DESIGN AND MECHANICAL ANALYSIS OF A NEW DENTAL IMPLANT THAT WOULD MIMIC NATURAL TOOTH WITH A PERIODONTAL LIGAMENT

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submitted by **ÖMER PEKTAŞ** in partial fulfillment of the requirements for the degree of **Master of Science in Mechanical Engineering Department, Middle East Technical University** by,

Prof. Dr. Canan Özgen ________________
Dean, Graduate School of **Natural and Applied Sciences**

Prof. Dr. Suha Oral ________________
Head of Department, **Mechanical Engineering**

Assist. Prof. Dr. Ergin Tönük ________________
Supervisor, **Mechanical Engineering Dept., METU**

Examining Committee Members:

Prof. Dr. Eres Söylemez ________________
Mechanical Engineering Dept., METU

Assist. Prof. Dr. Ergin Tönük ________________
Mechanical Engineering Dept., METU

Prof. Dr. M. Kemal Özgören ________________
Mechanical Engineering Dept., METU

Prof. Dr. Reşit Soylu ________________
Mechanical Engineering Dept., METU

Prof. Dr. Ahmet Keskin ________________
Oral & Maxillofacial Surgery Dept.,
Faculty of Dentistry, Ankara University

Date: 08 – 02 – 2012

I hereby declare that all information in this document has been obtained and presented in accordance with academic rules and ethical conduct. I also declare that, as required by these rules and conduct, I have fully cited and referenced all material and results that are not original to this work.

Name, Last name : Ömer PEKTAŞ

Signature :

ABSTRACT

DESIGN AND MECHANICAL ANALYSIS OF A NEW DENTAL IMPLANT THAT WOULD MIMIC NATURAL TOOTH WITH A PERIODONTAL LIGAMENT

Pektaş, Ömer
M.Sc., Department of Mechanical Engineering
Supervisor: Assist. Prof. Dr. Ergin Tönük

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Dental implant is an artificial dental root that is used to construct dental restorations, similar to the original teeth, in order to regain the function of missing teeth of patients experiencing tooth loss. At the interface between the jawbone and the roots of natural teeth, a thin, elastic, shock absorbing tissue, called the periodontal ligament (PDL), forms a cushion which provides certain mobility to the natural teeth. The restorations supported by dental implants, however, involve completely rigid structures. This causes overloading of the implant while bearing functional loading together with neighboring natural teeth, which leads to high local stresses within the implant system and in the jawbone.

The aim of this thesis study was to develop a novel dental implant model involving resilient components in the upper structure (abutment) in order to mimic the mechanical behavior of the PDL. Within the scope of the study, a complete mechanical design of a new dental implant model was made. The proposed model was optimized based on the design objectives by using Finite Element Method. The optimal design was verified to overcome the problem of loosening of the abutment screw (a common complication in previous designs), yield very similar axial mobility behavior as that of a natural tooth and withstand biomechanical loads without failure. In addition, as a support of a dental bridge in combination with a natural tooth, the proposed design was demonstrated to provide uniform load sharing with the natural tooth and substantially reduced magnitude of peak stresses within the construction, compared to a rigid counterpart.

Keywords: Dental Implant, Elastic Abutment, Periodontal Ligament, Mechanical Design, Finite Element Analysis

ÖZ

PERİODONTAL LİGAMANI TAKLİT EDEBİLECEK ESNEK YAPI İÇEREN DİŞ İMPLANTI TASARIMI VE MEKANİK ANALİZİ

Pektaş, Ömer
Yüksek Lisans, Makine Mühendisliği Bölümü
Tez Yöneticisi: Yrd. Doç. Dr. Ergin TÖNÜK

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Diş implantı, diş kaybı yaşayan hastaların eksik dişlerinin fonksiyonlarını geri kazanmak amacıyla yapılan doğal dişlere benzer restorasyonlarda kullanılan yapay diş köküdür. Periodontal ligaman adı verilen, şok emici özelliğe sahip, ince, elastik doku doğal diş kökleri ile çene kemiği arayüzünde bir yastık görevi görerek dişe belli bir hareket kabiliyeti kazandırır. Diş implantı destekli restorasyonlar ise tamamen rijit bir yapıya sahiptir. Bu durum, implant destekli dişin komşu doğal dişler ile beraber fonksiyonel yükleri karşılaması esnasında daha fazla yüke maruz kalmasına, implant yapısı ve implant çevresindeki kemik dokusu içerisinde yüksek lokal gerilimlerin oluşmasına neden olur.

Bu tez çalışmadaki amaç, üst kısmında (abutment) içerdiği esnek yapı sayesinde, periodontal ligamanın mekanik davranışını taklit edebilecek yeni bir diş implantı modelinin geliştirilmesidir. Tez kapsamında, esnek yapıya sahip özgün bir diş implantı modelinin mekanik tasarımı eksiksiz olarak tamamlanmıştır. Önerilen implant modeli, tasarım hedefleri doğrultusunda, Sonlu Elemanlar Analizi yöntemi kullanılarak optimize edilmiştir. Optimize edilen diş implantı modelinin (bugüne kadar önerilen tasarımların birçoğunda ortak sorun olan) üstyapı sıkıştırma vidasının gevşemesini engellediği, doğal dişlere yakın bir şekilde eksenel hareket sağladığı ve biyomekanik yüklere dayanabildiği doğrulanmıştır. Bunun yanı sıra önerilen esnek yapılı modelin, doğal dişlerle beraber köprü protezi dayanağı olarak kullanılması durumunda, tamamen rijit yapıya sahip bir diş implantına kıyasla, doğal dişlerle yükü daha homojen bir şekilde paylaştığı, köprü protezi ve implant yapısı içerisinde daha düşük gerilimlerin oluştuğu gösterilmiştir.

Anahtar kelimeler: Diş İmplantı, Esnek Üstyapı, Periodontal Ligaman, Mekanik Tasarım, Sonlu Elemanlar Analizi

To my family…

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INTRODUCTION

1.1 An Overview of Dental Anatomy

Dental anatomy is a field of biology and medical science which deals with the structure and functions of teeth. It studies the appearance, development, classification and parts of teeth [1].

Teeth are small bonelike structures, root end of which are firmly embedded in upper and lower jaws whereas other end is exposed. The fundamental roles of teeth are biting and chewing for grinding food, that is, teeth provide mechanical digestion in mouth as the beginning of digestive system. Besides, teeth primarily affect the function of speaking. Crooked teeth and unbalanced dental occlusion cause inaccurate pronunciation of letters and words. Another major concern is the contribution of teeth to the aesthetical appearance which significantly affects personal relations, thus, social life. People having irregular dental structure or missing teeth, usually avoid oral communication.

All the teeth lie on the jawbone that maintains stability and mobility for the mouth and teeth during mastication. The upper jawbone is called the maxilla and the lower jawbone is called the mandible [2]. The part of the jaw that surrounds the tooth roots and forms alveolus, tooth sockets, is called alveolar bone. Similar to the other bones in the body, alveolar bone is in a continual state of flux which includes recurrent destructive and formative activities in the bone structure throughout the life time so that the bone remains functional and healthy[3].

Teeth are composed of multiple tissues with different biological and mechanical properties. As it can be observed from Figure 1.1, a tooth has two main parts, namely, the crown and the root. The crown is the upper visible portion, outer surface of which is covered by enamel, a non-living extremely hard substance [2]. Underneath the enamel there exists an alive, cream colored, hard material called dentin with the pulp chamber in the center [4]. The dentin surrounds and protects the pulp which contains nerves and blood vessels [1].

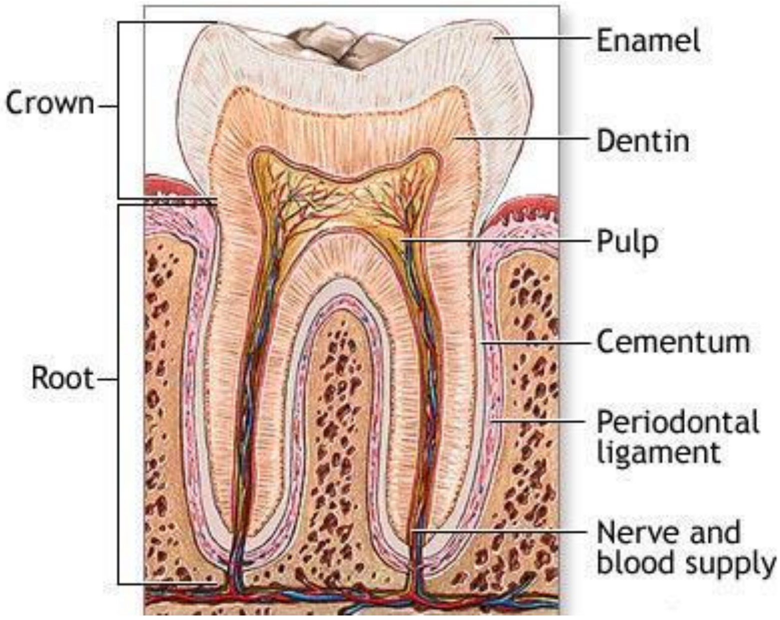


Figure 1.1 Anatomical structure of a tooth [5]

The roots extend into the jaw bone and embedded in tooth sockets. They are covered by a thin calcified layer called cementum which meets enamel at the neck of the tooth. The lower portion of dentin remains inside the cementum. In order to provide circulation and sensation, a canal of blood vessels and nerves passes through each root reaching the area of pulp [2].

