoalestic behaviour form T_g to the softening temperature (T_s), and the viscoplastic behaviour from T_s to melting temperature (T_m). Hence, the modulus of elasticity values in the viscoelastic and viscoplastic regions have been obtained from the theoretical calculations taking care of the viscoalestic and viscoplastic behaviors. These calculations have been explained in detail in part 4.3.

The variation of the modulus of elasticity (E) of the porcelain depending on the temperature has been depicted in Figure 4.10.

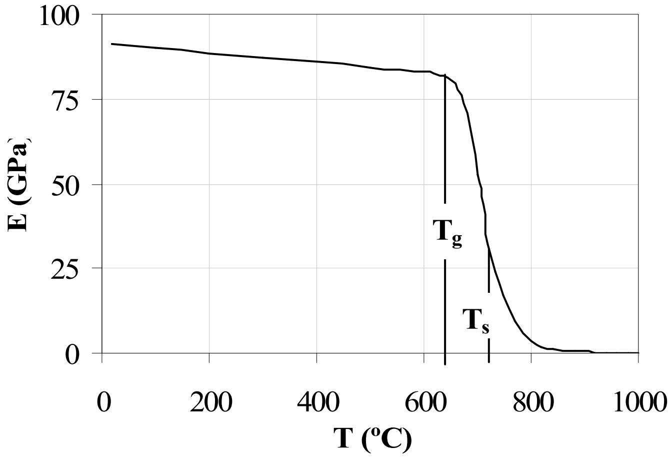


Figure 4.10 Changing of the modulus of elasticity depending on the temperature for porcelain.

4.3 Calculation of Modulus of Elasticity of Porcelain in Viscoelastic and Viscoplastic Regions

Viscoelastic materials show creep behaviour under constant load, at a temperature T_i and time Δt_i as seen in Figure 4.11a, whereas stress relaxation occurs under constant length (Figure 4.11b).

In this study, instead of E_{o_i} that gives ε_{0_i}, E_i that gives ε_i has been calculated (Figure 4.12) and used in the thermal stress analyses. By this way, the effects of viscoelastic behaviour of materials has been included in the elastic behaviour of the porcelain and taken into account in the results. From Figure 4.11a;

$$\varepsilon_i = \varepsilon_{0_i} + \varepsilon_{cr_i} \quad \Rightarrow \quad \frac{\sigma_{0_i}}{E_i} = \frac{\sigma_{0_i}}{E_{0_i}} + \frac{\sigma_{0_i}}{E_{cr_i}} \qquad [4.3]$$

Where; ε_i, ε_{cr_i} and E_{cr_i} are called as total strain, creep strain and creep modulus, respectively.

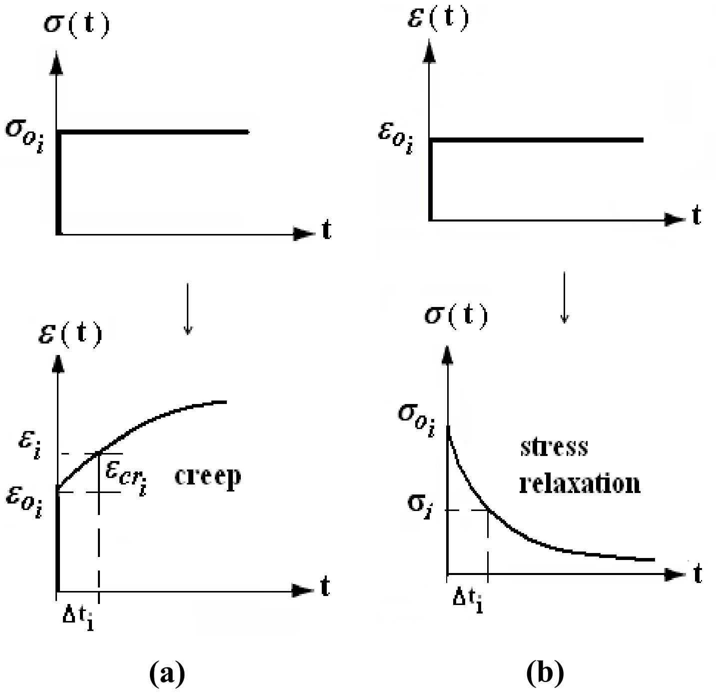


Figure 4.11 Typical mechanical behaviour of viscoelastic materials; **(a)** Creep, **(b)** Stress relaxation.

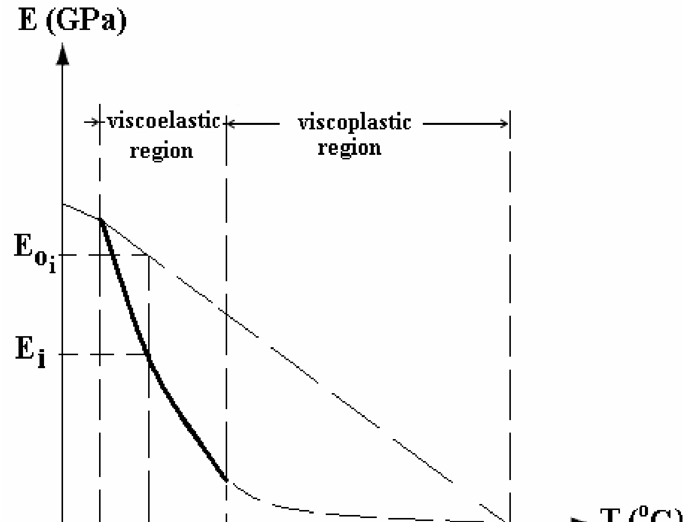


Figure 4.12 The variation of modulus of elasticity with temperature in the viscoelastic and viscoplastic regions.

$$\frac{1}{E_i} = \frac{1}{E_{0_i}} + \frac{1}{E_{cr_i}} \quad [4.4]$$

$$\frac{\varepsilon_{cr_i}}{\varepsilon_{0_i}} = \frac{\sigma_{0_i}/E_{cr_i}}{\sigma_{0_i}/E_{0_i}} = \frac{E_{0_i}}{E_{cr_i}} \Rightarrow \frac{\varepsilon_{cr_i}}{\varepsilon_{0_i}} = \frac{E_{0_i}}{E_{cr_i}} \Rightarrow \frac{1}{E_{cr_i}} = \frac{\varepsilon_{cr_i}/\varepsilon_{0_i}}{E_{0_i}} \quad [4.5]$$

$$\frac{1}{E_i} = \frac{1}{E_{0_i}} + \frac{\varepsilon_{cr_i}/\varepsilon_{0_i}}{E_{0_i}} \Rightarrow E_i = \frac{E_{0_i}}{1 + \frac{\varepsilon_{cr_i}}{\varepsilon_{0_i}}} \quad [4.6]$$

To calculate E_i from Equation [4.6], the creep ratio ($\frac{\varepsilon_{cr_i}}{\varepsilon_{0_i}}$) given in Figure 4.13 is required. For Kelvin-Voigt creep model, creep ratio curve could be defined as an Avrami type equation (Asaoka & Tesk, 1990).

$$\frac{\varepsilon_{cr_i}}{\varepsilon_{0_i}} = 1 - e^{-\frac{\Delta t_i E_{0_i}}{\eta_i}} \quad [4.7]$$

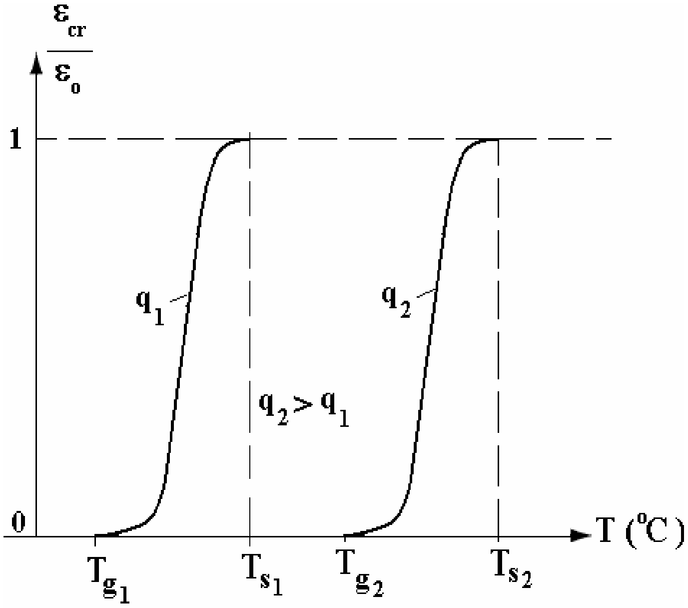


Figure 4.13 The variation of creep ratios with temperature in the viscoelastic region.

In Equation [4.7], η_i is called as the viscosity of the material at temperature T_i. The limit values of the curve of Avrami type equation change between 0 and 1 (Figure 4.11). By the knowing the values of T_g and T_s, maximum and minimum limits become apparent and the curve could be drawn more accurately. In this way, alteration in modulus of elasticity at various temperatures could be calculated by Equation [4.7] with the help of this curve.

Creep relaxation curves of PFM (porcelain-fused-to-metal) established numerically for different cooling rates by using Kelvin-Voigt model (Asaoka&Tesk, 1990). Δt_i was calculated according to the cooling rate. When the cooling rate was increased, curves were formed nearly parallel to each and reaching other to the high temperature interval. For this reason, firstly it is necessary to calculate T_g and T_s according to cooling rate (q).

T_g and T_s values of the veneering porcelain used have been reported as 583°C and 656°C,respectively by the producer firm. These T_g and T_s values are calculated by dilatometric analyses according to ISO 9693 metal-ceramic standards which is determined by heating the specimens at a low cooling rate (+5°C/min). If the cooling rate was 5°C/min in this study, it could be concluded that T_g and T_s will be the same with T_g and T_s values given by the manufacturer. However the measured cooling rate at viscoelastic region is about 700°C/min. In this situation T_g and T_s values approach to high values at dental laboratory conditions.

Fairhurst et al. (1981) reported the variation of T_g according to heating or cooling rates as follows:

$$T_{g_q} = T_{g_1} + B\log(q) \qquad [4.8]$$

Also Dehoff & Anusavice (1986) calculated T_g for higher cooling rates by the help of the same equation. Where; q is cooling or heating rate, T_{g_1} is the glass transition temperature at 1°C/min cooling ratio and, B is a coefficient determined according to type of porcelain type used.

Among the specimens used in the study of Fairhurst et al. (1981), T_{g_5} value of the porcelain, similar to T_{g_5} of Vita-VMK95's (583°C) has been determined and the coefficient B of this porcelain has been used in this study.

From Equation [4.8], the value of T_g for VMK95 at 1°C/min cooling ratio;

$$T_{g_5} = T_{g_1} + B\log(5) \;\Rightarrow\; T_{g_1} = T_{g_5} - 29.3\log 5 = 583 - 29.3\log 5 \;\Rightarrow\; T_{g_1} \cong 563^\circ C$$

The value of T_g at the cooling rate measured in the viscoelastic region (700°C/min);

$$T_{g_{700}} = T_{g_1} + B\log 700 = 563 + 29.3\log 700 \;\Rightarrow\; T_{g_{700}} \cong 646^\circ C$$

T_s value increases at same rate with T_g as seen in Figure 4.11. For this reason, Equation [4.8] has been also used for calculating T_s.

$$T_{s_5} = T_{s_1} + B\log(5) \;\Rightarrow\; T_{s_1} = T_{s_5} - 29.3\log 5 = 652 - 29.3\log 5 \;\Rightarrow\; T_{s_1} \cong 632^\circ C$$

$$T_{s_{700}} = T_{s_1} + B\log 700 = 632 + 29.3\log 700 \;\Rightarrow\; T_{s_{700}} \cong 715^\circ C$$

After determining T_g and T_s, the creep relaxation curve has been drawn as given in Figure 4.11. The creep rate $\frac{\varepsilon_{cr_i}}{\varepsilon_{o_i}}$ for different T_i temperatures, has been read from this curve and put in Equation [4.6]. Then, the value of E_i has been calculated from this equation. In this way, the modulus of elasticity curve in the viscoelastic region has been determined and plotted.

It can be said that, the analyses performed at the maximum firing temperature gives more precise results. In this case, the viscoplastic behaviour of the porcelain at high temperatures should be taken into consideration. Very low yield stresses and plastic deformations are the cause of low stresses occurred in the viscoplastic region of the porcelain. When the modulus of elasticity curves of the materials that shows the same behaviour with the porcelain in the literature were investigated, it has been seen that the curve in the viscoplastic region can be drawn appositely to the curve in the viscoelastic region. Therefore, in this study, the modulus of elasticity curve in the viscoplastic region has been made to pass at lower levels. By this way, even the plastic deformations have not been calculated, the effects of them on the stresses have been included to the results.

Eventually, the effects of the viscoelastic and viscoplastic behaviours have been reflected into the elastic properties with the help of this method. Thus, the porcelain used in this study has been described as a linear elastic material at all temperatures. The variation of the modulus of elasticity (E) including the viscoelastic and viscoplastic regions of the porcelain has been depicted in Figure 4.10.

CHAPTER FIVE
EXPERIMENTAL AND NUMERICAL STUDIES

5.1 Determination of the Cooling Curves of the Furnace

Modeling of the cooling media completely is important extremely for determining the thermal stresses occurred after the firing process. However, this situation only can be possible with true definition of the convection and radiation coefficients. Non-regular geometry of the crown does not give permission to theoretical calculation of the convection coefficient. Besides, this cooling media comes to a state of more complex condition, due to fact that a great part of the thermal missing happens from the radiation and the crown surfaces enter into a different radiation interaction with the furnace and room media. Because of all of these reasons, defining of the cooling media completely is hard possible. In the result of matching of the cooling curves obtained from the experimental measurements with the cooling curves found from the finite element analyses, it has been seen that the problem can be passed.

So as to be able to define the cooling media completely, the cooling curves of the external and internal surfaces of the crown prepared from Ni-Cr substructure material and VMK 95 has been first detected experimentally. For this reason, a metal-ceramic crown restoration from Ni-Cr substructure and VMK 95 porcelain has been prepared for a central incisor tooth. For the experimental measurements, two thermocouple wires of 0.3 mm diameter and a temperature measurement device which collects the data has been calibrated together for the interval of 20°C-1000°C.

While one of the thermocouple wires has been fixed on the outer surface of the porcelain, the other wire has been fixed on the inner surface of the substructure material (Figure 5.1). Then, the crown has been placed in the furnace in this form. Other ends of the thermocouple wires have been tied to the temperature measurement device outside as seen in Figure 5.2 and this device has been connected to the computer. Since there are no openings to reach the furnace chamber when the furnace lid is closed during the firing process, the central part of the thermocouples remain below the lid.

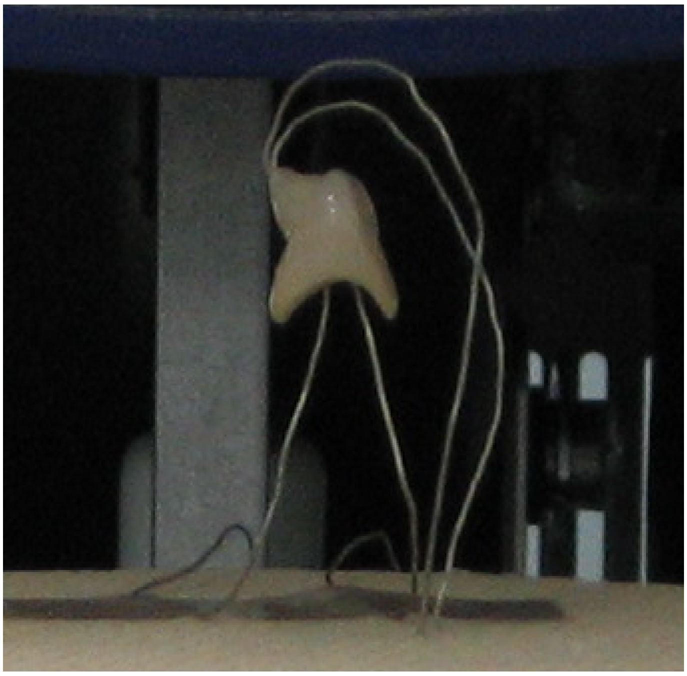


Figure 5.1 The locations of the thermocouples on the crown.

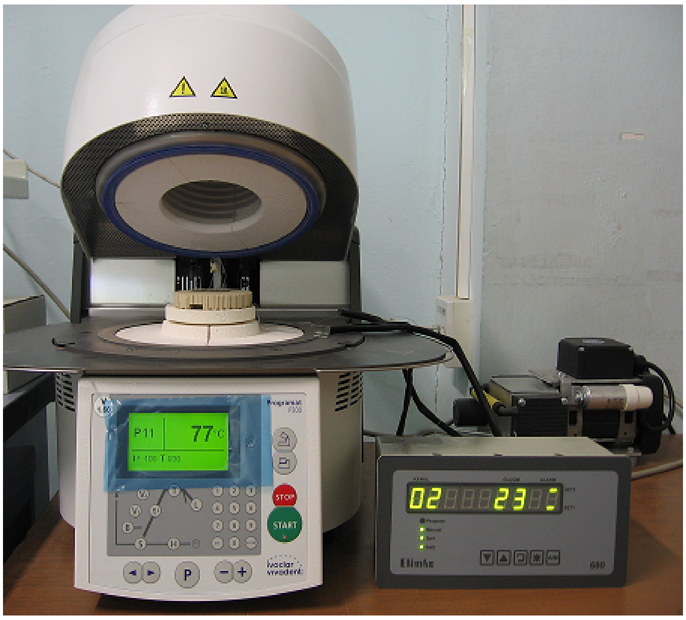


Figure 5.2 The mechanism of the temperature measurement.

Since the thermocouples remain inside the furnace chamber, their sheaths which resist the maximum firing temperature (960°C) are quite thick. In this case, it is impossible to pass the sheathed wires below the lid and perform the process. Since the thermocouples are required to be thin enough for the lid to be closed completely, the thermocouples have been used in unsheathed form. A special care has been given to the surfaces to be insulated in which the thermocouples contact.

Then, the standard firing process used to fire the prepared crown has been carried out on this mechanism prepared. The temperature measurements have been recorded directly on the computer with intervals of 10 sec beginning the opening of the furnace lid at the maximum temperature and the experimental cooling curves have been obtained at the end of the measurements. As seen in Figure 5.3, it has been determined that the cooling curves of the inner and outer surfaces of the crown are very close to each other. The most important factors of obtaining these results are that the dimensions of the crown is small, the thicknesses of the crown and between the surfaces taken the measurement are very thin.

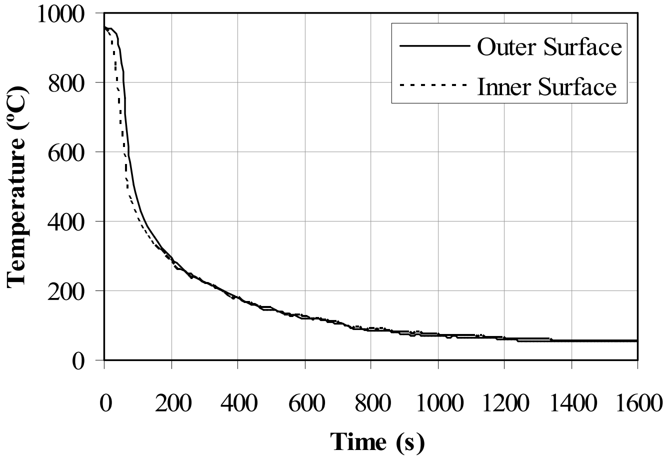


Figure 5.3 Experimental cooling curves obtained from the measurements.

5.2 Calculation of Equivalent Convection Coefficients

In the numerical thermal stress analyses made on the crown, a convection coefficient needs to provide the cooling curve exactly in the furnace. The convection coefficient, including the effects of both the convection in the furnace and the radiation occurred with the outer media after opening the furnace lid, is called as equivalent convection coefficient (h_{eq}).

Because of this, firstly, it has been calculated theoretically by using the experimental cooling curves. The experimental cooling curves depicts exponential function character in the shape of $A.e^{-bt}$. But, the cooling curves obtained from the experimental measurements can not be explained with only an exponential function, owing to the fact that a very rapidly temperature decrease occurs at the beginning o the cooling process. Therefore, the experimental cooling curve has been divided into pieces in between the certain temperatures and each part has been expresses by a different exponential function (Incropera et al., 2007), as seen in Figure 5.4.

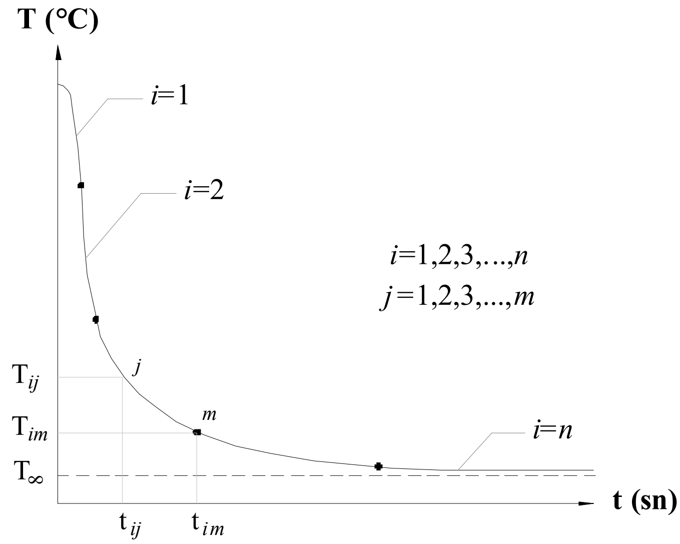


Figure 5.4 Dividing into pieces of the cooling curve.

For each i piece on this graphic;

$$\frac{T_{ij} - T_{\infty}}{T_{im} - T_{\infty}} = \exp(-b_i.t_{ij}) \qquad [5.1]$$

$$b_i = \frac{h_{eş_i} A_s}{V.d_{ij}.C_{p_{ij}}} \qquad [5.2]$$

From Equations [5.1] and [5.2], equivalent convection coefficient (h_{eq}) of each i piece can be obtained as below:

$$h_{eş_i} = \frac{V.d_{ij}.c_{p_{ij}}}{A_s.t_{ij}}.\ln\frac{T_{im} - T_\infty}{T_{ij} - T_\infty} \qquad [5.3]$$

Where, T_{ij}, T_∞, T_{i_m}, A_s, d, V and C_p are called as instant temperature of the piece at certain time, ambient temperature, maximum temperature of the piece, the surface area affected by the convection, density, volume and specific heat, respectively.

Then, the numerical study has been done to understand that whether the convection coefficients obtained by calculating for each pieces reflect to the cooling curve exactly of the furnace in the finite element method (FEM) or not. For this reason, the three dimensional finite element model of the produced crown has been established with 19269 nodes and 12346 couple field elements (Figure 5.5). In this model, the opaque layer has been included in the volume of the porcelain, since its composition and properties are very close to the dentine porcelain.

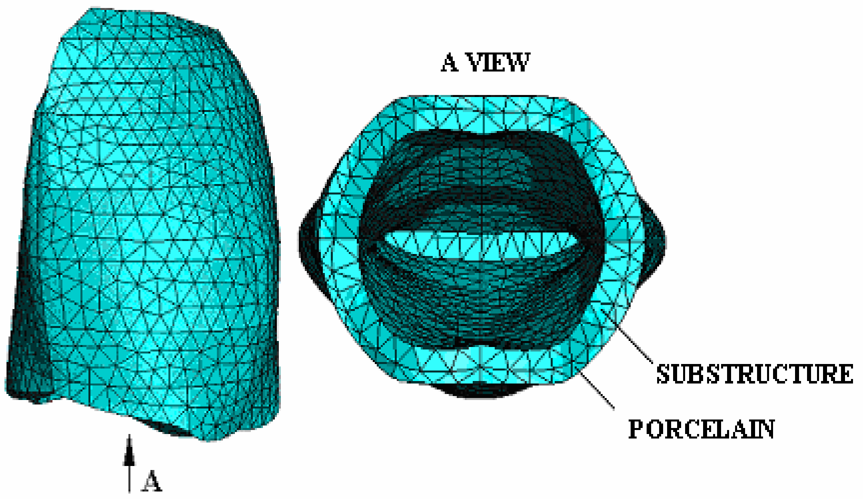
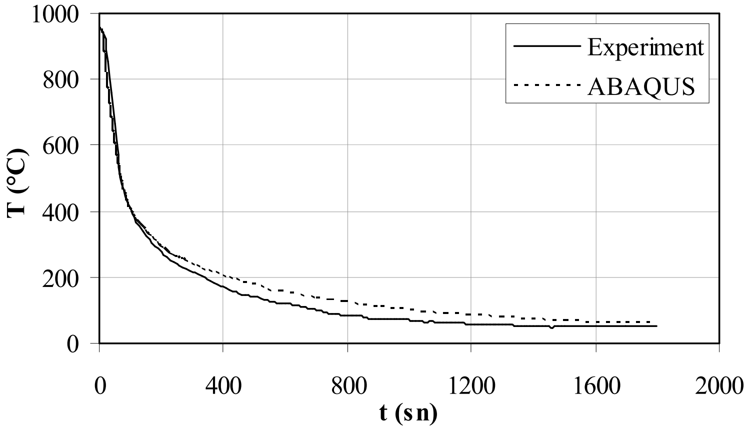
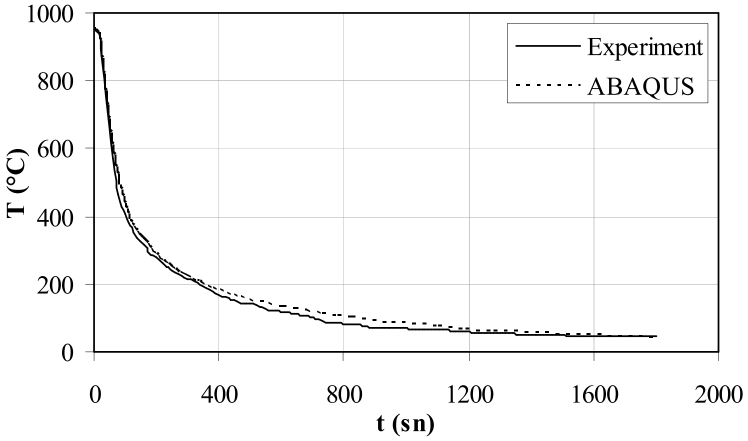


Figure 5.5 Finite element model of the crown used to experimental studies.

The convection coefficients calculated by the help of the experimental cooling curves have been applied to the outer surfaces exposed to air of the finite element model in ABAQUS finite element program. In the result of thermal stress analyses done, the numerical cooling curve has been compared with the experimental cooling curve. At initial, the differences observed between the numerical and experimental cooling curves, as seen in Figure 5.6a, have been reduced to minimum thanks to the adjustments done in the equivalent convection coefficients, and then the equivalent convection coefficients that give a very close curve to the experimental cooling curve have been obtained (Figure 5.6b).



(a)



(b)

Figure 5.6 Comparison of experimental cooling curve with the numerical curves obtained; **(a)** from the first numerical analysis, **(b)** after from the adjustments.

The equivalent convection coefficient values obtained by harmonizing the experimental and numerical cooling curves are given in Table 5.1 for certain temperature values.

Table 5.1 The equivalent convection coefficients for certain temperatures.

T (ºC)	**960**	**375**	**355**	**135**	**125**	**50**
h_{eq} (W/m^{2}K)	23,5	14,5	3,5	2,3	1,4	1,1

In a temperature value among given temperature values, ABAQUS finite element program calculates the value of the equivalent convection coefficient at this temperature by doing the interpolation.

The equivalent convection coefficients, calculated from the experimental and numerical studies, includes both convection and radiation effects, at the same time. Hence the elapsed time until that the temperature of the furnace reaches from its maximum value to room temperature can be modeled as numerically. Finally, the thermal stress analyses, done by this method, give more accurate results because of providing the real media in the furnace.

5.3 Thermal Stress Analyses in ABAQUS

In this study, ABAQUS finite element program has been used for thermal stress analyses. During the analyses, some steps are followed in the finite element program.

Firstly, the three dimensional (3D) finite element models have been established in *Part module*. These models are in 3D modeling space, deformable type and solid shape. In these models, the opaque layer has been included in the volume of the porcelain, since its composition and properties are very close to the dentine porcelain. Therefore, basically, it has been used two volumes named as substructure and porcelain.

In *Property module*, the physical, thermal and mechanical properties of the substructure materials and porcelain have been entered to the model. In this phase, ABAQUS finite element program allows to be entered the material properties depending on temperature. After entering the material properties, these properties have been defined to the related volumes. Then, *Assembly module* has been used to create the assembly from the volumes.

In *Step module*, the type of the analysis, the time of the analysis and the number of incerementations can be entered. For this study, the type of the analysis has been chosen as *Coupled temp-displacement* owing to the fact that it is the stress analysis depending on temperature. According to the experimental cooling curves, the time of the analysis has been chosen as *1800 sn*. Also the maximum number of the increments has been taken as 1000.

In *Interaction module*, the equivalent convection coefficient values depending on temperature obtained from the experimental and numerical studies have been entered to the inner and outer surfaces of the models. Thus, the thermal stress analyses give more accurate results because of providing the real media in the furnace.

In *Load module*, due to the fact that the analyses done are thermal stress analyses, any load has been entered to the model. But, in this step, the starting temperature of the cooling process has been defined on the inner and outer surfaces of the model according to the cooling curves. In other words, the starting temperature of the cooling process has been entered as boundary condition of the analyses.

In *Mesh module*, the crown model produced has been divided to finite elements. There are 19269 nodes and 12346 couple field elements on the model divided to finite elements. The element type has been chosen as suitable for the thermal stress analysis. The element chosen (C3D10M) is a 10-node modified thermally coupled second-order tetrahedron and hourglass control element.