Another structural part that has a crucial role in dental anatomy is the periodontal ligament (PDL). It is a collagenous connective tissue, completely covering tooth roots and separating tooth from the alveolar bone [2]. The periodontal ligament is a living membrane having both nerve and blood supply. It forms an elastic cushion at the interface of jawbone and tooth roots so that slight movement of a tooth is enabled. Stating differently, instead of a rigid connection, such an interface provides flexibility. As a result of this, the shocks that would have probably been induced by the forces created during biting or chewing are absorbed by uniform distribution of induced stresses so that teeth and alveolar bone structures are protected from being harmed [1].

1.2 Loss of Teeth and Methods of Treatment

Tooth loss is an undesirable condition that is especially encountered by adult people. It occurs when one or more teeth, due to several reasons, loosen and fall out. The case of complete loss of natural dentition is termed as edentulism, whereas loss of some teeth is called partial edentulism [6].

1.2.1 Causes of Tooth Loss

Most of the cases of tooth loss are commonly attributed to the reasons that are traced as below.

1. Physical trauma or injury

An external effect such as a blow or injury may lead to exertion of an amount of force exceeding the limits that dental tissues can physically endure and this may cause teeth to loose and fall out or break.

2. Tooth decay

Clinically termed as dental caries or dental cavities, decaying of tooth is incurred by a biochemical process. The bacteria growing in the remains of food accumulated in the cavities around the base region of teeth form organic acids. These acids gradually destroy the dental enamel and other underlying tissues. This situation is often characterized by intensive pain and bad breadth. Over a period of time the decaying teeth loosen and fall or require extraction [7].

Some of the factors that trigger tooth decay are not brushing teeth resulting in poor oral hygiene, lack of dental care, excess intake of starch and sugars, malnourishment etc.

3. Periodontal disease

When periodontal tissues are damaged due to excessive compressive stresses or infection develops around these tissues, this is referred as periodontal disease. It is responsible for loss of teeth more than any other cause.

If occlusal forces creating intensive pressure and tension fields exceeding the limits of periodontal ligament act on tooth surfaces, periodontal tissues usually suffer. Excessively high pressure causes the collagen fibers of the periodontal ligament to be compressed and stretched resulting in undue tension in areas of high pressure. If the fibers are extended too much, they begin to separate from the dental junctions, enlarging the periodontal pocket, which leads to penetration of bacteria and harmful chemicals, posing the risk of infection. On the other hand, the disconnected fibers can no longer stimulate bone regeneration and bone supporting the tooth begins to resorb. This leads to loosening and eventually falling out of the tooth [3].

In the Journal of Periodontology, dental researchers list the risk factors for tooth loss due to periodontal disease as follows:

- Being older than the age of 35
- Being male
- Never getting professional dental care
- Never using a toothbrush
- Smoking
- Having diabetes
- Having high blood pressure
- Having rheumatoid arthritis

The last factor leads to the fact that the front teeth are more prone to be lost due to periodontal diseases compared with the teeth at the back of the mouth [8].

1.2.2 Consequences of Tooth Loss

It is essential to replace or treat a missing tooth; otherwise, neglecting the treatment for a long time may lead to serious and sometimes irrecoverable drawbacks. Millions of people lose their capacity of chewing and face with aesthetical problems in addition to the effects on anatomical structures like periodontal tissues, alveolar bone and neighboring teeth [9].

1. Effects on the jaw bone

Throughout the course of life, there is a continual process of regeneration of alveolar bone similar to all other bone tissues in the body. Osteoblasts are the cells of bone that make new bone, whereas osteoclasts are the cells responsible from absorbing bone [10]. The activities of these cells balance each other. The osteoclasts continuously produce enzymes that destroy and dissolve organic components of the bone matrix. On the other hand, osteoblasts produce new bone on the side of bone that is in contact with the periodontal ligament. However, the formation of new cells requires stimulation that is created by the forces exerted on teeth and transmitted to the jaw bone through tooth roots and periodontal connection.

When a tooth is lost, the major stimuli that triggers bone formation around surrounding bone becomes absent, since no stress field is induced on that region. As a result, the formative activities of osteoblasts that counterbalances the destructive activities of osteoclasts stop and the bone in the region of lost tooth resorbs, loses its density and gets weakened. The width and height of the bone decreases and the form of the bone is lost. As for the worst scenario, the loss of all teeth leads to complete bone loss and deformation of the shape [11].

2. Effects on other teeth

Losing a tooth adversely affects both its functional opponent and the neighboring teeth. When a tooth loss occurs, not only the alveolar bone in the socket of lost tooth resorbs, but the bone surrounding roots of the tooth which is the functional opponent in the other jaw also deforms. This is because no longer the functional occlusion is maintained and the forces encouraging the continual structural changes are not applied on the previously opponent tooth. Hence, the opponent tooth becomes a non-functioning tooth whose periodontal ligament gradually becomes thinner and loses its ligament-like properties and functions. Besides, the tooth in the opposing jaw tends to move or extend towards the missing tooth location in time as illustrated in Figure 1.2.

The neighboring teeth are also influenced by a tooth loss. In case of a tooth loss, the harmonious dental occlusion is lost and occlusal imbalance develops. The neighboring teeth of the lost tooth begin to shift and tilt towards the empty location (Figure 1.2). As a result, the forces created during biting or chewing are no longer transmitted along the long axes of these teeth. This may cause periodontal disease which may possibly result in further tooth loss [3].

There are some other consequences of tooth loss related with the aforementioned effects on anatomical structures. As the jaw bone resorbs and loses its shape, problems with speech and aesthetical appearance arise, which unfavorably affect social life. Moreover, as the number of missing teeth increases, chewing certain food becomes difficult leading to nutritional problems. On the other hand, the remaining teeth are subjected to more stresses during chewing, which may cause weakening of these teeth.

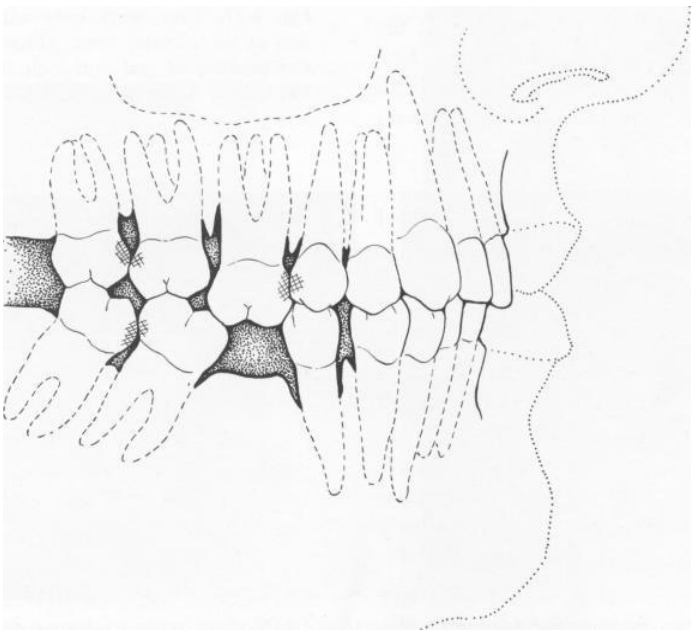


Figure 1.2 Tilting of teeth due to a missing tooth [12]

1.2.3 Methods of Treatment of Tooth Loss

In order to avoid the drawbacks of tooth loss explained in previous section, medical treatment should be applied. For the treatment of tooth loss, a number of different techniques are implemented as described below.

1.2.3.1 Dental Bridges

Dental bridge is a conventional method of replacing missing teeth. It is made by placing crowns on the neighboring teeth at each side of empty location and joining these crowns by a false tooth, called pontic, in the middle. In order to fit the crowns, natural and healthy teeth at both sides are cut to a proper geometry (Figure 1.3). Then the dental bridge is placed, adjusted and cemented to the cut teeth by special adhesives. Dental bridges are also known as fixed partial dentures [13].

Figure 1.3 The procedure for dental bridge construction [14]

The adjacent teeth which are cut for a dental bridge application lose their resistance against external effects and become more prone to be adversely affected. In the cases of inconvenient bridge restorations, the teeth under bridge may decay in time requiring replacement of the bridge. It is also necessary to replace the bridge when adaptation problems arise between gum and the bridge. The removal of the bridge causes trauma on the supporting teeth. Another drawback related with dental bridges is that no force is transmitted through the false tooth at the middle to the bone since it has no root. Hence, the bone may begin to resorb.

1.2.3.2 Removable Partial and Complete Dentures

Dentures consist of false replacement teeth that are attached to gum-colored plastic bases made of acrylic resin and connection is maintained by a metal framework. Removable partial dentures can be used when there are remaining natural teeth. As it can be seen in Figure 1.4a, they include metal clasps or other attachments that are used to attach dentures to existing teeth and hold the denture in place. Partial dentures form bridges filling the gap due to missing teeth.

The complete dentures, on the other hand, are used as a treatment for the cases of complete tooth loss. They cover the entire upper and/or lower jaw (Figure 1.4b). Removable complete dentures are placed directly on the gum tissue without any fixation and can be removed easily [13].

The disadvantage of removable dentures is that during chewing, speaking or laughing, the denture may loosen, move or slide from its place resulting in discomfort for the patient.

(a)

(b)

Figure 1.4 Removable dentures: a) partial denture [15], b) complete denture [16]

1.2.3.3 Dental Implants and Overdentures

Dental implant is an artificial dental root that is used to regain the function of one or more missing teeth of edentulous patients (Figure 1.5). It is the safest, most functional and efficient way among the techniques for missing tooth replacement applied by dentists. A dental implant is basically composed of two parts. The lower part is surgically inserted into the jaw bone and fixed in the location of priorly existing natural tooth roots. It is directly in contact with bone. During a period of healing, bone grows around the implant and provides a strong structural support similar to the roots of natural teeth.

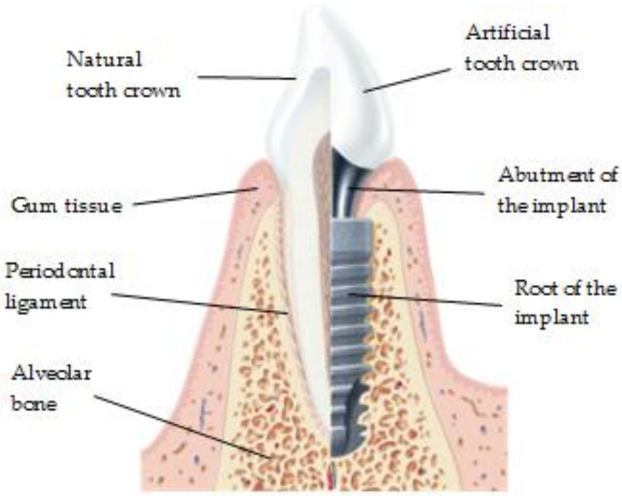


Figure 1.5 Dental implant as an artificial tooth root [17]

The upper part, called abutment, is mounted on the portion of the lower part that is exposed to the oral cavity beyond the level of gum line. The false tooth, called the crown that replaces the missing tooth is fitted onto the abutment (Figure 1.6).

Dental implants have significant advantages over other treatments of tooth loss. Compared with dental bridge applications, there is no need to sacrifice the healthy adjacent teeth since the neighboring teeth are by no means affected during a dental implant treatment. The dental implant is directly screwed into the cavity of the missing tooth. Another prominent benefit of dental implants is that the shrinkage of the alveolar bone is prevented by providing an interface for the forces on the crown to be transmitted to the bone and to trigger new cell formation as in the natural tooth case. Therefore, the necessary stimulation for bone regeneration not existing in dental bridge applications is maintained resulting in preserved bone shape.

(1) **(2)** **(3)**

(a) **(b)**

Figure 1.6 A typical dental implant system: (a.1) root, (a.2) abutment, (a.3) crown [18] and (b) single tooth replacement [19]

The crowns supported with dental implants look like original teeth and can restore the chewing efficiency comparable to that of natural teeth. With dental implants, discomfort caused by slipping or cracking and pain due to misplacement of removable dentures are avoided. This can be achieved by using overdentures.

An overdenture is a fixed type of denture which is supported by remaining teeth or dental implants [20]. An implant-supported overdenture is a commonly used technique which has several advantages over an ordinary removable denture. In order to hold the denture in place, usually ball type abutments are utilized (Figure 1.7). The overdenture is clipped on the abutment and can be removed easily for cleaning or other purposes.

Overdentures are mechanically stable and provide proper transmission of forces created during chewing. When an implant supported overdenture is used, less bone resorption occurs. On the other hand, the adaptation of the patient to the overdenture will be quicker due to its stability compared to a removable denture.

(1) **(2)** **(3)**

Figure 1.7 Implant-supported overdentures: (1) lower jaw overdenture, (2) a dental implant with ball type abutment, (3) upper jaw overdenture [19]

Another prominent advantage of including implant-support together with natural tooth support for an overdenture is that teeth that have been loosened due to weakening of periodontal tissues can be saved. This is maintained by additional stability gained by the artificial supports, reducing the possibility of torsion on supporting natural tooth and preventing further loosening of these teeth and traumatization of neighboring teeth [3].

1.3 Dental Implantology

Dental implantology is a rapidly advancing field of science that deals with the problem of tooth loss and focuses on providing esthetically pleasing, functionally effective and biologically compatible replacement choices for missing teeth. In the Glossary of Prosthodontic Terms, a dental implant is defined as "*a prosthetic device made of alloplastic material(s) implanted into the oral tissues beneath the mucosal or/and periosteal layer, and on/or within the bone to provide retention and support for a fixed or removable dental prosthesis; a substance that is placed into or/and upon the jaw bone to support a fixed or removable dental prosthesis*" [20]. Figure 1.8 illustrates the use of dental implants for different cases of tooth loss.

Dental implant is an alternative method of treatment of single or multiple tooth loss and it is, in terms of side effects, oral comfort and long term results, superior to the conventional techniques like dental bridges, partial or complete removable dentures. The major concern in dental implant treatments is approximating natural dentition with respect to appearance and functions as far as possible.

(a) **(b)**

Figure 1.8 Dental implants for (a) single and (b) multiple tooth replacement [21]

The need and use of dental implants is increasingly continuing, which is attributable to several factors, including

- tooth loss related to age,
- anatomic consequences of edentulism,
- poor performance of removable prostheses,
- psychological aspects of tooth loss,
- predictable long-term results of implant-supported prostheses and
- advantages of implant-supported prostheses[22].

1.3.1 A Brief History of Dental Implantology

The attempts to overcome the problem of tooth loss and to find out a way of replacing missing teeth date back to as old times as the history of human beings. From ancient time to the modern age, several materials, many geometric forms, surgical and prosthetic methods have been tried until today's state of the art dental implants are developed.

With regard to the ancient implantology, scarcely existing archeological reports demonstrate the attempts of different prosthetic devices used as natural and functional replacements. The findings show that ancient Egyptians, Greeks, Romans, Chinese, Indians and Arabs have used transplantation procedures and devices. In addition to naive artificial units such as shaped stone, wood, cast iron and carved sea shells, bone and natural teeth taken from various animals and even teeth sold by the poor or slaves have been tried as implantation material [23]. In 1931, archeologists found a woman mandible of Mayan origin having tooth-shaped pieces of shell placed into the sockets of missing teeth. Later, in 1970, a Brazilian dental academic, Professor Amadeo Bobbio, investigated that mandibular specimen and in the radiographs he took, he observed bone formation around the implant-like structures [24].

In 18th century, teeth donated by humans are replaced for missing teeth of wealthier people in exchange for some fee; however, immune system reactions precluded the adaptation of foreign materials. In 1785 and 1786 Le Mayeur implanted one hundred and seventy donated teeth, but none of them had successful results [23].

In 19th century, predicating on a false assumption that precious materials would be well tolerated by biological tissues, gold, silver, platinum and some other metal alloys were used as implant materials, which resulted in extremely poor long-term results [3]. These implants have been placed in alveolar sockets immediately after extraction to create suitable replacements. The rejection of these type metallic inclusions was basically due to the fact that they were not inert. The experimental study carried out by Venable et al. (1937) showed that the metals that are not inert tend to ionize on contact with body fluids resulting in production of metallic salts causing excessive proliferation of some tissues whereas inhibiting bone formation [25]. The prevention of bone formation around the implant leads to failure of the implantation.

The modern breakthroughs in dental implantology emerged as a result of so-called serendipity of a Swedish orthopedic surgeon, Professor Bränemark, in 1952. While studying bone healing and regeneration around ‘the rabbit ear chamber’, which was a chamber of titanium designed and developed as a part of research conducted in Cambridge University, he observed in microscopic level that bone had grown around titanium surface in so close proximity that he was unable remove the chamber form the rabbit femur. Dr. Bränemark investigated this phenomenon through further studies on animal and volunteer human tissues, which all contributed to unveiling the biocompatible properties of titanium. Having initially been considered to be appropriate for applications in the field of orthopedics such as knee or hip surgery, later titanium is realized to be utilized as anchorage for dental prosthesis and artificial crowns [26]. In 1965, the first titanium implant was purposely inserted into the jawbone of a human volunteer, Gösta Larsson, with the aim of providing an artificial root for prosthetic teeth [24]. Also, in 1967, Leonard Linkow presented his blade-form titanium implants providing mechanical stability and function for partial and complete dentures [26].

During the period between 1970 and 1980, researchers carried out many experimental studies to obtain better designs and geometric forms for titanium dental implants some of which are the IMZ Implants, TPS Implants, ITI Hollow-Cylinder Implants [27]. Throughout this period, Dr. Bränemark continued his research and in 1971 he introduced titanium hollow screw implants which resulted in increased success rate, clinical applicability and reduced rate of complications compared to blade-form implants. In 1977, Dr. Bränemark published a paper which covered all the data obtained in his studies, and this report provided the scientific data for the development of currently implemented implantation procedures and implant systems [26]. In 1978, he established a commercial partnership with a Swedish defense company, Bofors AB. In 1981, based on the partnership, Nobel Biocare, one of the largest current dental implant producers in the world, was founded with the aim of focusing directly on dental implantology.

In 1982, the Toronto Conference on Osseointegration in Clinical Dentistry set the first guidelines for successful implant dentistry [28]. The successful integration of hollow screw geometry into bone and high biocompatible characteristics of titanium resulted in that screw form dental implants have become the preferred method of tooth replacement and a standard dental treatment technique. Providing a high rate of success and a wide range of restorative options, today, dental implants, under various brand names, are extensively used worldwide. Current studies are mainly focused on improving aesthetics, reducing healing period and simplifying the use of dental implants.

1.3.2 Classification of Dental Implants

The factors determining the success of a dental implant are biocompatibility with living tissues, easiness of operative procedure, time required for healing, being able to endure biomechanical forces and transmit them to jawbone properly. In order to optimize these parameters, several implant materials are utilized; numerous designs for geometric forms and procedures for enhancing surface properties are developed.

The dental implants that have been involved in the market can be classified with respect to various parameters some of which are materials, geometric form, surface properties, method of implantation, implantation time and prosthetic restoration type. Here the first three parameters will be explained.