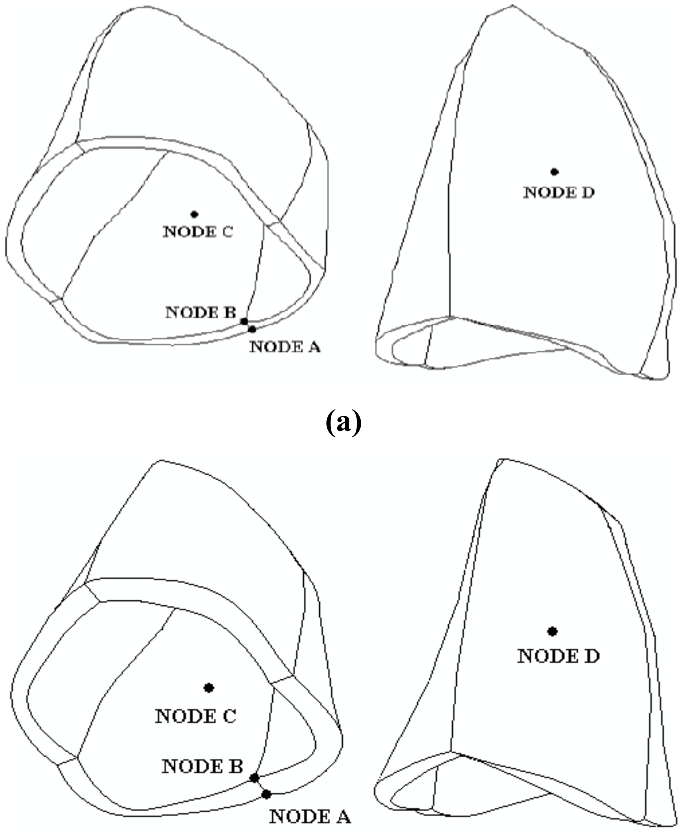
In Job module, the analysis which will be done has been named and started the finite element analysis. Also, errors and warnings occurred during the analyses can be followed in this module. In Visualization module provides graphical display of finite element models and results. XY data curves, for example Stress-Temperature curve, can be drawn wit the help of the result information from the output database.

In the result of these analyses, thermal stresses depending on the time and temperature have been obtained and drawn graphically. Also, the residual stresses on the substructures and porcelain, and the equivalent plastic strains on the substructures have been determined. The thermal stresses remained on the crown at the end of the cooling process are called as the residual stresses.

CHAPTER SIX
RESULTS AND DISCUSSIONS

6.1 Results Obtained from Finite Element Analyses

In the result of all of the transient thermal stress analysis, it has been seen that the maximum stresses happen at the interfaces of the substructure and porcelain, especially highly concentrated at the cervical margin, mesial and distal corners of the crown. This observation is supported by most of the clinical studies (Papazoglou & Brantley, 1998; Ozcan & Niedermeier, 2002; Ozcan et al., 2006.). In their studies, it has been reported that the cracks are seen at the cervical region in veneering material. Therefore, in this study, the transient and residual stress values has been taken the certain points as seen in Figure 6.1a, b.



(a)

(b)

Figure 6.1 Location of the critical nodes on the finite element model, **(a)** porcelain, **(b)** substructure material.

6.1.1 Thermal stresses of Ni-Cr/Porcelain crown

6.1.1.1 Thermal stresses according to α1 for Ni-Cr/VMK95

The results of the transient thermal stress analysis for Ni-Cr/VMK 95 crown are given in Figures 6.2 (a, b) and 6.3 (a, b) depending on time and temperature. In this analysis, the thermal expansion coefficient values at high temperatures of the porcelain have been taken as constant.

Figure 6.2a and b depicts the thermal stresses in porcelain depending on time and temperature, respectively. As seen in these figures, the maximum stress occurs at node B and its value is 110 MPa, approximately. At the beginning of the cooling process, the stresses reach to the maximum values at all nodes investigated.

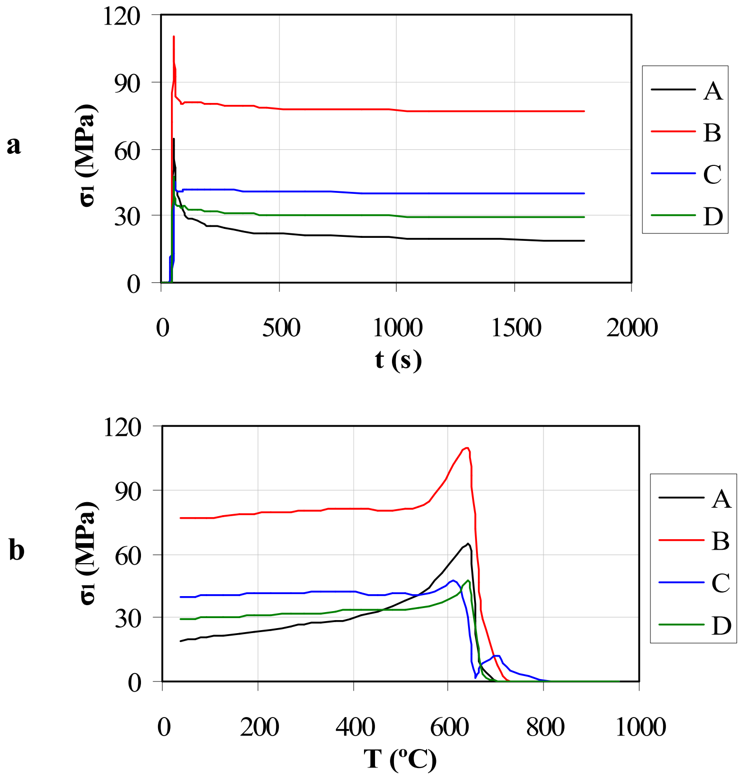


Figure 6.2 Thermal stresses in porcelain depending on; **(a)** time, **(b)** temperature.

Figure 6.3a and b shows the thermal stresses in Ni-Cr substructure depending on time and temperature, respectively. The maximum stress occurs at node A interacted to Node B of the porcelain. The maximum value of the thermal stress is 175 MPa, approximately. Similarly to porcelain, at the beginning of the cooling process, the stresses reach to the maximum values at all nodes investigated.

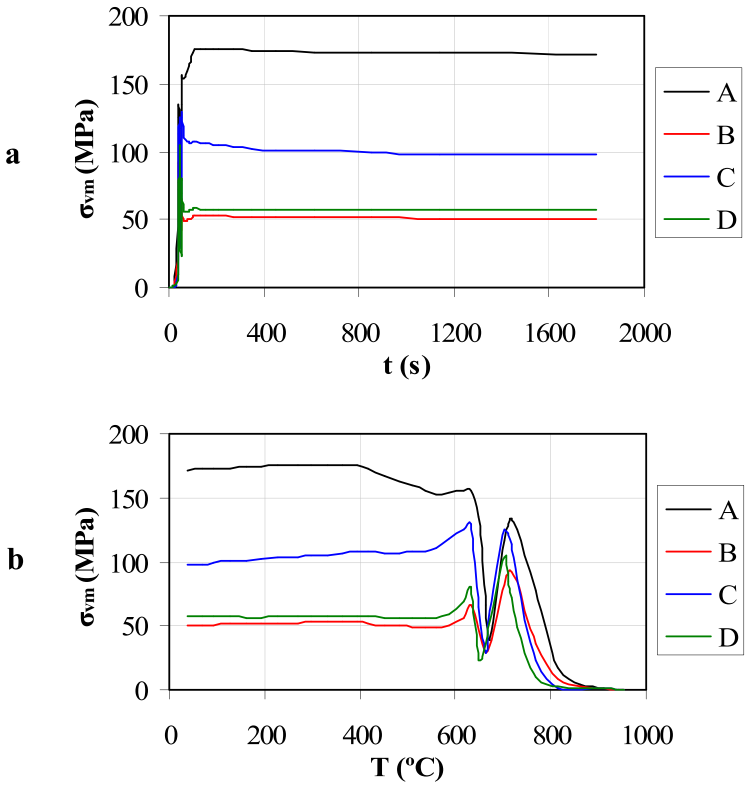
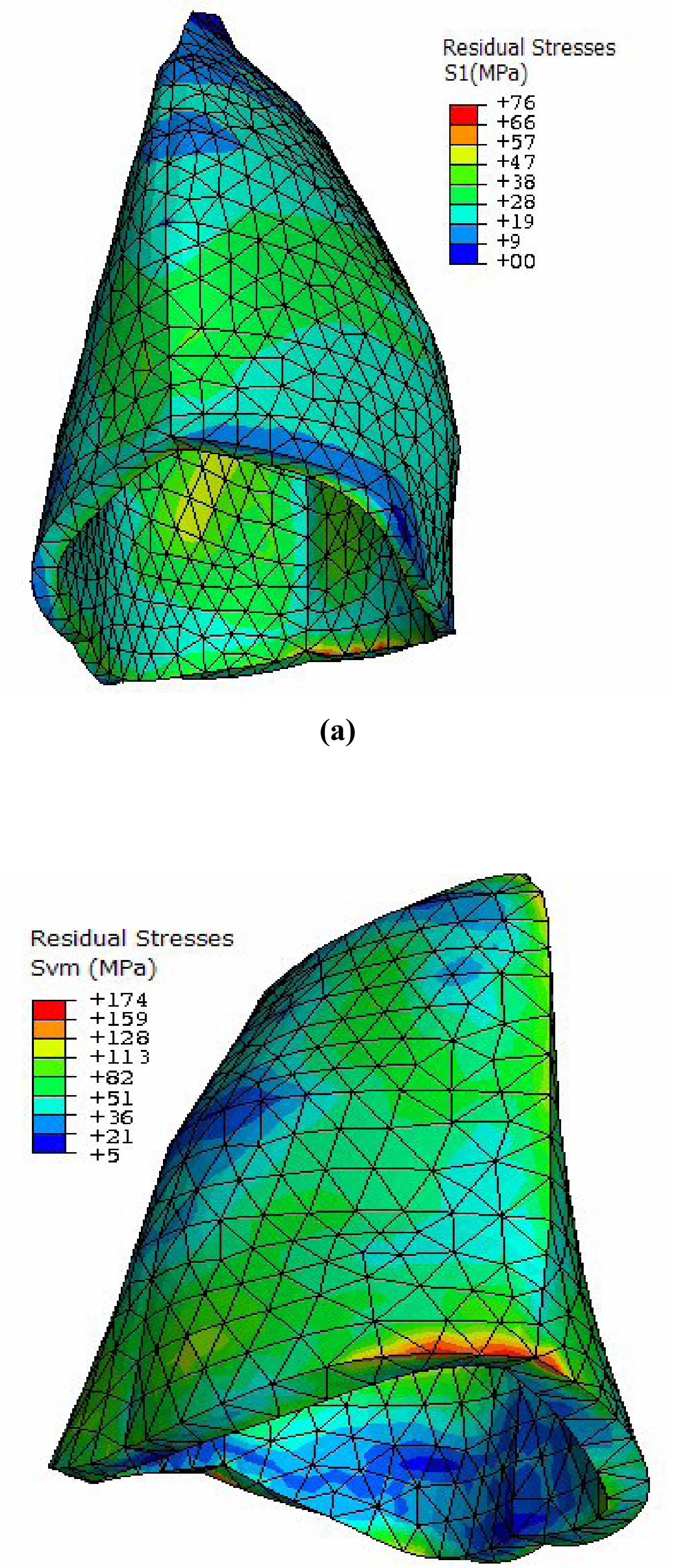


Figure 6.3 Thermal stresses in Ni-Cr depending on; **(a)** time, **(b)** temperature.

After the cooling process, the stresses remained on the crown are called as residual stresses. Figure 6.4a and b points out the residual stresses remained on the porcelain and Ni-Cr substructure, respectively. As seen in these figures, the residual stresses intensify at node B of porcelain and at node A of Ni-Cr. The values of them for porcelain and Ni-Cr are 76 MPa and 174 MPa, respectively.



(a)

(b)

Figure 6.4 Residual stresses for α1 remained on; **(a)** porcelain, **(b)** Ni-Cr.

6.1.1.2 Thermal stresses according to α2 for Ni-Cr/VMK95

The transient thermal stress values obtained according to α2 for Ni-Cr/VMK 95 crown are given in Figures 6.5 (a, b) and 6.6 (a, b) depending on time and temperature. In this analysis, the thermal expansion coefficient values at high temperatures of the porcelain have been taken in the increased form.

Figure 6.5a and b points out the thermal stresses in porcelain depending on time and temperature, respectively. As seen in these figures, the maximum stress occurs at node B and its value is 210 MPa, approximately. At the beginning of the cooling process, the stresses reach to the maximum values at all nodes investigated.

As seen in Figures 6.2 and 6.5, since the thermal expansion coefficient which has the increased values at the high temperature are used in the analyses, it is seen that the maximum thermal stresses increase.

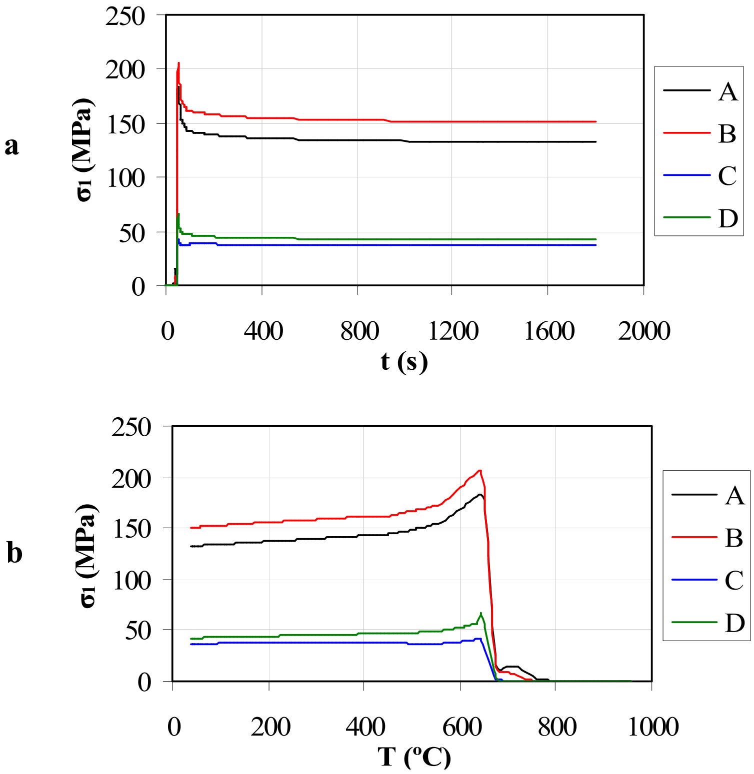


Figure 6.5 Thermal stresses in porcelain depending on; **(a)** time, **(b)** temperature.

Figure 6.6a and b depicts the thermal stresses in Ni-Cr substructure depending on time and temperature for α2, respectively. The maximum stress occurs at node A interacted to node B of the porcelain. The maximum value of the thermal stress is 230 MPa, approximately. Similarly to porcelain, at the beginning of the cooling process, the stresses reach to the maximum values at all nodes investigated.

If Figure 6.6 compares with Figure 6.3, it can be seen that the maximum thermal stresses increase by using the thermal expansion coefficient which has the increased values at the high temperature.