1.3.2.1 Implant Materials

Dental implant materials are selected with regard to required mechanical properties as well as biocompatibility. Recently, esthetical quality has also become a leading concern.

Dental implants are subjected to high levels of mechanical loads during chewing and biting. It is reported that the bite force is in the range of approximately 100 to 800 N for natural dentitions and the normal chewing force is in the range of 75 to 135 N depending on the location of the tooth on the jaw [29]. The material should have mechanical properties such as tensile strength, and Young's modulus sufficient to endure these loads.

Another important consideration in the material choice for dental implants is biocompatibility. Since a dental implant is in direct contact with living bone tissue, it should be well tolerated without arising complications that may occur due to biological or chemical reactions with toxic effects in an environment involving body fluids.

The materials for the commercially available dental implants that have been used in clinical dental treatments can be traced as metals, ceramics and polymers.

a. Metals

Various types of metals such as gold, platinum, stainless steel, cobalt-chromium alloy have been tried to serve as implant material; however, in the long run most of these metals caused negative tissue reactions and non-biocompatible properties of them were observed. The best choice to fulfill the required properties has been found to be titanium, which is used for construction of the majority of the implants available on the market.

Titanium is a nontoxic, reactive material. When it is in contact with any other electrode, an oxide layer is formed on the surface and this layer resists chemical attacks. It is also inert in living tissues so that it is not rejected by these tissues. Bone regeneration and bonding on the surface of a titanium dental implant occur perfectly [27].

The tensile strength of titanium is high, whereas elastic modulus is low (approximately 100-110 GPa) compared to other metallic biomaterials like stainless steel and Co-type alloys having modulus of elasticity about 206 and 240 GPa, respectively. Therefore, the elastic modulus of titanium is closer to the value for bone that is reported to range between 17 and 28 GPa [30]. In order to further enhance mechanical and also chemical characteristics, several alloys of titanium are developed. Iron, nitrogen, vanadium, carbon, hydrogen and many other materials are incorporated into titanium alloys. The most widely used material for dental implants is annealed Ti-6Al-4V, which is a titanium alloy including aluminum and vanadium. Ti-6Al-7Nb, an alloy developed for surgical implants is also attractive for dental applications [30], [31].

The mechanical properties of pure titanium and titanium alloys are presented in Table 1 [32]. The ultimate tensile strength and yield strength of alloys are significantly enhanced without increasing elastic modulus too much.

Table 1.1 Mechanical properties of titanium and its alloys **[32]**

Material	***Young's modulus [GPa]***	***Yield strength [Mpa]***	***Ultimate strength [Mpa]***	***Ultimate strain [%]***
Ti pure - Grade 1	102.7	170	240	24
Ti pure - Grade 2	102.7	275	345	20
Ti pure - Grade 3	103.4	380	450	18
Ti pure - Grade 4	104.1	485	550	15
Ti-6Al-4V (Annealed)	110 - 114	825-869	895-930	6-10
Ti-6Al-7Nb	114	880-950	900-1050	8-15

b. Ceramics

A number of ceramic materials such as alumina, hydroxyapatite [$Ca_{10}(PO_4)_6(OH)_2$] (HA), tricalcium phosphate ($Ca_3(PO_4)_2$) and bio glasses have been used as implant materials.

Among these, hydroxyapatite and tricalcium phosphate have been disapproved to be implant materials due to their brittle characteristics [33]. On the other hand, it is reported that ceramic implants have low flexural strength values insufficient to withstand occlusal load. Moreover, they have a high degree of solubility, causing hydrolysis when subjected to water and becoming prone to fracture [34]. Another disadvantage of ceramic implants is that they have one-piece structure which precludes the use of angled abutments.

Zirconium dental implants are among the newer ceramic implants used recently. They provide promoted aesthetic quality, biocompatibility and mechanical structure. The most distinctive property of zirconium implants is that they are indistinguishable from natural teeth regarding their appearance since they are purely white as can be seen in Figure 1.9. However, their inferior mechanical characteristics when compared with titanium alloys, lead manufacturers to use ceramic materials as coatings on metallic implants instead of using as implant material.

Figure 1.9 A metal free ceramic dental implant [35]

The most common ceramic coating applications are hydroxyapatite, titanium dioxide and zirconium dioxide coatings implemented using several techniques.

c. Polymers

With the purpose of obtaining a dental implant that would be able to simulate small elastic movements of the periodontal ligament of a natural tooth, some polymers including polyurethane, polyoxymethylene, and polyamide have been used as dental implant materials. However, the insufficient mechanical properties and biological compatibility have constrained their application [34]. Therefore, the studies about the use of polymers have been shifted towards investigation of elastic components to be incorporated into implant systems.

1.3.2.2 Geometric Forms of Dental Implants

Many alternative designs for dental implants have been developed in order to maintain a stable fixation between implant and bone tissues. Among these alternatives two workable types are commonly used for the treatment of missing teeth, namely, root-form and subperiosteal implants [3]. The selection of appropriate type depends on the amount and location of available bone on the jaw.

a. Root-Form Implant

The most prevalent type of dental implants used in clinical applications is the root-form dental implant which is also known as endosseous or endosteal type. These types of implants are placed directly into jawbone as in the case of a natural tooth root. They have a cylindrical geometry having a screw-like form on the surface (Figure 1.10). Root-form implants can be used for anchorage of single or more artificial teeth as well as for the fixation of partial or complete dentures [36].

Figure 1.10 A root-form dental implant fixed on the bone [37]

b. Sub-Periosteal Implant

Contrary to root-form implant, the sub-periosteal implant is composed of a metal framework that is set over the bone to serve for the role of multiple tooth roots without including any part fixed inside the bone (Figure 1.11). Rather, the upper part of the framework protruding through the gum supports the artificial dentures. The use of this kind of an implant is limited to the cases when there exists very little alveolar bone resulting from bone resorption. On the other hand, it can be used only if all the teeth on upper or lower jaw are missing. The dentures for the patients having minimal bone height can be held on their jawbone by using such an implant system [3].

Figure 1.11 A Sub-periosteal type dental implant placed on the jawbone [38]

1.3.2.3 The Surface Properties of Implants and Osseointegration

Being mostly used as a dental implant material, titanium is readily accepted by human body without causing allergic reactions or toxicity. However, in order to further enhance the biocompatibility of dental implants to increase the long-term survival rate, several surface modification techniques are implemented. The basic aim is to increase the rate of osseointegration, which is defined as ‘the process and resultant apparent direct connection of an exogenous materials' surface and the host bone tissues, without intervening fibrous connective tissue present' [20].

The most common techniques for promoting biocompatibility are surface roughening and surface coating. The surface roughness at macro scale is achieved by grit sandblasting (with TiO_2, Al_2O_3 or calcium phosphate grits) and at micro scale by acid etching. As the roughness of the surface increases, the contact area between implant and bone enlarges providing an improved surface condition for better osseointegration. For surface coating, biocompatible materials like TiO_2 and hydroxyapatite are employed [34]. It has been reported that hydroxyapatite coating onto titanium dental implants maintains early adaption with bone; therefore, it is a widely adopted method [22]. Plasma spraying, sputtering and sol-gel coating are the basic methods for coating implant surfaces. Moreover, electrochemical methods such as thermal oxidation, anodic oxidation and micro arc oxidation are also utilized for surface modification of dental implants [34].

1.4 The Role of the PDL and Concept of Elasticity for Dental Implants

As far as the natural teeth are considered, a distinctive and very essential characteristic stands out. There exists an interface on the contact surface of a living tooth's root and jawbone, called periodontal ligament. It is a flexible, shock-absorbent layer which provides the tooth root to be affixed on the jawbone steadily and also prevents the forces created during chewing and biting, from becoming excessively large by providing uniform distribution of forces along the root surfaces. It is experimentally obtained that, in the case of direct contact without an elastic interface, the forces acting upon the tooth roots and the jawbone are significantly larger. In addition, periodontal ligament forms a cushion between the contact surfaces so that it helps avert the large local stresses and harmful effects of high stress levels on the anatomical structures. By the way, one of the foregoing reasons of tooth loss is degeneration of the periodontal ligament [3].

When a dental implant is placed into the root of a lost tooth, it is impossible for the periodontal membrane to still maintain its position and continue its functions as in the original case. Hence, the artificial root in the form of a thread is in direct contact with the jawbone, which causes large forces and large stresses to occur. As a consequence, both the life of the dental implant diminishes and resorptions take place in the jawbone.

In order to cope with the aforementioned problems arising from the inelastic structure of the present dental implants, a new design involving a pliant, shock-absorbent interface in the upper structure of the dental implant system can be implemented. Such an elastic component that would mimic certain mechanical functions of periodontal membrane in the natural anatomical structure will play a very crucial role in a dental implant application by distributing the forces uniformly on the implant and decreasing the local loads.

1.5 The Advantages of a Resilient Dental Implant

The previous studies revealed significant advantages of a resilient dental implant compared to a rigid type implant system. The essential contributions of elastic behavior in a dental implant are given below. Accordingly, an elastic dental implant:

- By shock absorbing nature, avoids excessive, local biomechanical overload that would cause component fracture, prosthesis fracture and early implant failure due to early crestal bone loss; in other words, provides protective damping against occlusive shock forces, decreases the magnitude of induced stresses in case of impact loads, maintains progressive loading for implant-supported prostheses.
- Diminishes or delays the loads transmitted to implants and to the bone.
- Attenuates the stress concentrated at the implant-bone interface in case of impact loads.
- Keeps the induced strain gradients in bone within physiologic limits, by uniformly distributing stresses and damping the occlusal forces, which reduces peak stresses (stress concentrations)
- Precludes micro-fractures in bone and bone resorption due to overloading; reduces the decrease in density experienced by the bone around the implant during the course of time.
- Helps avert relative micromotions that are out of admissible range between implant and bone, which favorably affects the implementation of one-stage surgery, termed as immediate loading.
- Provides some mobility as in the case of a natural tooth, which helps especially in implant-tooth co-supported dentures by preventing differential mobility throughout the prosthesis that causes uneven load sharing between the supports of the system.