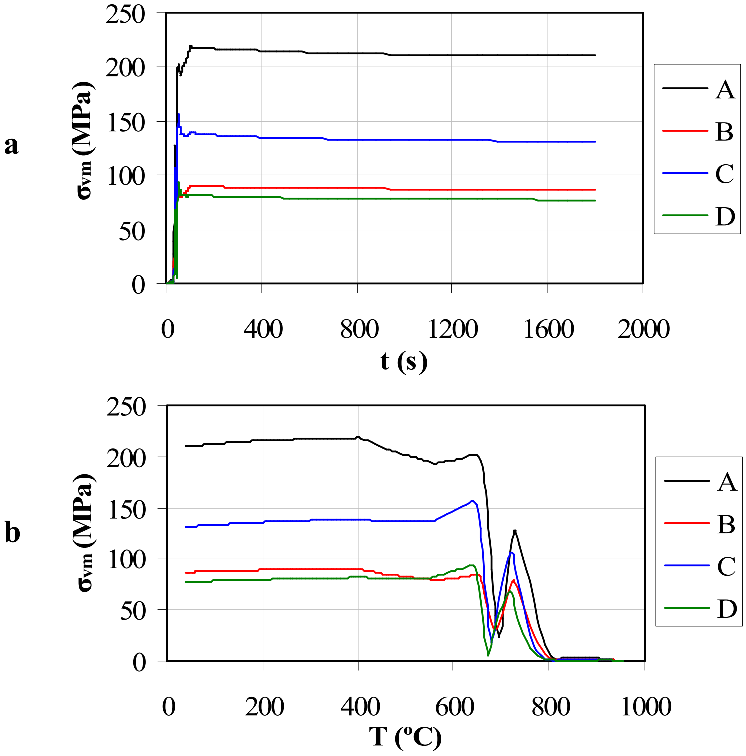
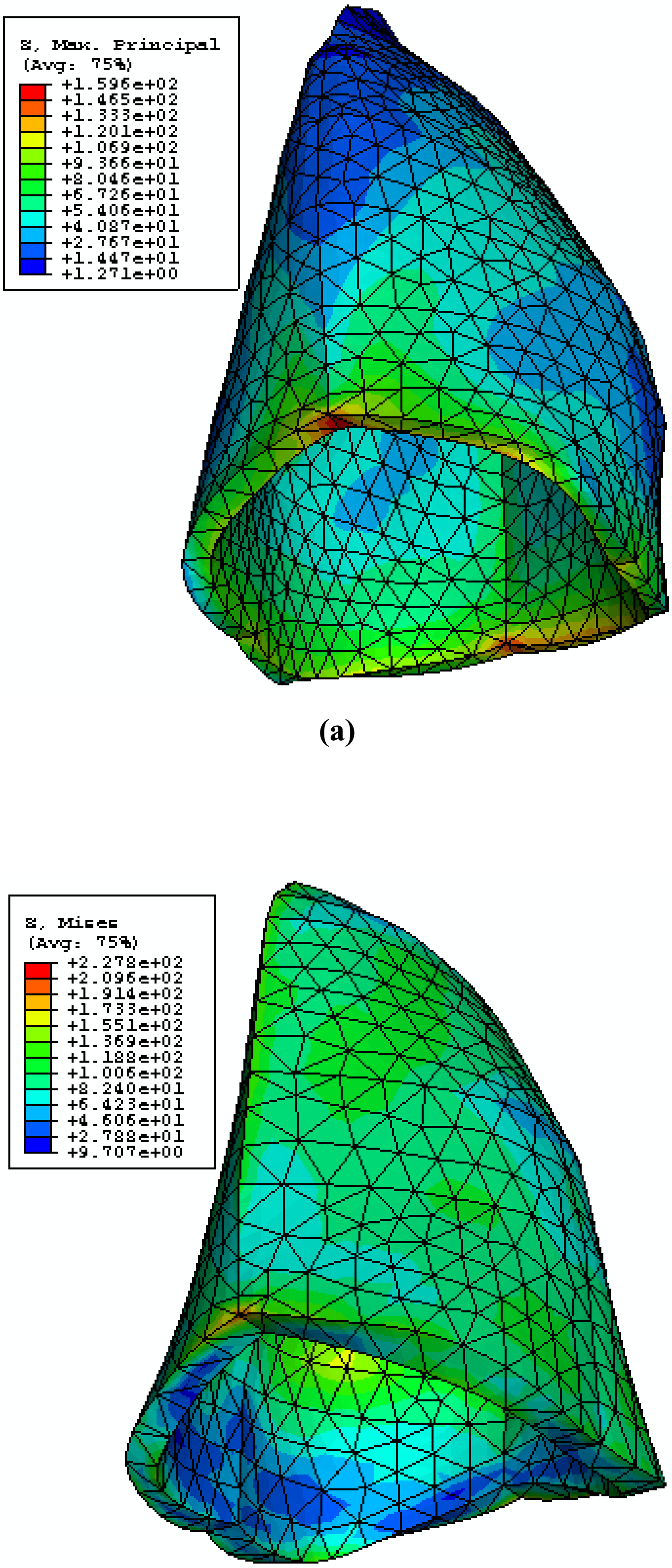


Figure 6.6 Thermal stresses in Ni-Cr depending on; **(a)** time, **(b)** temperature.

Figure 6.7a and b shows the residual stresses remained on the porcelain and Ni-Cr substructure for α2, respectively. As seen in these figures, the residual stresses intensify at node B of porcelain and at node A of Ni-Cr. The values of them for porcelain and Ni-Cr are 160 MPa and 228 MPa, respectively. Similarly, it is seen that the residual stresses also increase by using α2.



(a)

(b)

Figure 6.7 Residual stresses for α2, remained on; **(a)** porcelain, **(b)** Ni-Cr.

6.1.2 Thermal stresses of Co-Cr/Porcelain crown

6.1.2.1 Thermal stresses according to α1 for Co-Cr/VMK95

When the results of Co-Cr/VMK 95 crown’s transient thermal stress analysis were investigated, it has been seen that the maximum stresses occurred on the porcelain and Co-Cr substructure were intensified at the inner corner points of the porcelain and outer corner points of Co-Cr, as seen in Figure 6.10a and b, respectively. Therefore, nodes E and F have been also added to investigations done with α1 for Co-Cr/VMK 95 crown.

Figure 6.8a and b depicts the thermal stresses in porcelain depending on time and temperature, respectively. As seen in these figures, the maximum stress occurs at node E and its value is 158 MPa, approximately.

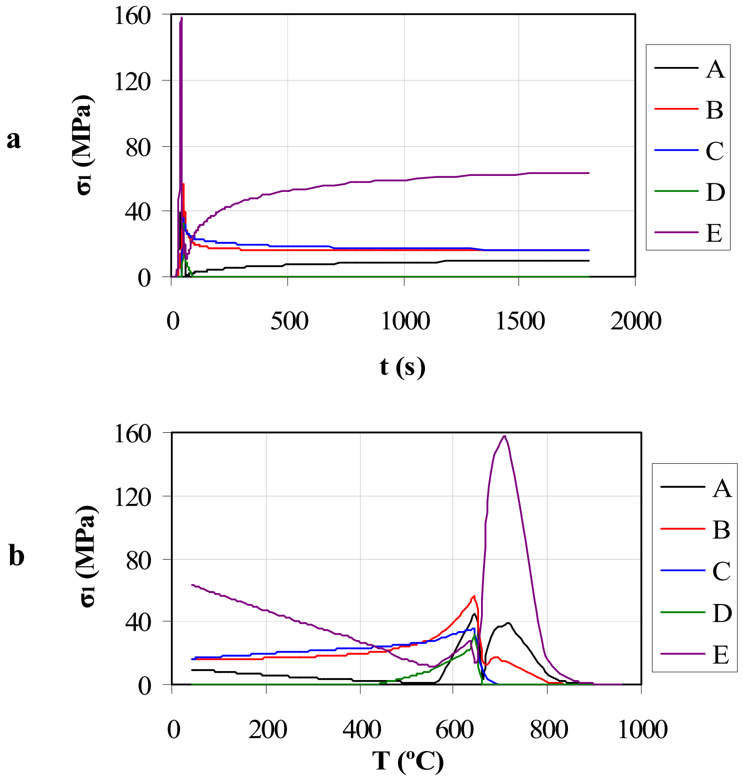


Figure 6.8 Thermal stresses in the porcelain of Co-Cr/VMK95 crown for α1 depending on; **(a)** time, **(b)** temperature.

Figure 6.9a and b shows the thermal stresses in Co-Cr substructure depending on time and temperature, respectively. The maximum stress occurs at node F interacted to Node E of the porcelain. The maximum value of the thermal stress at this node is 329 MPa.

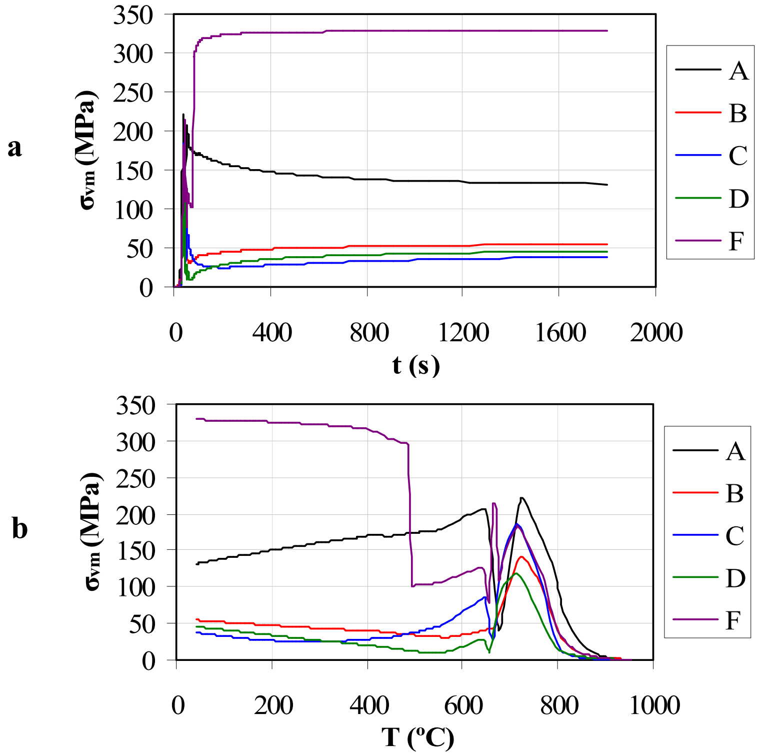
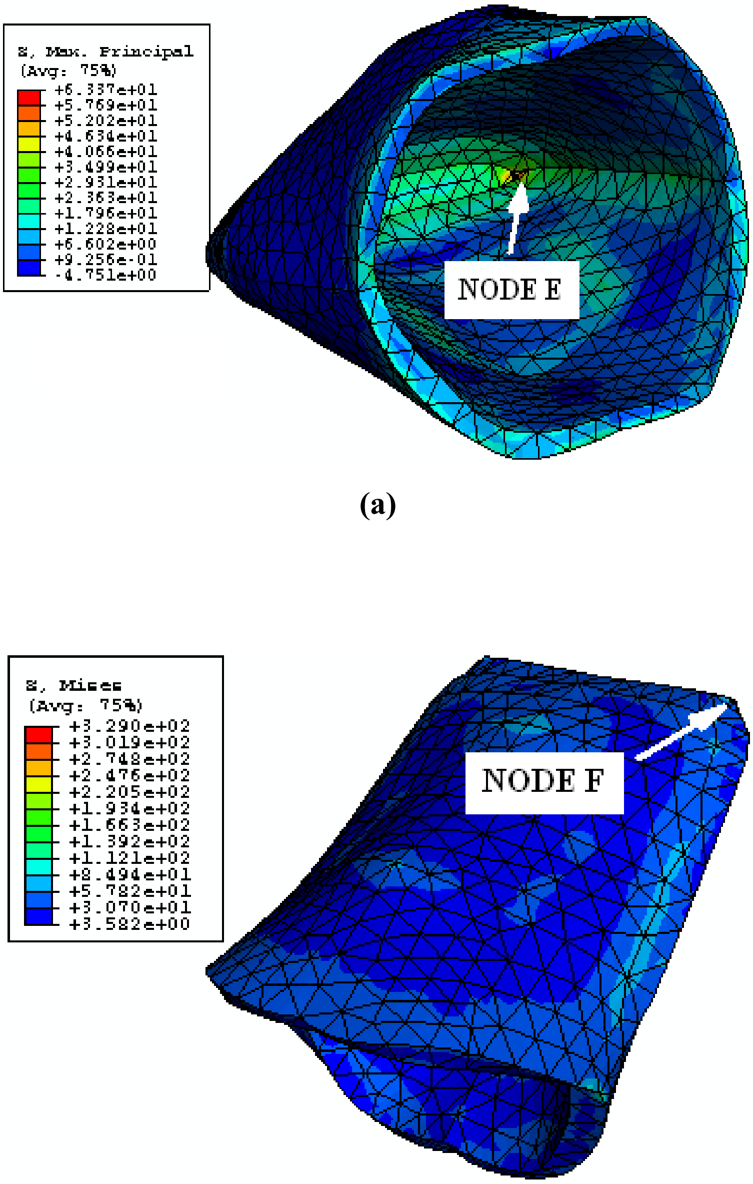


Figure 6.9 Thermal stresses in Co-Cr of Co-Cr/VMK95 crown for α1 depending on; **(a)** time, **(b)** temperature.

In this analysis, looking at the general of the analysis, it is seen that the maximum value of the thermal stresses are collected in these E and F nodes. However, this situation can be based on the problems from the model. In addition, it is explain that the fractures happen in the cervical margin, mesial and distal corners of the crown in the clinical studies. Therefore, the maximum stress values at node B for porcelain and node A for Co-Cr, considered in the clinical studies, must be investigated.

The maximum stress values at these nodes are 56 MPa at node B of the porcelain and 221 MPa at node A of Co-Cr.

Figure 6.10a and b points out the residual stresses remained on the porcelain and Co-Cr substructure, respectively. As seen in these figures, the residual stresses intensify at node E of porcelain and at node F of Co-Cr. The values of them for porcelain and Co-Cr are 64 MPa and 329 MPa, respectively. For nodes B and A, these values are 16 MPa and 132 MPa, respectively.



(a)

(b)

Figure 6.10 Residual stresses for α1, remained on; **(a)** porcelain, **(b)** Co-Cr.

6.1.2.2 Thermal stresses according to α2 for Co-Cr/VMK95

The transient thermal stress values obtained according to α2 for Co-Cr/VMK 95 crown are given in Figures 6.11 (a, b) and 6.12 (a, b) depending on time and temperature. In this analysis, the thermal expansion coefficient values at high temperatures of the porcelain have been taken in the increased form.

Figure 6.11a and b points out the thermal stresses in porcelain depending on time and temperature, respectively. As seen in these figures, the maximum stress occurs at node B and its value is 196 MPa, approximately. At the beginning of the cooling process, the stresses reach to the maximum values at all nodes investigated.

As seen in Figures 6.8 and 6.11, since the thermal expansion coefficient which has the increased values at the high temperature is used in the analyses, it is seen that the maximum thermal stresses increase.

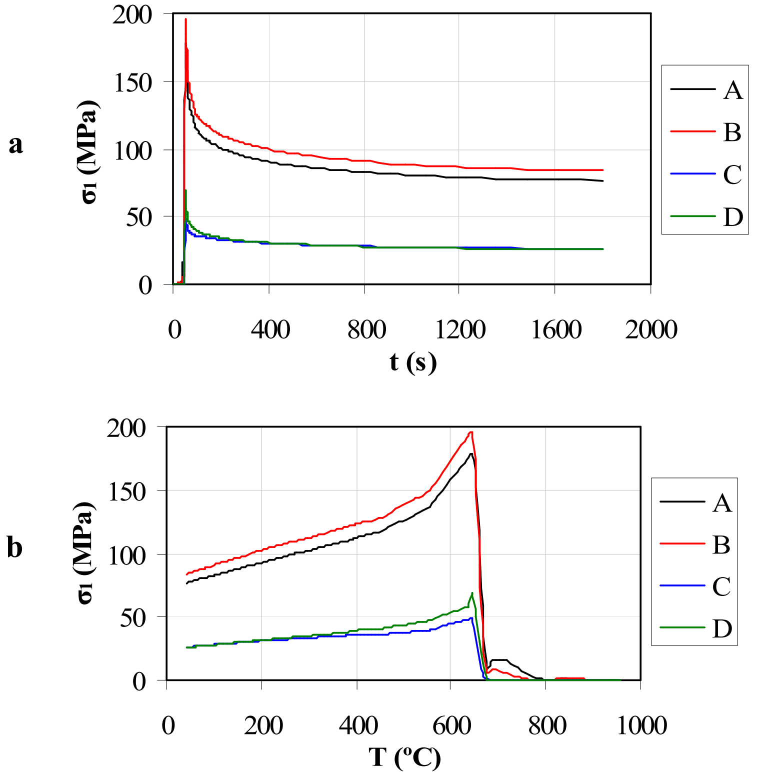


Figure 6.11 Thermal stresses in the porcelain of Co-Cr/VMK95 crown for α2 depending on; **(a)** time, **(b)** temperature.

Figure 6.12a and b depicts the thermal stresses in Co-Cr substructure depending on time and temperature for α2, respectively. As seen in this figures, the maximum stress occurs at node A interacted to node B of the porcelain. The maximum value of the thermal stress is 249 MPa, approximately. Similarly to porcelain, at the beginning of the cooling process, the stresses reach to the maximum values at all nodes investigated.

If Figure 6.12 compares with Figure 6.9, it can be seen that the maximum thermal stresses for node A increase by using the thermal expansion coefficient which has the increased values at the high temperature.

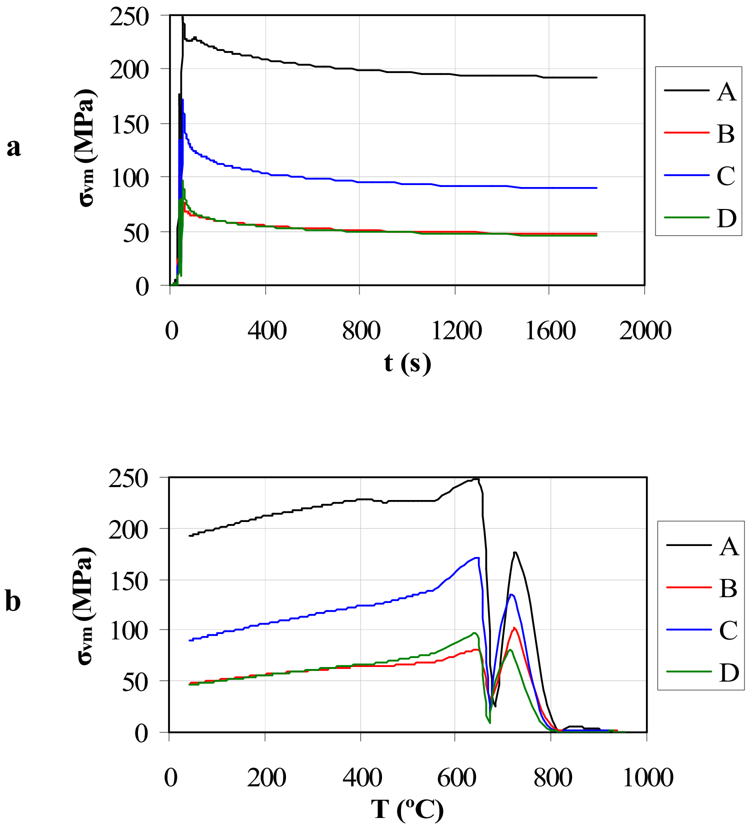
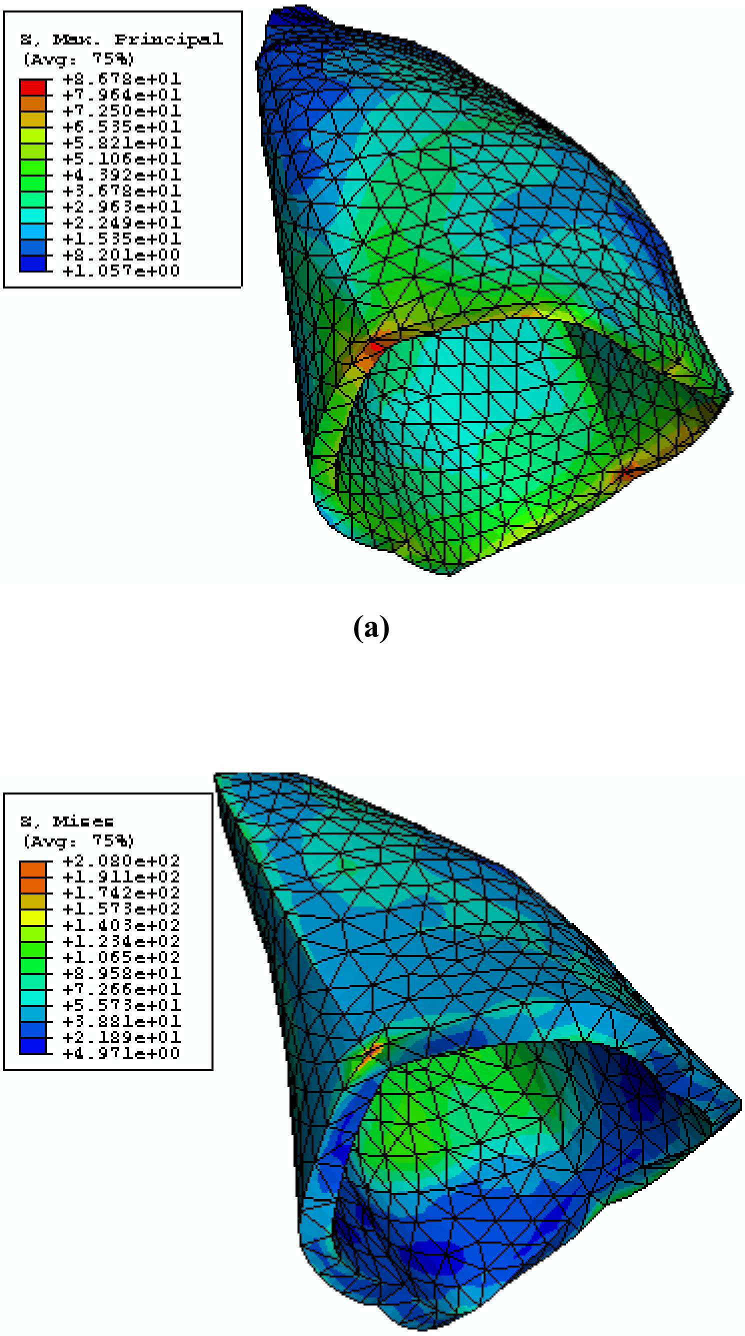


Figure 6.12 Thermal stresses in Co-Cr of Co-Cr/VMK95 crown for α2 depending on; **(a)** time, **(b)** temperature.

Figure 6.13a and b shows the residual stresses remained on the porcelain and Co-Cr substructure for α2, respectively. As seen in these figures, the residual stresses intensify at node B of porcelain and at node A of Co-Cr. The values of them for porcelain and Co-Cr are 87 MPa and 192 MPa, respectively. Similarly, it is seen that the residual stresses also increase by using α2.



(a)

(b)

Figure 6.13 Residual stresses for α2, remained on; **(a)** porcelain, **(b)** Co-Cr.

6.1.3 Thermal stresses of Pd-Ag-Au/Porcelain crown

6.1.3.1 Thermal stresses according to α1 for Pd-Ag-Au/VMK95

When the results of Pd-Ag-Au/VMK 95 crown's transient thermal stress analysis were investigated, it has been again seen that the maximum stresses occurred on the porcelain and Pd-Ag-Au substructure were intensified at the inner corner points of the porcelain and outer corner points of Pd-Ag-Au alloy, as seen in Figure 6.16a and b, respectively. Therefore, additionally, nodes E and F have been included to investigations done with α1 for Pd-Ag-Au/VMK 95 crown.

Figure 6.14a and b depicts the thermal stresses in porcelain depending on time and temperature, respectively. As seen in these figures, the maximum stress occurs at node E and its value is 135 MPa, approximately.

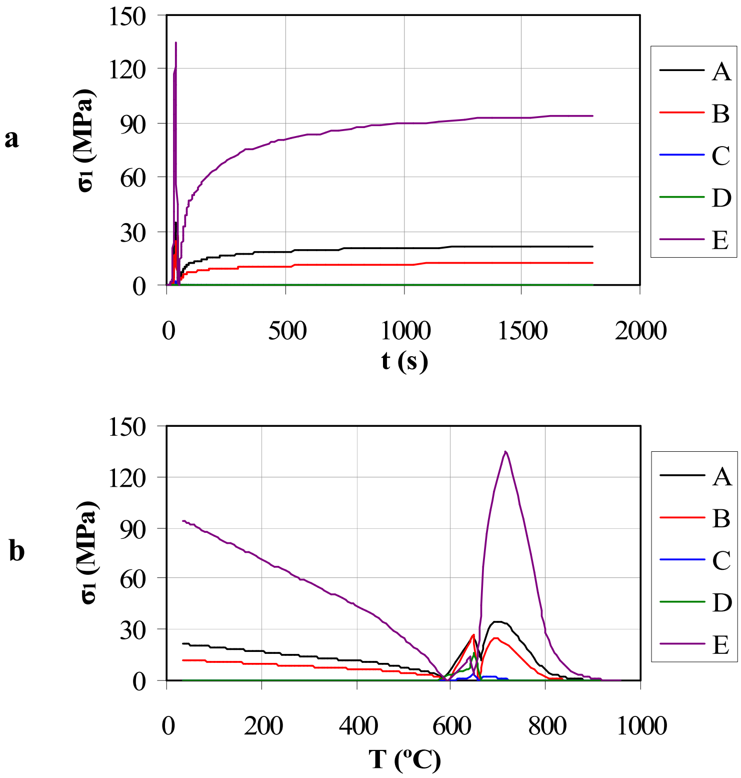


Figure 6.14 Thermal stresses in the porcelain of Pd-Ag-Au/VMK95 crown for α1 depending on; **(a)** time, **(b)** temperature.

Figure 6.15a and b shows the thermal stresses in Pd-Ag-Au substructure depending on time and temperature, respectively. The maximum stress occurs at node F interacted to Node E of the porcelain. The maximum value of the thermal stress at this node is 135 MPa.

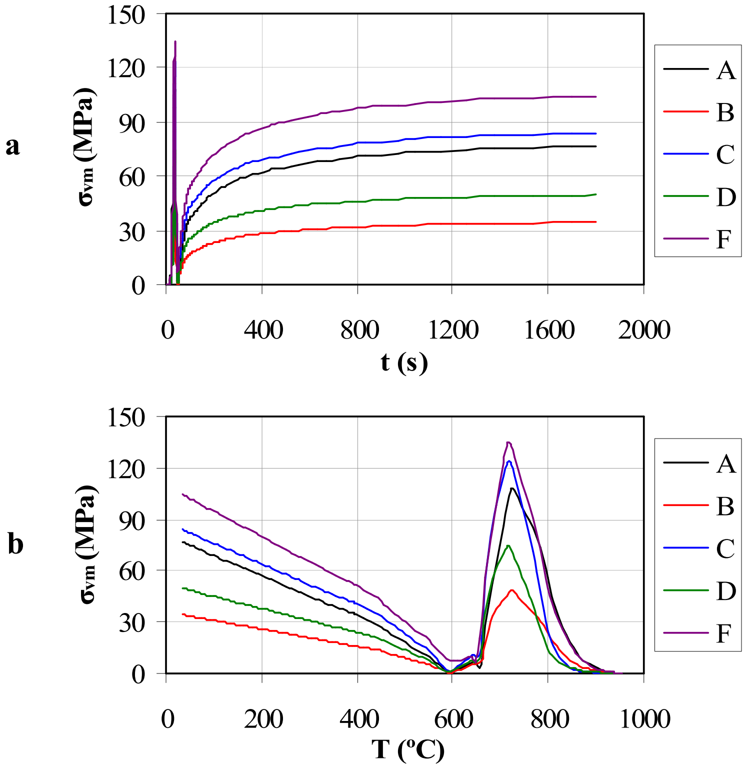
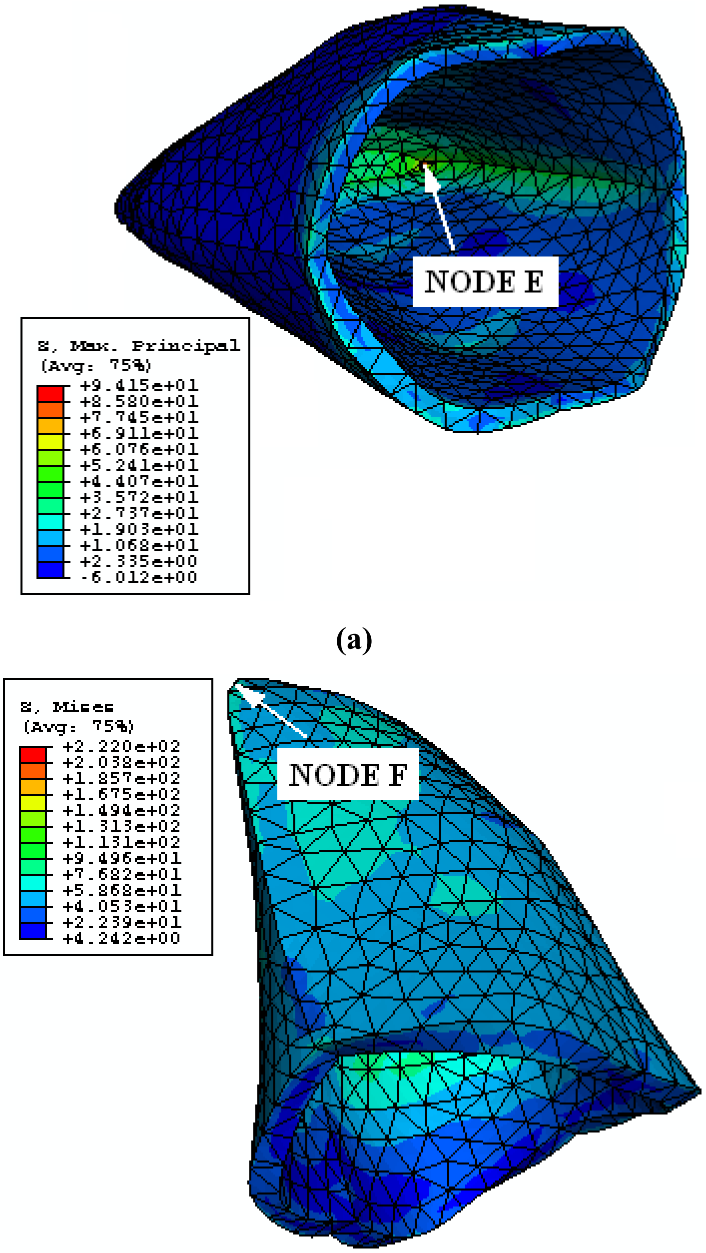


Figure 6.15 Thermal stresses in Pd-Ag-Au of Pd-Ag-Au/VMK95 crown for α1 depending on; **(a)** time, **(b)** temperature.

Considering the general of this analysis, it is seen that the maximum value of the thermal stresses are collected in these E and F nodes. However, this situation can be based on the problems from the model. Because of this situation, the maximum stress values at node B for porcelain and node A for Pd-Ag-Au, considered in the clinical studies, must be taken into account.

The maximum stress values at these nodes are 27 MPa at node B of the porcelain and 108 MPa at node A of Pd-Ag-Au.

In Figure 6.16a and b, it is seen the residual stresses remained on the porcelain and Pd-Ag-Au substructure, respectively. The residual stresses intensify at node E of porcelain and at node F of Pd-Ag-Au. The values of them for porcelain and Pd-Ag-Au are 94 MPa and 104 MPa, respectively. Whereas, for nodes B and A, these values are 12 MPa and 77 MPa, respectively.