- By compensating dissimilar mobility, avoids implant overload and relieves the stress concentration that would occur on a rigid type implant supporting a fixed denture in combination with a natural tooth under static loading.
- Reduces the rate of instances of overdenture problems such as failure of framework, bar devices or other retentive elements.
- Helps prevent bone resorption and following loosening of osseointegrated dental implant that are caused by the greater stress concentration on rigid type implants at high load levels when functioning in concurrence with natural teeth during mastication.
- Provides the patients with increased comfort and the feel of a live tooth.

For free standing implants, elastic components act as stress absorbing damping elements under dynamic load, whereas for an implant connected with a natural tooth to support a denture, the elastic elements are effective both as a stress distributor under static loading and stress absorber under dynamic loading.

1.6 The Problems Related with the Previous Designs

Each of the benefits of an elastic dental implant listed in the previous section is revealed via different studies explained in the literature survey, in Chapter 2. Although each study contributed significantly to the basis of elastic dental implant concept, they have not yielded a practical and applicable solution for clinical use yet. The problems involved in the previous approaches can be traced as follows.

- Dissimilar and deficient mobility characteristics in the axial direction of the implant as compared with natural teeth
- Loosening of the internal connecting screw, that is used to fasten the abutment to the implant, due to loss of preload during loading, fracture of screw and/or abutment due to inadequate mechanical properties of elastic components/connectors (as in the IMZ implants)
- Fracture of the internal screw due to transmission of main forces directly to the screw and stress concentrations at the head of the screw and near the initial screw threads, which are stated as potential sites of failure
- Failure of elastic parts due to high stress levels

- Frequent requirement for replacement and maintenance of the abutment and fixation screw due to failure, which is highly undesirable in a dental implant treatment
- Lacking mechanical integrity, involving not stable, reliable or durable mechanical designs in terms of longevity of the system under different load cases
- Very complicated geometric designs in terms of manufacturability and easiness of use, which makes mass production and clinical implementation impractical
- Lacking certain material specifications for elastic components, rather assuming a material whose mechanical behavior is analogous to that experimentally obtained for natural teeth to be available
- Limited versatility, in other words, being applicable only for supporting removable complete dentures in complete tooth loss cases (not for single tooth loss cases)
- Injury of gum tissue and gingiva inflammation in the designs in which the elastic parts are positioned at the implant neck region and have contact to the surrounding gingiva since the mobility of the part might cause irritation of this soft tissue during loading
- Inevitable moisture contamination of elastic parts that are exposed to the oral environment, requiring cleaning and/or maintenance at certain time intervals
- Not enabling modifications (like cutting to a specific geometry) on the abutment before fastening the artificial crown, while this is a common practice employed by prosthodontists
- Remaining only at the conceptual design stage, involving no detailed designs
- Ignorance of nonlinearities of the contact regions between elastic and rigid parts during mechanical design and analyses of proposed designs

1.7 Motivation and Objective of the Thesis

Having stated the advantages of a resilient dental implant (in Section 1.5) and the problems involved in the previous attempts to come up with such a design (in Section 1.6), the objective of this thesis study is to make a design of a novel dental implant system that would mimic the mechanical behavior of the periodontal ligament in natural teeth and tackle with the problems that appear in the previous designs. Particularly, the major points addressed in this study are:

1. Overcoming the problem of loosening of the internal fastening screw,
2. Obtaining a very similar mobility behavior (force-displacement relationship) for the dental implant as that of a natural tooth by simulating the mechanical behavior of the periodontal ligament, which will especially be useful for implants supporting dental bridges in combination with natural teeth,
3. Coping with the failure of the internal screw and elastic components due to stress concentration by design optimization,
4. Avoiding the soft tissue irritation problem by making a design of dental implant that confines the elastic components within a closed system,
5. Achieving a simple, solid mechanical design with a reliable mechanical integrity so as to provide an elastic dental implant practical for mass production and proper for clinical implementation.

1.8 Scope and Outline of the Thesis

The scope of the thesis is explained by making a descriptive outline for parts of the thesis.

Chapter 1 is devoted to supply the basic related introductory information in the field of dentistry and dental implantology. In particular, the role of the periodontal ligament for natural teeth, the importance of simulating its mechanical behavior in dental implants and the problems with the previous attempts are explained.

Chapter 2 is organized to compile sufficient and essential background information from the literature, within the context of the framework of thesis objectives, in order to be able to provide a contributory study from the mechanical engineering discipline to the interdisciplinary field of dental implants. Specifically, the biomechanical considerations for natural teeth and for dental implants, the approaches and important considerations about the concept of incorporating an elastic interface into dental implants are handled.

Chapter 3 deals with description of the methodology implemented to achieve the objectives of the study. In addition to the design procedure of the proposed dental implant system at the beginning, the optimization of the design and the comparison with a rigid counterpart by using finite element method are explained in detail.

Chapter 4 presents all the results obtained by applying the analytical procedure and the finite element analyses in the previous chapter.

Chapter 5 is devoted to summarize the work done and to briefly state the important conclusions drawn from the results obtained. Finally, some recommendations for future research on the subject are made.

CHAPTER 2

LITERATURE SURVEY

2.1 Biomechanical Considerations for Natural Teeth

In order to design a mechanical part as a replacement for a natural tooth, it is necessary to understand the inherent biomechanics of interrelated natural structures like tooth, periodontal ligament and surrounding jawbone. From the mechanical point of view, these anatomical structures involve interfacial tissues that support masticatory forces and external loads by providing proper force transmission among each other.

2.1.1 Loads on Natural Teeth

Natural teeth are subjected to different load levels during chewing and biting. The mechanical loads and corresponding response in terms of tooth displacement, termed as tooth mobility, have been investigated by several researchers. When a natural tooth is loaded, the force exerted has both horizontal and vertical components, combination of which produces tooth movement in all directions.

The highest levels of masticatory forces are reached when the teeth on upper and lower jaw are in contact during chewing cycles. The periods of contact last about 100 milliseconds adding up to 15 to 30 minutes each day. However, parafunctional habits like bruxism cause the load bearing tissues to be subjected to forces up to 11 N for several hours per day [39].

The maximum forces, created due to chewing and biting, have been determined by several experimental studies. Graf et al. conducted a study to measure the functional forces generated during mastication by placing transducers within fixed and removable prosthesis. The results indicated that in humans, maximum biting forces could reach up to 800 N in the molar region, and 100 to 200 N in the incisor region [40]. On the other side, Anderson reported that maximum occlusal forces due to chewing and swallowing (70 to 150 N) are considerably lower than maximum bite forces. Moreover, it was stated that in most cases these forces do not exceed even 10 N [41].

It is asserted that bite forces are affected by gender, age, height, weight, type of functional dental occlusion depending on facial type (short, average or long) and presence of parafunctional habits. Abu Alhaija et al. carried out a study to compare the effects of different factors on maximum occlusal bite force (MBF) in first molar region, in Jordanian students. According to the results of this study, those with a short face had highest MBF (679 ± 117 N) while the long face types had the lowest MBF (453 ± 98 N). The average MBF (599 N) for males was higher than the value for females (546 N). On the other hand, a positive correlation was derived between average MBF, weight and height. The average MBF in Jordanian adults was 573 N [42]. Bakke reported that bite force has an increasing pattern with age (until the age of 50), weight and height [43]. Carlsson et al. recorded higher level of bite forces for those having parafunctional habits like bruxism [44].

The state of dentition is another factor having impact on the maximum bite force. It is reported that patients with complete removable dentures have values just one third to one fourth of those with complete natural dentition, while individuals using implant-supported dentures have almost similar values as ones with a full natural dentition [39, 43]. This result shows that implant treatment provides a mechanical replacement with a close load bearing capacity as in the natural case. With respect to another study, the results of which justify this conclusion, Bousdras et al. reported that the maximum vertical bite force in humans with implant-supported superstructures was found to approach 690 N to 800 N [45], which is the value of maximum bite force with natural teeth for humans recorded by Graf et al. [40]

The force-displacement relationship for a natural tooth is nonlinear and time dependent, which is the fundamental mechanical consequence of the presence of fibrous tissue, called periodontal ligament, surrounding and supporting tooth roots. The functions of this layer with respect to mechanical perspective are explained in details in the following section.

2.1.2 Mechanical Behavior of the Periodontal Ligament

Periodontal ligament (PDL), the connective tissue separating the tooth from its bony socket, dominates the biomechanics of normal healthy teeth. When teeth on upper and lower jaw are in good occlusion, each tooth meets its functional opponent and forces in the direction of the long axis of teeth are induced. However, during chewing, teeth are under continual forces from multiple directions either simultaneously or alternately. The role of the PDL becomes prominent and should be underscored at this point. The essential function of the PDL is to provide retention and support for the tooth in its socket in jawbone. This is accomplished by the connective tissue fibers (Figure 2.1) which are arranged in a pattern that a single one or a group of fibers always counteract the forces acting, irrespective of the direction of forces. As a consequence, the fibers of the PDL not only provide support for the tooth, but also constrain and in a sense control its displacement so that the stresses transmitted to the jawbone are kept within normal physiologic limits [3]. On the other hand, the micromovements of PDL provide the distribution forces along the root surfaces so that high levels of local stresses are precluded [46].