(a)

(b)

Figure 6.16 Residual stresses for α1, remained on; **(a)** porcelain, **(b)** Pd-Ag-Au.

6.1.3.2 Thermal stresses according to α2 for Pd-Ag-Au/VMK95

Figure 6.17a and b shows the thermal stresses in porcelain depending on time and temperature for α2, respectively. As seen in these figures, the maximum stress occurs at node B and its value is 155 MPa, approximately.

From the comparison of Figure 6.14 and 6.17, since the thermal expansion coefficient which has the increased values at the high temperature is used in the analyses, it is seen that the maximum thermal stresses increase.

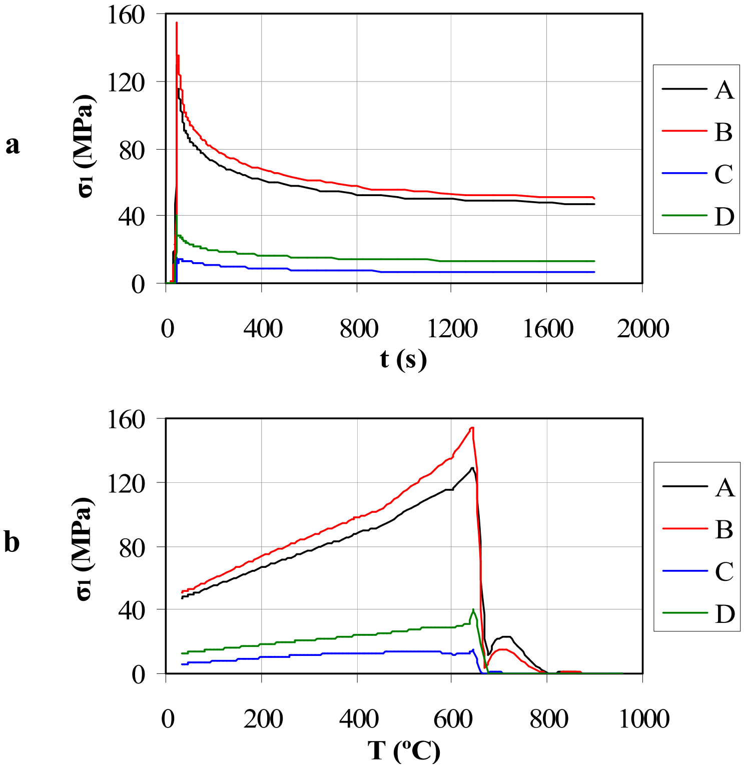


Figure 6.17 Thermal stresses in the porcelain of Pd-Ag-Au/VMK95 crown for α2 depending on; **(a)** time, **(b)** temperature.

Figure 6.18a and b points out the thermal stresses in Pd-Ag-Au substructure depending on time and temperature for α2, respectively. As seen in this figures, the maximum stress occurs at node F. The maximum value at this node is 260 MPa, approximately. In addition, the maximum thermal stress value at node A interacted to node B of the porcelain is 49 MPa.

If Figure 6.18 compares with Figure 6.15, it can be seen that the maximum thermal stresses at nodes A and F decrease and increase by using α2, respectively.

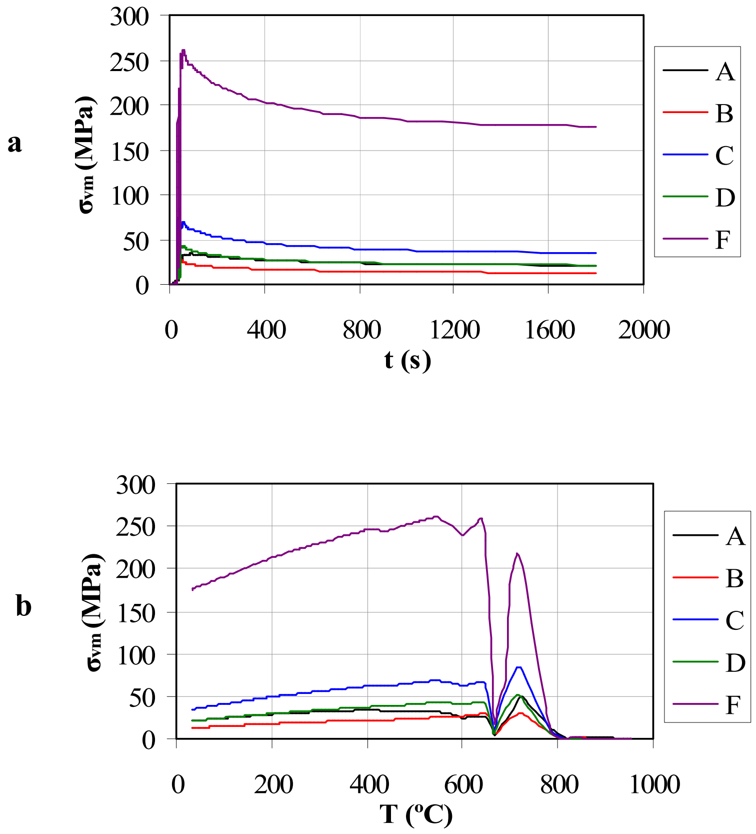


Figure 6.18 Thermal stresses in Pd-Ag-Au of Pd-Ag-Au/VMK95 crown for α2 depending on; **(a)** time, **(b)** temperature.

Figure 6.19a and b depicts the residual stresses remained on the porcelain and Pd-Ag-Au substructure for α2, respectively. As seen in these figures, the residual stresses intensify at node B of porcelain and at node F of Pd-Ag-Au. The values of them for porcelain and Pd-Ag-Au are 53 MPa and 176 MPa, respectively. Besides, the residual stress value at node A of Pd-Ag-Au is 21 MPa.

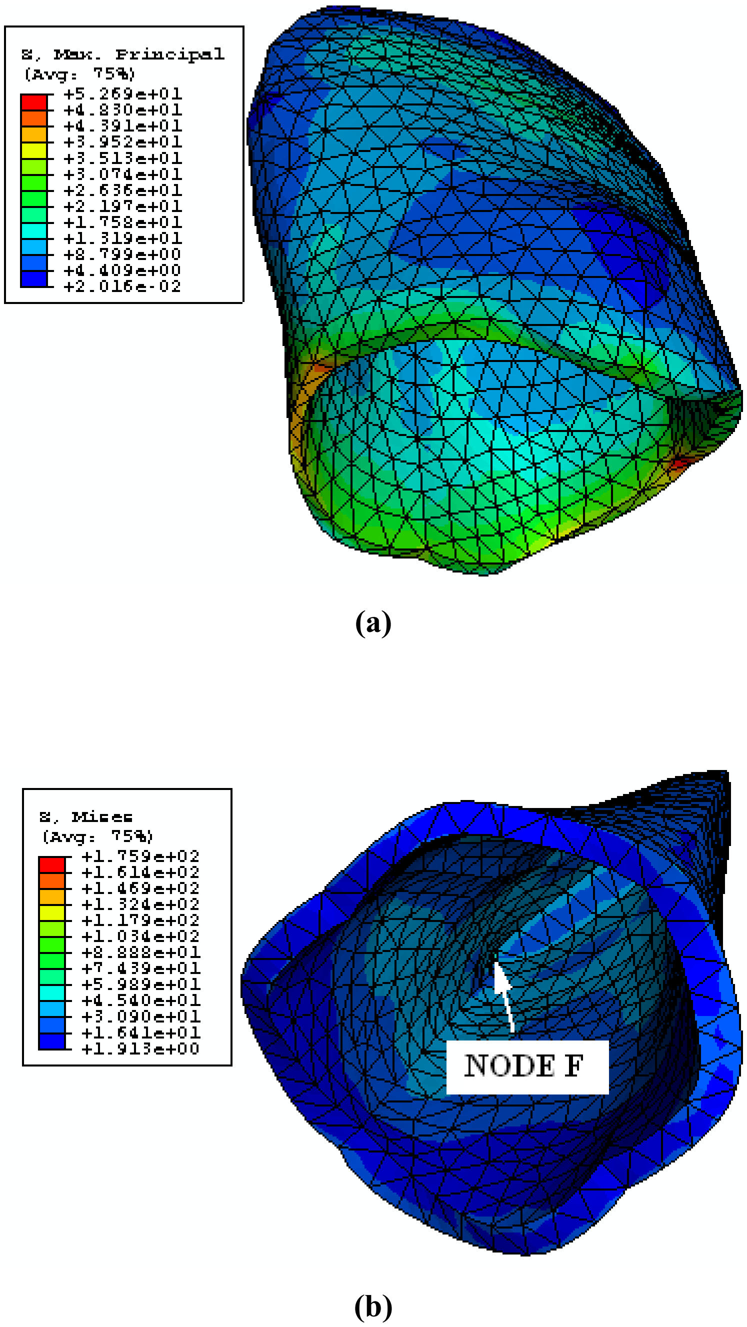


Figure 6.19 Residual stresses for α2, remained on; **(a)** porcelain, **(b)** Pd-Ag-Au.

6.1.4 Thermal stresses of Au-Pt/Porcelain crown

6.1.4.1 Thermal stresses according to α1 for Au-Pt/VMK95

Since the results of Au-Pt/VMK 95 crown's transient thermal stress analysis were investigated, the maximum stresses occur at the inner corner points of the porcelain and outer corner points of Au-Pt alloy, as seen in Figure 6.22a and b, respectively. Because of this, nodes E and F have been included to analyses done with α1 for Au-Pt /VMK 95 crown.

The thermal stresses, obtained in porcelain depending on time and temperature, are given in Figure 6.20a and b, respectively. As seen in these figures, the maximum stress happens at node E and its value is 138 MPa.

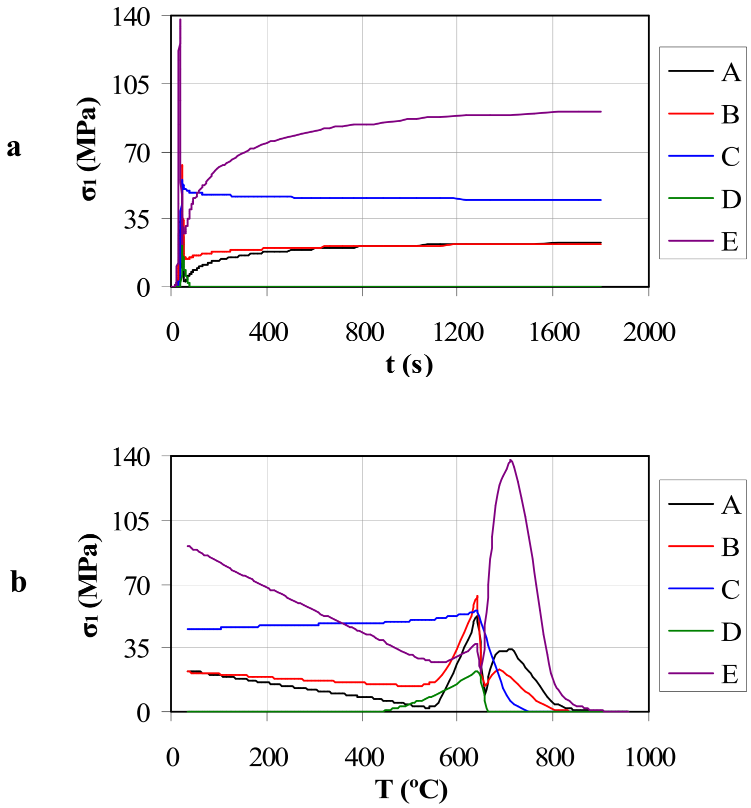


Figure 6.20 Thermal stresses in the porcelain of Au-Pt/VMK95 crown for α1 depending on; **(a)** time, **(b)** temperature.

Figure 6.21a and b depicts the thermal stresses in Au-Pt substructure depending on time and temperature, respectively. The maximum stress occurs at node F interacted to Node E of the porcelain. The maximum value of this thermal stress is 155 MPa.

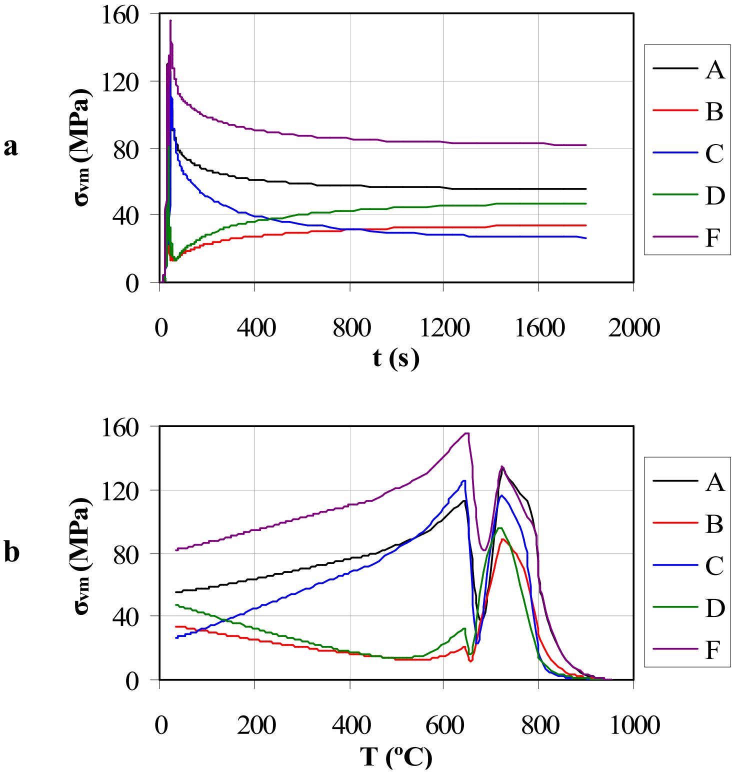
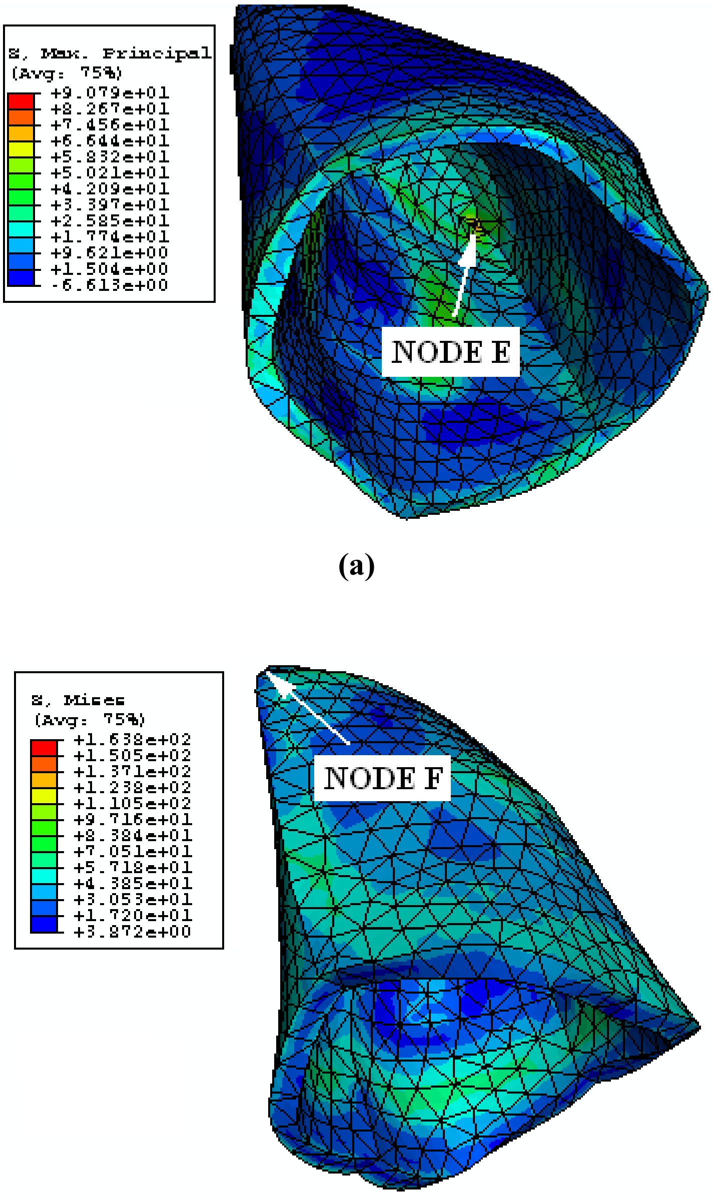


Figure 6.21 Thermal stresses in Au-Pt of Au-Pt/VMK95 crown for α1 depending on; **(a)** time, **(b)** temperature.