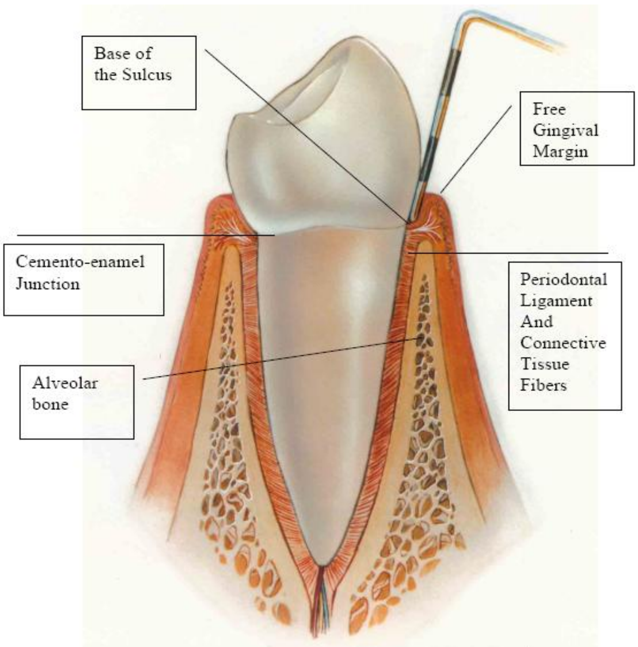


Figure 2.1 Connective tissue fibers in periodontal ligament of a healthy tooth [46]

2.1.2.1 Experimental Studies on the Mechanical Behavior of the PDL

Dental occlusal analysis, carried out to investigate displacement and deformation behaviors of periodontal ligament in response to an applied load, has been an important issue for researchers working on dental biomechanics. There have been many attempts to reveal mechanical properties and behavior of PDL by applying experimental techniques both on animals and humans.

Prior to explaining load-displacement characteristics of PDL, it is meaningful to mention about this layer's physical and mechanical properties. It has been obtained that varying from tooth to tooth and in different areas of the same root, for humans, the thickness of PDL in a healthy tooth in favorable occlusion is in the range from 0.15 to 0.39 mm [48]. The layer is thinnest at the middle region of the root. It is the functional movement of the tooth that maintains the typical thickness; otherwise, when opposing tooth is lost and occlusal forces become absent, the ligament starts to get thin and lose density [39].

Poisson's ratio of PDL is reported to be in the range of 0.45-0.49 (i.e. nearly incompressible like many other soft tissues), whereas a certain value for modulus of elasticity is not stated since the mechanical behavior of this tissue is pointed out to be non-linearly elastic. It was found that with the increment of load, Young's moduli increased exponentially [49]; specifically, the values determined were approximately 0.12 MPa under load ranging between 0 and 0.5 N, 0.25 MPa between 0.5 and 1.0 N, 0.44 MPa between 1.0 and 1.5 N, and between 0.69 and 0.96 MPa for loads within the range of 1.5-2.0 N. The corresponding diagram showing the relationship between the magnitude of applied load and the resultant tooth displacement is presented in Figure 2.2 [50].

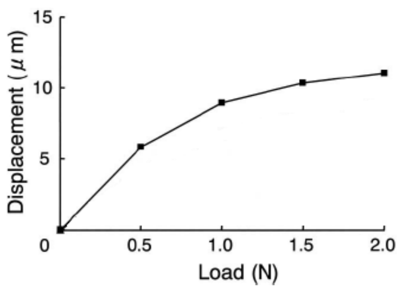


Figure 2.2 Tooth displacement as a function of magnitude of load [50]

For higher levels of load Rees and Jacobsen reported that a wide range of values (up to more than 1 GPa) has been used for the elastic modulus of periodontal ligament in previous studies. However, the authors concluded that, by assuming linear behavior, 50 MPa resulted good correlation between the finite element model for tooth loading and the experimental findings about tooth displacements. In addition, in this study an ultimate tensile strength of 2.4 MPa was reported [51].

The periodontal tissue has a unique and distinctive mechanical character. As a result of the membrane’s resiliency, a natural tooth shows physiologic micromovements in horizontal and vertical direction when loaded. The magnitude of displacement in horizontal direction is larger than that of vertical [27]. Morita et al. conducted a study to correlate displacement and deformation distributions with an experimentally obtained load-displacement curve by using digital image correlation analysis. For the experimental part, specimens consisting of a molar tooth, its periodontal ligament and a portion of alveolar bone were prepared by separating from a porcine mandible and a compressive force was applied using a table-top material tester. Based on the results obtained at the end of the study, it was concluded that the vertical displacement of a tooth can be characterized by a nonlinear biphasic behavior, as expressed in many previous studies. Regarding this characteristic nature of behavior, it was demonstrated that the first phase indicated deformation of the PDL, whereas the second indicated deformation of the alveolar bone and tooth. As represented in Figure 2.3, it was also noted that in the first phase (between 0 to -80 μm) the rate of increase of load was relatively small, whereas in the second phase (after -80 μm) this rate was higher resulting in diminished tooth mobility [52].

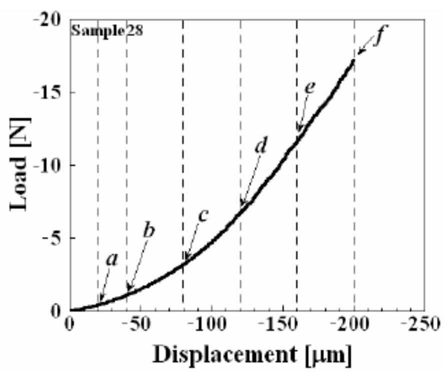


Figure 2.3 Load-displacement curve of the specimen under compressive load [52]

Taking the strong similarity between bovine and human PDL into consideration, Pini et al. also conducted an experimental study to examine the mechanical response of the bovine PDL subjected to uniaxial tension and compression. The specimens were tested in a custom made uniaxial testing machine. The stress-strain curve of the PDL obtained from a molar specimen is presented in Figure 2.4. The experimental procedure included firstly a compressive loading (CL) up to a reversal point followed by unloading (CU) to zero strain. After the initial compression cycle, the test specimens were loaded in tension until rupture occurred (TL). As it can be observed from Figure 2.4, the PDL's behavior under compression was biphasic such that stress did not increase much up to a level of compressive strain, after which a steep increase in stresses stood out. Another noteworthy point is the hysteresis loop exhibited in the compression cycle. In this study, the mechanical parameters such as elastic modulus (ε_f), maximal stress (S_{max}), corresponding maximizer strain (E_{max}) and strain energy density (U_E) were also analyzed and found to be influenced by PDL thickness and the tooth type (molar or incisor). With regard to the stress-strain relationship observed from the results, it was concluded that the classical linear assumption should be avoided and the PDL should be characterized by a nonlinear and time-dependent mechanical behavior [53].

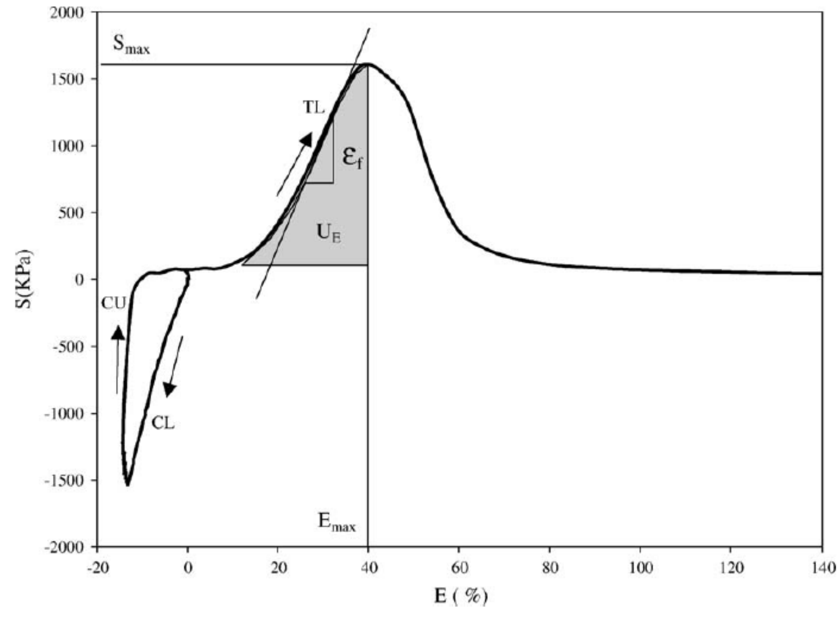


Figure 2.4 Stress-strain curve of the PDL obtained from a bovine molar tooth [53]

Toms et al. investigated the mechanical behavior of the PDL under orthodontic loading by using cadaveric specimens taken from mandibular premolars of human donors. The specimens were tested both under compression and tension load conditions. The results indicated that the human PDL has a nonlinear behavior depending on age, location of tooth on the jaw and direction of applied load [54].

The maximum tooth displacement that the PDL can allow and withstand is a subject that has been studied experimentally by several researchers. If the fibers constituting the PDL are subjected to intense forces exceeding the physiologic limits or abnormally misdirected forces are applied, these supporting fibers usually suffer. In case of such an extensive load, pain receptors indicate the sudden overload so as to protect the tooth from detrimental impacts [3]. Noda et al. reported the maximum tolerated value of tooth movement without causing any damage to be 0.26 mm, which is within two-thirds of the maximum width of PDL. The patients that were involved in the study did not report any pain until this limit of recorded tooth movement. This was attributed to the fact that blood circulation was still maintained over remaining one-third width of the PDL [48].