If this analysis is considered generally, it will be seen that the maximum value of the thermal stresses are intensified in E and F nodes. However, the problems from the model can cause to this situation. For this reason, the maximum stress values at node B for porcelain and node A for Au-Pt, considered in the clinical studies, must be considered.

The maximum stress values at these nodes are 62 MPa at node B of the porcelain and 134 MPa at node A of Au-Pt.

Figure 6.22a and b shows the residual stresses remained on the porcelain and Au-Pt substructure, respectively. The residual stresses collect at node E of porcelain and at node F of Au-Pt. The values of them are 91 MPa and 82 MPa for porcelain and Au-Pt, respectively. Also, for node B of the porcelain and node A of Au-Pt put forward in the clinical studies, these values are 22 MPa and 55 MPa, respectively.



(a)

(b)

Figure 6.22 Residual stresses for α1, remained on; **(a)** porcelain, **(b)** Au-Pt.

6.1.4.2 Thermal stresses according to α2 for Au-Pt/VMK95

Figure 6.23a and b displays the thermal stresses in porcelain depending on time and temperature for α2, respectively. The maximum stress occurs at node B of the porcelain structure, 199 MPa, as seen in these figures.

If Figures 6.20 and 6.23 compare in terms of the maximum thermal stresses, it will be seen that the maximum thermal stresses increase, since the thermal expansion coefficient which has the increased values at the high temperature is used in the analyses.

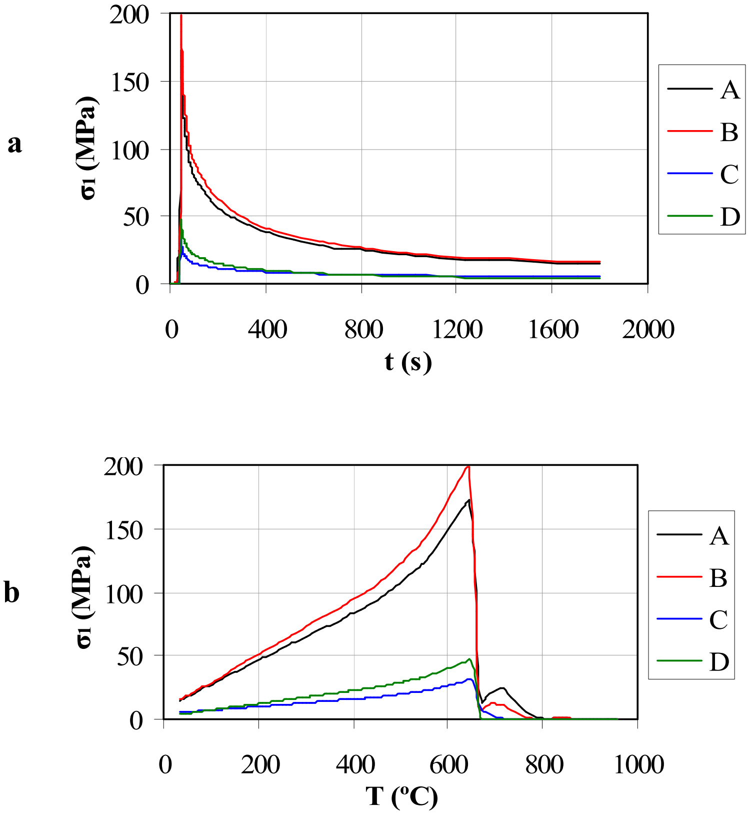


Figure 6.23 Thermal stresses in the porcelain of Au-Pt/VMK95 crown for α2 depending on; **(a)** time, **(b)** temperature.

The thermal stresses in Au-Pt substructure depending on time and temperature for α2 are given in Figure 6.24a and b, respectively. As seen in this figures, the maximum stress occurs at node F and its value is 233 MPa, approximately. In addition, the value of the maximum thermal stress at node A interacted to node B of the porcelain is 125 MPa.

In the comparison of Figure 6.21 and 6.24, it can be seen that the maximum thermal stress at node F increases and its value at node A decreases, by using α2.

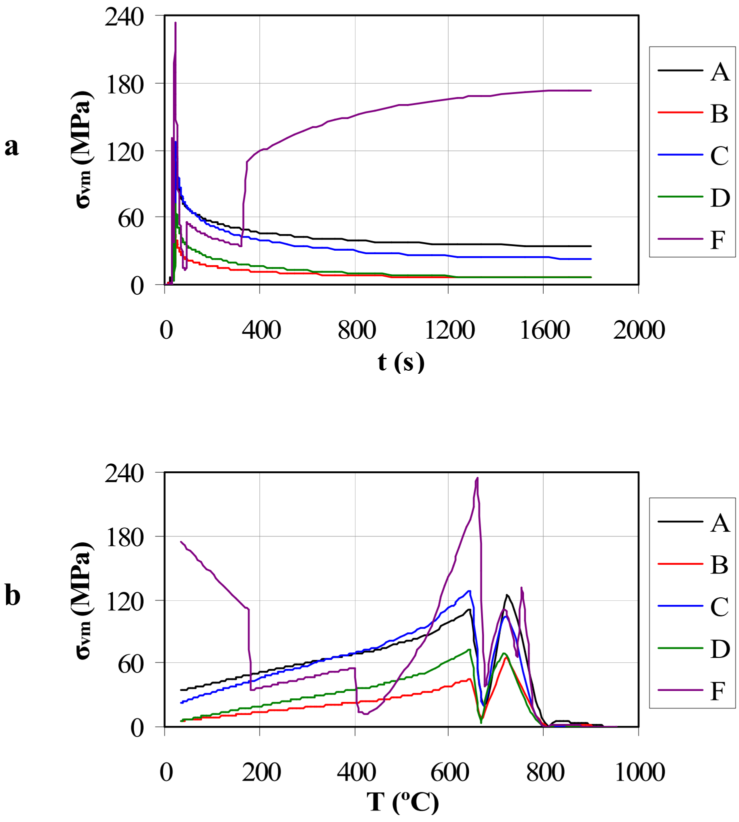
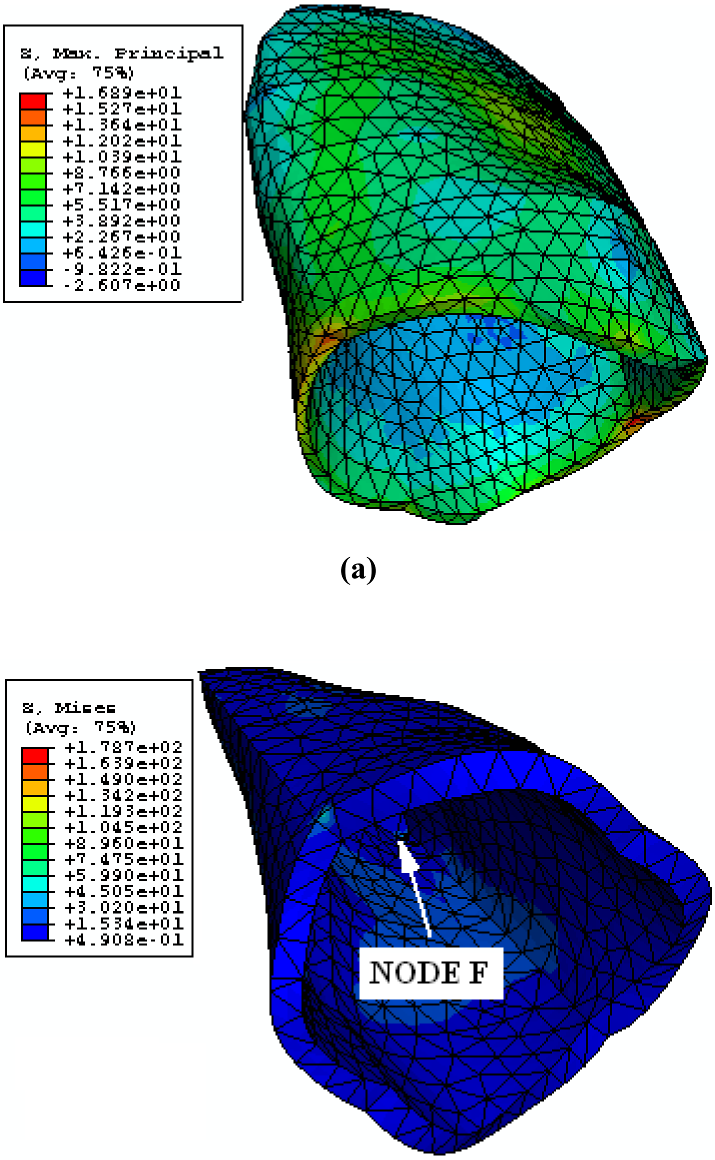


Figure 6.24 Thermal stresses in Au-Pt of Au-Pt/VMK95 crown for α2 depending on; **(a)** time, **(b)** temperature.

Figure 6.25a and b shows the residual stresses remained on the porcelain and Au-Pt substructure for α2, respectively. As seen in these figures, the residual stresses intensify at node B of porcelain and at node F of Au-Pt. The values of them for porcelain and Au-Pt are 17 MPa and 179 MPa, respectively. In addition to this, the residual stress value at node A of Au-Pt substructure has been found as 34 MPa.



(a)

(b)

Figure 6.25 Residual stresses for α2, remained on; **(a)** porcelain, **(b)** Au-Pt.

6.1.5 Equivalent plastic strains in the substructure materials

At the end of the cooling process, the equivalent plastic strains which occur in the metal alloy substructure materials are given in Figure 6.26a, b, c, d, e, f, g and h.

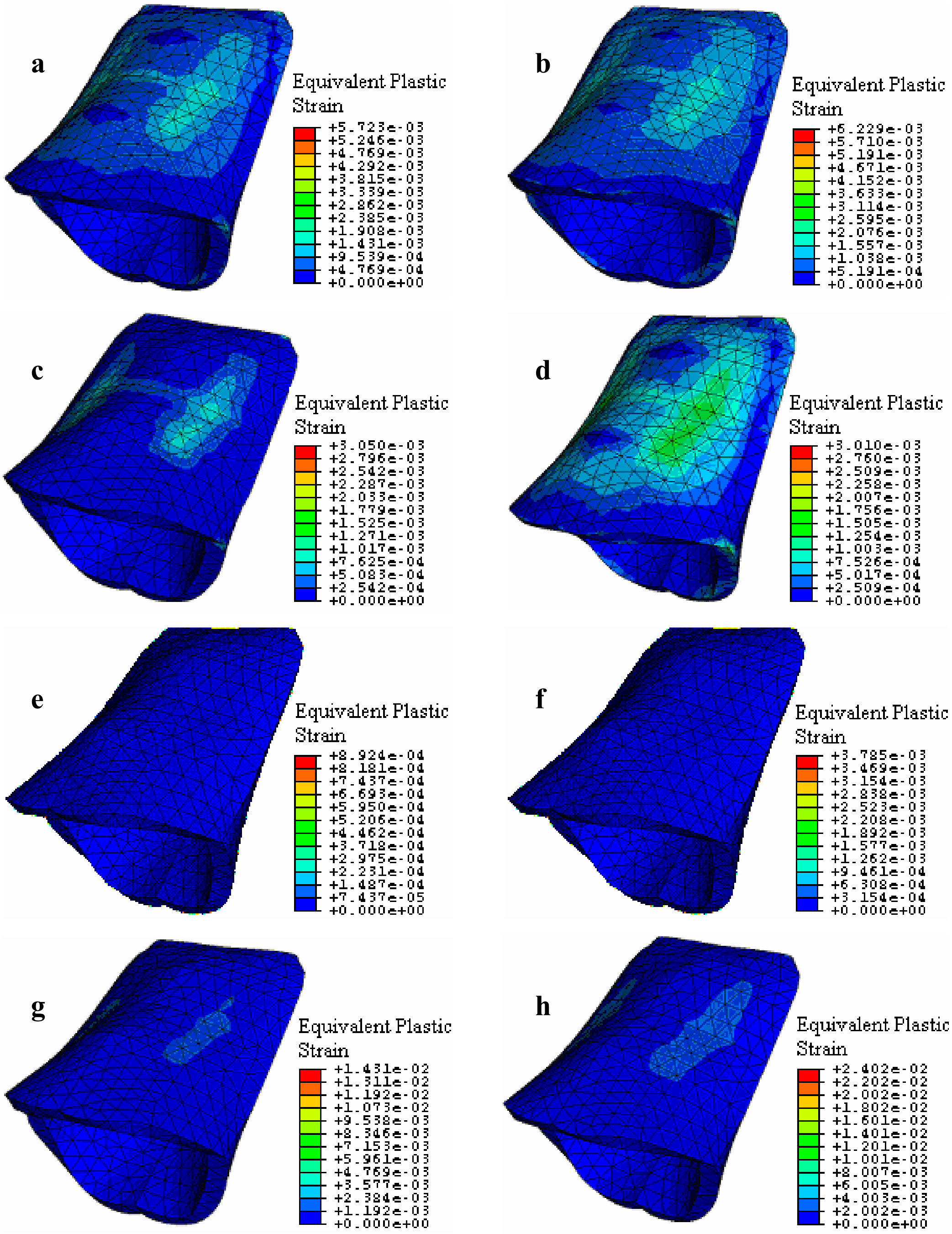


Figure 6.26 Equivalent plastic strains in; **(a)** Ni-Cr for α2; **(b)** Ni-Cr for α1; **(c)** Co-Cr for α2; **(d)** Co-Cr for α1; **(e)** Pd-Ag-Au for α2; **(f)** Pd-Ag-Au for α1; **(g)** Au-Pt for α2; **(h)** Au-Pt for α1

In this figures, it has been determined that the equivalent plastic strains happen in the local regions of the crown and also, are among the elastic limits. Therefore, using the elastic properties can be adequate in the analyses. However, higher cooling rates and different material pairs can cause increasing of the stresses and plastic deformations. In these situations, the behaviors of the metal substructure in the plastic region must be taken into consideration in the analyses.

6.2 Comparison of Maximum and Residual Stresses

The maximum and residual stress values of the porcelain and the substructure materials and the critical nodes on them are given in Table 6.1 and 6.2 for the thermal expansion coefficients α1 and α2, respectively.

From these tables, generally, it is seen that the stresses intensify at nodes B and E for VMK 95 porcelain and nodes A, F for the substructure materials. However, it is thought that the stresses concentration at node E for porcelain and node F for the substructure materials can be based on the problems from the model. In addition to this situation, it is explain that the fractures occur in the cervical margin, mesial and distal corners of the crown in the clinical studies. Therefore, in the comparison of the stress values, node B for porcelain and node A for the substructure materials have been considered as the critical nodes.