The experimental findings reported in the study conducted by Gei et al. verifies the results for tooth mobility obtained by Noda et al. Tooth mobility curves under compressive vertical force for a molar and an upper incisor based on experimental data are presented in Figure 2.5. It can be observed that greater forces are required for the same amount of displacement for molar compared with the incisor. In addition, for both ones the mechanical behavior appears to be biphasic throughout the range of displacement. After reaching a displacement of 0.02 mm, stiffness increases for a molar tooth. In this figure, the broken lines represent the results of two different linear interface models, whereas the solid line illustrates the results of the proposed nonlinear PDL model. Regarding these results, it can be apparently seen that linear interface models are not capable to predict the prominent characteristics that are caught by nonlinear modeling. Although the experimental curves were not exactly fitted by the curve obtained via nonlinear modeling, the general behavior involving two phases is approached observably to a large extent by using the nonlinear formulation as a constitutive law for modeling the mechanical behavior of the PDL [55].

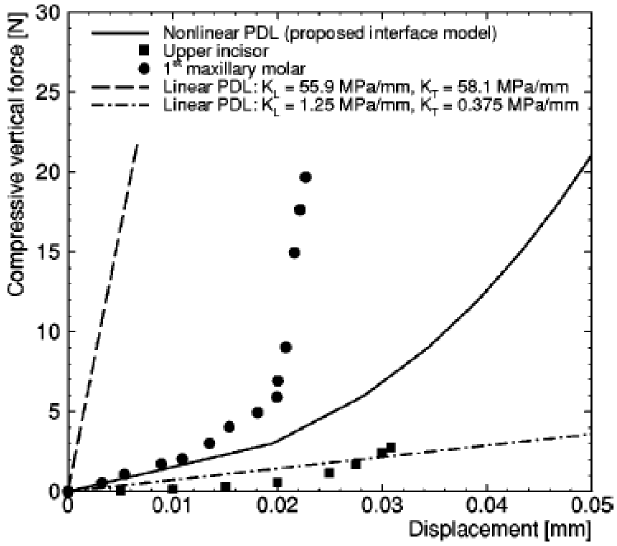


Figure 2.5 Tooth mobility curves in compression: the curves formed by symbols refer to experimental results, solid line refers to nonlinear model and broken lines refer to linear models [55]

2.1.2.2 Mathematical Models on the Mechanical Behavior of the PDL

The PDL strongly influences the stress state of the jawbone-teeth system. Owing to the fact that the length scales between the PDL and jawbone-teeth system are very different and the PDL has a very complex mechanical behavior, there exist significant difficulties in mechanical and analytical modeling of it. However, before designing reliable and effective dental implants, it is crucial to analyze and reveal the sophisticated biomechanical behavior of the PDL.

In order to investigate the typical nonlinear, anisotropic, time dependent response of the PDL, a constitutive model was generated in a collaborative study carried out by Natali et al. [56] The PDL presents anisotropic behavior due to the composite nature of the tissue caused by the distribution of the fibers, whereas time dependent viscous response is a consequence of rearrangement of the fibers in the solid matrix and corresponding microstructural modifications under loading. Hence a viscoelastic model was proposed and the constitutive relations were implemented in a finite element code. The biomechanical response was formulated by an Helmholtz free-energy density, defined as the difference between the instantaneous hyperelastic energy stored and the energy dissipated due to viscous effects given as

$$\psi(C,q_i)=W(C)-\frac{1}{2}\sum_{i=1}^{n}q_i:C \tag{2.1}$$

where C is the right Cauchy-Green tensor $C=F^TF$, F being the deformation gradient tensor. The instantaneous response, $W(C)$, accounts for the mechanical response when the tissue is subjected to strain rates so high that viscous effects can be neglected and it is defined as:

$$W = W_m(I_1, I_2) + W_f(I_4) + W_{mf}(I_1, I_2, I_3, I_4) \tag{2.2}$$

where W_m represents the contribution of the composite matrix, W_f is the contribution of the composite fibers, W_{mf} is the contribution from the interaction between matrix and fibers; I_1, I_2, I_3 are the invariants of the right Cauchy-Green strain tensor:

$$I_1 = tr(C), \quad I_2 = \frac{1}{2}[I_1^2 - tr(C^2)], \quad I_3 = J = \det(C) \tag{2.3}$$

while the fourth invariant I_4 represents the square of stretch λ of the collagen fibers:

$$I_4 = a_0Ca_0 = \lambda^2 \tag{2.4}$$

For the W_m term Mooney Rivlin model was used:

$$W_m = C_1(I_1 - 3) + C_2(I_2 - 3) \tag{2.5}$$

The second Piola-Kirchoff stress tensor representing the stress state of the body with reference to undeformed configuration was expressed as:

$$S(t) = 2\frac{\partial\psi(C(t), q_i(t))}{\partial C} \tag{2.6}$$

On the other hand, in vitro experiments were applied to some specimens of PDL taken from pigs to obtain stress relaxation and cyclic stress-strain curves. Then, the experimental results were compared with the numerical ones obtained by finite element analysis. In spite of the limited experimental data, the viscoelastic model was successfully validated as shown in Figure 2.6.

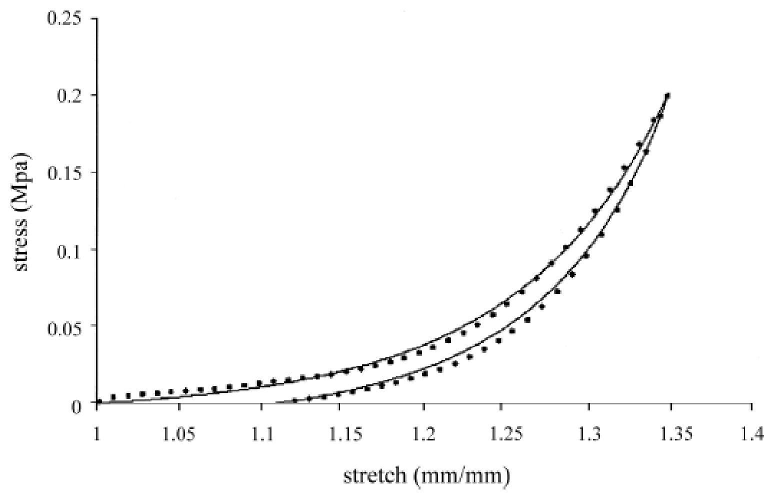


Figure 2.6 Stress versus stretch curve and hysteresis curve at a 0.1 mm/s elongation rate. (Dotted Line: Experimental Results, Continuous Line: Numerical Simulation Results) [56]

Limbert and Middleton proposed another novel constitutive model for PDL based on the definition of a compressible isotropic hyperelastic free energy function [57]. In their study, the material was assumed to be a continuum fiber-reinforced composite; material coefficients of Mooney Rivlin model corresponding to mechanical characteristics were obtained by using a least-squares-fit procedure to experimentally obtained data available in the literature. The stress-stretch curves obtained both analytically and experimentally are illustrated in Figure 2.7. It can be observed that the behaviors in tension and compression cases are distinct. In the compression case, as deformation increases, the required force for further movement should also be incremented, in other words, the layer shows stiffening effect for higher strain levels.

The spatial version of elasticity tensor given in Equation 2.7 was implemented into ABAQUS, a commercial finite element code, and an idealized compressible, anisotropic, nonlinear model of the PDL was developed.

$$(\mathrm{A}^{S})_{ijkl} = \frac{1}{J} F_{iI} F_{jJ} F_{kK} F_{lL} (2 \frac{\partial \mathrm{S}}{\partial \mathrm{C}})_{IJKL} \tag{2.7}$$

As it can be seen from the aforementioned studies, in addition to experimental studies, different approaches are used to reach a mathematical constitutive model that would describe the complex, nonlinear viscoelastic mechanical behavior of the periodontal ligament. Currently, this subject is an important research area, since an accurate and reliable model will enable to approach more to the natural tooth structure while designing novel dental implants. In Section 2.3, studies on developing a model for an elastic component in a dental implant that would simulate the PDL are reviewed.

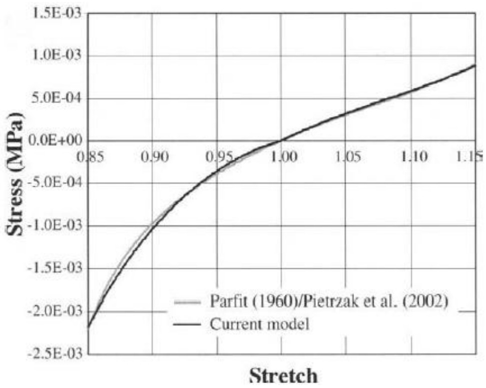


Figure 2.7 Stress-stretch curves for the PDL by fitting experimental data [57]

2.1.3 Structure and Mechanical Properties of the Jawbone

Bone, in general, is composed of two distinct osseous tissues with different structural and mechanical characteristics: the cortical bone and the cancellous bone (Figure 2.8). The jawbone involves the same bone architecture.

The cortical bone tissue is in the form of a dense, compact shell surrounding the more delicate cancellous bone. It contributes to the major functions of bone such as supporting the body, protecting organs, providing levers, storing and releasing chemical elements (like calcium). The cortical bone is harder, stiffer and stronger than the cancellous bone [58]. The average thickness of cortical bone of the jaw is reported, by Chaichanasiri et al., to be 2.2 mm, which is a value obtained by averaging from computed tomography (CT) images [59]. Miyamoto et al. reported that the increase in cortical bone thickness enhances dental implant stability [60].