Figure 6.27a and b depicts the comparison of the maximum stress values of the porcelain and the substructure materials for the thermal expansion coefficients α1 and α2, respectively.

The comparison of the residual stress values of the porcelain and the substructure materials are shown in Figure 6.28a and b for the thermal expansion coefficients α1 and α2, respectively.

Table 6.1 Maximum stresses, residual stresses and critical nodes in the substructure materials and porcelain according to α1.

Type of Crown	**Max. Stress (MPa)**	**Residual Stress (MPa)**	**Critical Node**
VMK 95	110 B	76 B	B
Ni-Cr	175 A	174 A	A
VMK 95	158 E 56 B	64 E 16 B	E
Co-Cr	329 F 221 A	329 F 132 A	F
VMK 95	135 E 27 B	94 E 12 B	E
Pd-Ag-Au	135 F 108 A	104 F 77 A	F
VMK 95	138 E 62 B	91 E 22 B	E
Au-Pt	155 F 134 A	82 F 55 A	F

Table 6.2 Maximum stresses, residual stresses and critical nodes in the substructure materials and porcelain according to α2.

Type of Crown	**Max. Stress (MPa)**	**Residual Stress (MPa)**	**Critical Node**
VMK 95	210 B	160 B	B
Ni-Cr	230 A	228 A	A
VMK 95	196 B	87 B	B
Co-Cr	249 A	192 A	A
VMK 95	155 B	53 B	B
Pd-Ag-Au	260 F 49 A	176 F 21 A	F
VMK 95	199 B	17 B	B
Au-Pt	233 F 125 A	179 F 34 A	F

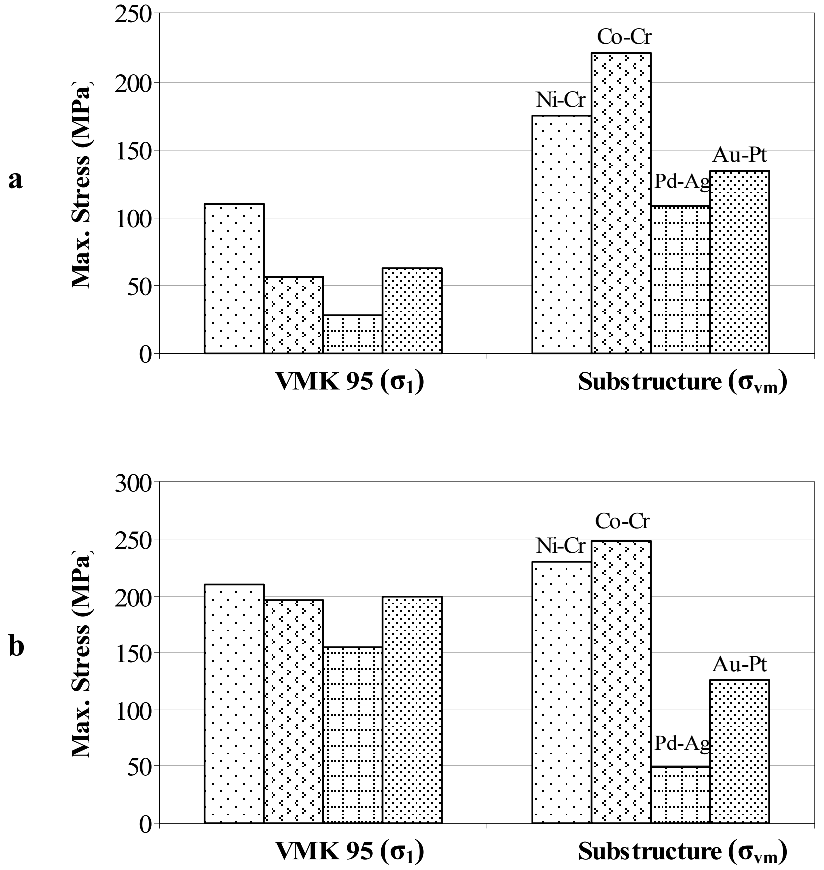


Figure 6.27 Comparison of the maximum stresses in the substructure materials and porcelain according to; **(a)** α1, **(b)** α2.

As seen in Figure 6.27a and b, Pd-Ag-Au/VMK95 material pair depicts more compatible behaviour in terms of the maximum stress values according to the other material pairs for α1 and α2. The minimum stress values for porcelain and the substructure materials happen in the Pd-Ag-Au/VMK95 crown. For the porcelain, the most critical material pair is Ni-Cr/VMK95 for α1 and α2. Furthermore, for the substructure materials, Co-Cr/VMK95 material pair is the most critical crown type. Because, Co-Cr alloy has the highest maximum stress values for α1 and α2. Notwithstanding, the maximum stress values occur in the substructure materials does not exceed the yield stress values given by the manufacturer firms.

Figure 6.27b shows the comparison of the maximum stress values of the porcelain and the substructure materials for the thermal expansion coefficient α2. When the thermal expansion coefficient which has the increased values at the high temperature (α2) is used in the analyses, it is seen that the maximum thermal stresses increase for porcelain in all of the material pairs. However, in the substructure materials, while the maximum stresses of Ni-Cr and Co-Cr increase, it is seen that the maximum stress values decrease for Pd-Ag-Au alloy or does not change for Au-Pt alloy.

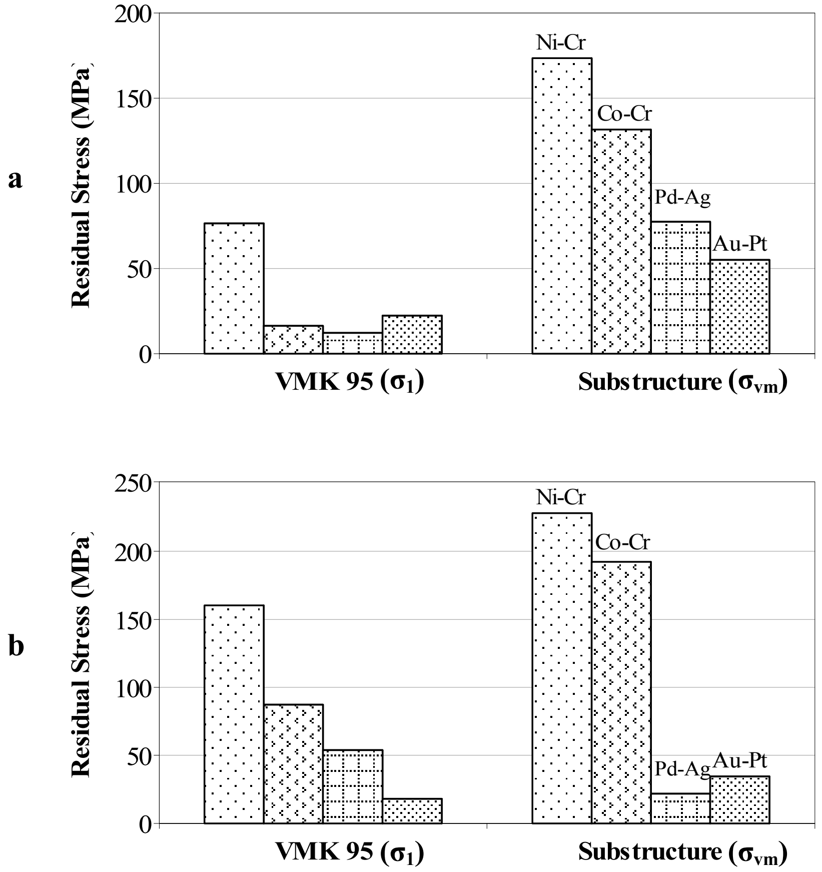


Figure 6.28 Comparison of the residual stresses in the substructure materials and porcelain according to; **(a)** α1, **(b)** α2.

As seen in Figure 6.28a and b, in terms of the residual stresses remained on the porcelain, the most compatible material pair is Pd-Ag-Au/VMK95 and Au-Pt/VMK95 material pairs for α1 and α2, respectively. Also, the most dangerous crown type is Ni-Cr/VMK95 for α1 and α2. Nevertheless, the arrangement of the residual stress values remained on the substructure materials from big to small is Ni-Cr, Co-Cr, Pd-Ag-Au, Au-Pt for α1 and Ni-Cr, Co-Cr, Au-Pt, Pd-Ag-Au for α2.

Figure 6.28b points out the comparison of the residual stress values of the porcelain and the substructure materials for the thermal expansion coefficient which has the increased values at the high temperature (α2). When α2 is used in the analyses, it is seen that the residual stresses of the porcelain increase in Ni-Cr/VMK95, Co-Cr/VMK95 and Pd-Ag-Au/VMK95 crowns. Also, for Au-Pt/VMK95 crown type, the residual stress value remained on the porcelain decreases. However, in the substructure materials, while the residual stresses of Ni-Cr and Co-Cr alloys increase, it is seen that the residual stress values decrease for Pd-Ag-Au and Au-Pt alloys.

CHAPTER SEVEN
CONCLUSIONS AND RECOMMENDATIONS

In this study, the effects of the thermal behaviour of dental porcelain at high temperatures has been reflected onto the elastic properties and it has been demonstrated that the transient and residual stresses, appearing after the firing process, could be calculated accurately by the elastic or elastic-plastic solution method. The results of this study are similar to past studies, carried out with other methods. These can be taken as evidence of validity and reliability of the present method. The results obtained from this dissertation have been given in below article by article.

- Generally, the change of the stress graphics depending on the time and temperature are in the same form approximately at all nodes of the finite element model in all of the material pairs. Especially, the stresses intensify at the interfaces, increase in some of the local points of the model and are under the strength limits on the great part of the crown.

- In the viscoplastic region, the stresses are in negligible levels and near the viscoelastic region the values of them increase. After a short time from the beginning of the cooling process, that is to say T_g, in the region where the porcelain shows the linear elastic behaviour, the stresses happen on the porcelain reach the maximum values. At the end of the cooling process, the stresses remained on the crown are called as the residual stresses.

- In such studies, not only the residual stresses but also the maximum stresses occur during the cooling process are taken into account in the evaluations. Forasmuch as, in spite of residual stresses come into question in the clinical studies, maximum thermal stresses are the most important reason of the micro cracks probably occur in veneering material during the cooling process. This situation reveals the importance of the glazing process applied on the veneering material at the last firing process. The glazing process applied provides closing the cracks.

- The results obtained from the analyses depict a similar distribution to Dehoff's study (Dehoff et al., 2006) which calculated the residual stresses in full ceramic restorations by viscoelastic FEA. They have concluded that the residual stresses could be a risk in clinical applications since the residual tensile stresses on the local regions of the veneer materials could be high enough to cause delayed or catastrophic failure when high occlusal forces are superimposed on them. This conclusion accommodates to the conclusions of this dissertation. It has been observed that if Δα ($\alpha_{substructure} - \alpha_{pocelain}$) is high, the risk of microcrack occurrence in the porcelain increases. Also the producer firm have noticed that, if metal alloys have higher thermal expansion coefficients than the range suggestion, it is often necessary to use a slow cooling programme in order to ensure that cooling from 900°C to 700°C is shorter than 3 minutes (Vita Metal Ceramics, VMK 95, nd).

- In most of clinical studies (Papazoglou & Brantley, 1998; Ozcan & Niedermeier, 2002; Ozcan et al., 2006), it was reported that, cracks were seen at the cervical region in veneering material. In this dissertation, too, it has been observed that the residual and transient stresses concentrate at the cervical margin of the crown. Therefore, it has been seen that the clinical studies support this dissertation.

- The maximum stresses occur at the interfaces between the porcelain and the substructure materials, and cervical regions of both porcelain and substructure materials. At the outer surface of the porcelain and the inner surface of the substructure materials, the stresses are lower levels. Furthermore, if it is thought that the opaque layer is at the interface between the porcelain and the metal alloy, it can be said that the stresses on the opaque layer are higher than that on the porcelain. When the equivalent plastic strains on the substructures investigate, it is seen that the equivalent plastic strains happen at the local points where the maximum stresses occur and are in negligible levels.

- The maximum thermal stresses happen on the metal substructure materials are under the strength limits given by the manufacturer firms. This situation depicts that a problem will not come out in terms of safety of the metal alloy. Besides, in terms of the crown geometry, the negligible levels of the equivalent plastic strains show that there is not a matter originated from the metal alloy.

- Inside the metal alloy substructures, it has been seen that Pd-Ag-Au alloy presents the more compatible behaviour in terms of safety of the porcelain veneering material according to the other material pairs. In other words, Pd-Ag-Au/VMK95 material pair is the most suitable material pair. Furthermore, it has been seen that more critical stresses occur in the crown with Ni-Cr and Co-Cr metal alloys. In these alloys, Co-Cr metal alloy is more safety than Ni-Cr alloy in terms of the thermal stresses.

- The most important result obtained from this dissertation is that the minimum differences between thermal expansion coefficients of porcelain and substructure materials affect markedly to the thermal stresses. This situation is emphasized from a lot of studies in the literature. Also, according to thermal expansion coefficient, the porcelain manufacturer firms determine the substructure materials which are harmonious with their porcelains.

- Particularly, for Ni-Cr/VMK95 material pair having α2, it has been seen that the residual stresses joined with the occlusal forces can be caused to the faults of function and the damages. As a matter of fact, Dehoff et al. (2006) observed such damages for full ceramic restoration in the clinical studies. Additionally, the thickness of the porcelain structure has an important effect on the stresses. Because, it has been determined that the maximum residual stresses occur in the thinnest regions of the porcelain.

The following recommendations may be listed to shed light on further researches:

- The stress analyses done from the maximum furnace temperature has been provided that it is reached the more accurate results in both metal alloy substructure materials and porcelain veneering material.

- It has been seen that defining the porcelain as a linear elastic material by including its different mechanical behaviour at all temperatures provides a great facility in terms of finite element analyses.

- To calculate the modulus of elasticity of the porcelain in the viscoelastic region, knowing only T_g and T_s temperatures is enough. In this wise, the changing curve of the modulus of elasticity of the porcelain at the high temperatures can be drawn more sensible with the help of its viscoelastic properties. For the similar studies, this situation is an important advantage of the method applied. Calculating T_g and T_s temperatures depending on the cooling rates enables to use this method for the different cooling rates, too.

- Measuring the cooling curves experimentally and superposing with the curves obtained from the finite element analyses provides reflecting the effects of radiation and convection to the results and determination of the media exactly. Thus, even the great wrongs done in the properties which effect only on the temperature distribution (k, d and C_p) can be eliminated by superposing the experimental and numerical cooling curves.

- It is probable to mention the various measures reduced the residual stresses. However, some disadvantages resulted from this measures can occur. Although the residual stresses can be reduced by decelerating the cooling rate, the cooling slowly can change the crystal structure of the porcelain and consequently, the mechanical behaviour of the porcelain can be affected from this change.

- The most important factor is to use more compatible material pairs with each other. Using the substructure materials such as Pd-Ag-Au and Au-Pt alloys which show more close behaviour to the thermal and mechanical behaviours of the porcelain reduces to the residual stresses on a large scale. But, the most important disadvantage of such alloys are that more expensive than worthless metal alloys such as Ni-Cr and Co-Cr. Furthermore, concentrating the stress intensify at the interfaces point out occurring more critical conditions in the opaque layer located between metal alloy and porcelain. This situation presents that the opaque material having the similar properties with the porcelain need to be compatible with metal substructure, too.

- In addition, it has been determined that rounding the sharp corners on the crown is an important and practical precaution reduced to the stress intensifies.

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