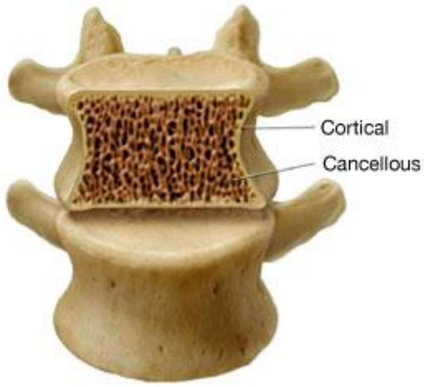


Figure 2.8 The tissues forming the bone architecture [58]

The cancellous bone, also called as trabecular bone or spongy bone, is a porous, less dense, less stiff, softer and weaker tissue. It has a higher surface area in comparison with the cortical bone; therefore, it is metabolically more active and seriously affected by bone remodeling. If the continuous state of degeneration is uncoupled with regeneration, bone density starts to diminish resulting in bone resorption. The decrease in bone density is an unfavorable situation for dental implant treatment since the amount of bone cells surrounding the implant directly influences the stability of the implant [58, 61].

The mechanical properties of bone is related to bone mass, bone remodeling frequency, bone size and area, bone microarchitecture and the degree of bone mineralization. In addition, factors like age, anatomical site, liquid content, etc. are influential on the mechanical properties. Adopted from literature, typical values of Young’s modulus and Poisson’s ratio, for bone tissues, dental tissues and artificial porcelain crown, are presented in Table 2.1.

Table 2.1 Mechanical properties of bone and dental tissues [49, 59, 62, 63]

Material	Young's Modulus (GPa)	Poisson's Ratio
Cortical Bone	14.5	0.32
Cancellous Bone	1.37	0.30
Enamel	84.1	0.20
Dentine	18.6	0.31
Porcelain Crown	67.2	0.30

Bone density, an important parameter indicating the bone quality and affecting the biomechanical behavior under loads, can vary in different locations, ages, gender, states of health etc., remaining in the range of 1.7-2.0 g/cm^3 for cortical bone and 0.23-1.0 g/cm^3 for cancellous bone [64].

2.2 Biomechanical Considerations for Dental Implants

Dental implants are used as a replacement of lost or partially damaged teeth to regain masticatory functions. Although an artificial replacement can never serve perfectly as in the natural case, the aim is to approach the ideal structure as much as possible.

A dental implant is a product resulting from the implementation of concepts and knowhow involved in multiple disciplines including biomechanics, surface chemistry, biology and physics. Long-term success of a dental implant depends on how well and broadly the interrelated factors from different disciplines are handled during the design of the dental implant. Besides, the host factors, conditions like bone quality and quantity existing in the implantation site of the patient, are also influential on the success of dental implant treatments.

The service conditions in the mouth are hostile for intraorally placed parts. A dental implant is in direct contact with the host bone, which requires special surface properties. On the other hand, all the teeth continuously remain in an environment of saliva, an aerated aqueous solution with a pH value in the range of 5.5 to 7.5 [65] and varying temperatures of 33.2 to 38.2 °C [66]. However, sufficiently satisfactory results have been obtained by making optimization regarding parameters like biocompatibility, strength, fatigue durability, nontoxicity, corrosion resistance, and also aesthetics in the design of dental implants [67].

In this section, dental implants are dealt with from biomechanical point of view and requirements with respect to biomechanical performance are discussed in detail. Firstly, the biomechanics of implant-bone interface including the effects of quality and quantity of bone and the concept of bone remodeling is reviewed. Secondly, the biomechanics underlying dental implant systems are explained and factors to be considered in the design of a dental implant are compiled. Thirdly, in line with the aim of the thesis, the studies involving the concept of an elastic interface for dental implants carried out up to the present are presented.

2.2.1 Biomechanics of Implant-Bone Interface

As far as its mechanical role is concerned, a dental implant system has the primary function of supporting the artificial crown and transmitting occlusal loads acting on the crown to bone tissue. The interaction between the implant and the supporting bone tissue is of great importance for success of the implant. In addition to the surface characteristics of the implant, the quality and quantity of bone and bone remodeling phenomena are the key factors determining the nature of the interaction.

2.2.1.1 Effects of External Bone Quality and Quantity

The success rates of dental implants strongly depend on the initial stability of the implant immediately after surgical placement and in early stages of healing. Previous studies have indicated that varying bone quality and quantity in different anatomical regions are responsible for the initial clinical performance.

In the regions where bone density is low, it is difficult to maintain stable fixation due to insufficient contact between implant surface and bone. As a result possibly occurring micromotions at the interface will adversely affect the osseointegration of the implant with bone [68]. On the other hand, bone strength is directly related with bone density. If the cortical bone layer is thin and the cancellous bone has a comparatively low density, the capability of bone to endure occlusal loads will decrease [27]. In addition, as bone density diminishes, the stiffness tends to decrease [69]. Hence, it can be concluded that bone density determines the mechanical properties of bone, affects the stresses and strains under loading, thereby, is an important parameter for implant survival. The most favorable qualitative state of bone for dental implant success is the case where homogenous thick cortical bone surrounds the dense cancellous bone. It is also reported that the bone quality of mandibular bone is superior to that of maxillary bone [68].

Beside the qualitative properties, bone quantity is also important for a dental implant application. The vertical bone height is required to be adequate so that remaining bone region beyond the implant could resist the occlusal loading. In some cases, the bone height is not sufficient even to place a dental implant. This deficiency of bone volume is mostly due to bone resorption. In such cases so-called artificial bone cement is added to deficient regions [27].

2.2.1.2 Bone Remodeling Around Dental Implants

Bone remodeling is a continual state of adaptation of bone to dynamic mechanical stimuli. It is composed of recurrent successive processes, namely, resorption and regeneration. For an equilibrium state (of strain, stress or strain energy density of cyclic loading), these processes are always active and perfectly balance the impact of each other.

Bone resorption occurs as a result of certain decrease in the magnitude of mechanical load, whereas bone regeneration is induced by a certain increase in mechanical load transmitted to bone tissue during chewing cycles or biting [64]. In other words, when strain intensity is higher than equilibrium strain, formative activities of osteoblasts accelerate and net bone deposition occurs. However, when strain intensity becomes lower than equilibrium strain, degenerative activities of osteoclasts becomes predominant and net bone resorption takes place. The induced local strain is associated with applied load, geometric and material properties of bone, while remodeling depending on local strain modifies these bone properties. These processes follow each other in a closed feedback loop [61].

Instead of static loading, bone remodeling is triggered by dynamic loading with short durations of loading. It is reported that extended loading duration reduces the adaptation effectiveness of bone to the loading condition [64]. Apart from that, if the magnitude of applied load becomes excessively large beyond the physiological limits that bone tissue can withstand, high stresses and strains occur at the implant-bone interface, thereby damaging the osseointegration and raising the risk of implant failure [27].

In case of a tooth loss, if not treated properly by a dental implant application, the transmitted load to bone tissue which is the major stimuli for regeneration process becomes absent leading to uncoupled resorption process. As a consequence bone density and bone height appears to decrease, which means that both bone quality deteriorates and bone quantity diminishes. Such a case will unfavorably affect a potential implant treatment, even may preclude surgical placement of a dental implant due to inadequate bone quality and quantity. Moreover, Genna reported that the chance of failure of dental implants that are placed in regions of the jawbone with lower density is higher [70]. From this point of view, bone quality maintained by bone remodeling is a critical factor to ensure longevity of a dental implant.

Bone remodeling should be taken into consideration in the design of dental implants. It is stated that different designs may lead to different success rates of implant restoration [64]. Moreover, A. Eser et al. carried out a study to predict biomechanics of bone around cylindrical screw dental implants with different macrogeometric designs under simulated immediate loading conditions and concluded that bone response differs in terms of remodeling for two distinct implant designs [71].

2.2.2 Biomechanics of the Implant System

There are several factors affecting the biomechanics of a dental implant system, thereby, the success or failure of the implant. These can be traced as:

- Magnitude, direction and location of applied loads on the artificial prosthetic teeth,
- Type and geometry of the prosthesis (whether it is a single tooth or a bridge composed of connected multiple teeth),
- The number and location of the implants on the jawbone,
- The rigidity of the implant system and implant-bone interface,
- The angulation of the abutment and the nature of the connection between the artificial tooth and implant components,
- The geometrical design parameters of the implants such as shape, length, diameter etc.
- The material and surface properties of the implant.

All the factors mentioned above have impact on the stability of the structure and the magnitude and pattern of stress throughout the implant system. Hence, in order to better understand the biomechanical properties of a dental implant system and to evaluate the efficiency of such prosthetic systems, it is crucial to disclose the effect of each factor to a more detailed extent.

2.2.2.1 External Loads and Force Transmission

The nature of applied loads and the manner how these loads are transferred from prosthesis to bone through implant system are critical factors that determine the stress distribution on the overall structure [72]. The implant's survival depends on whether stresses remain in the physiologic limits of bone and below mechanical strength of the parts constituting the implant system.

The previous successful clinical results indicate that osseointegrated dental implants, as shown in Figure 2.9, can withstand bite forces comparable to the values on natural dentitions [27, 65]. Therefore, from an implant-supported tooth, it can be expected to safely maintain its structure and functions without complications at load levels up to 800 N, which is the value reported to be the maximum tolerable vertical bite force for a natural tooth (as presented in Section 2.1.1).

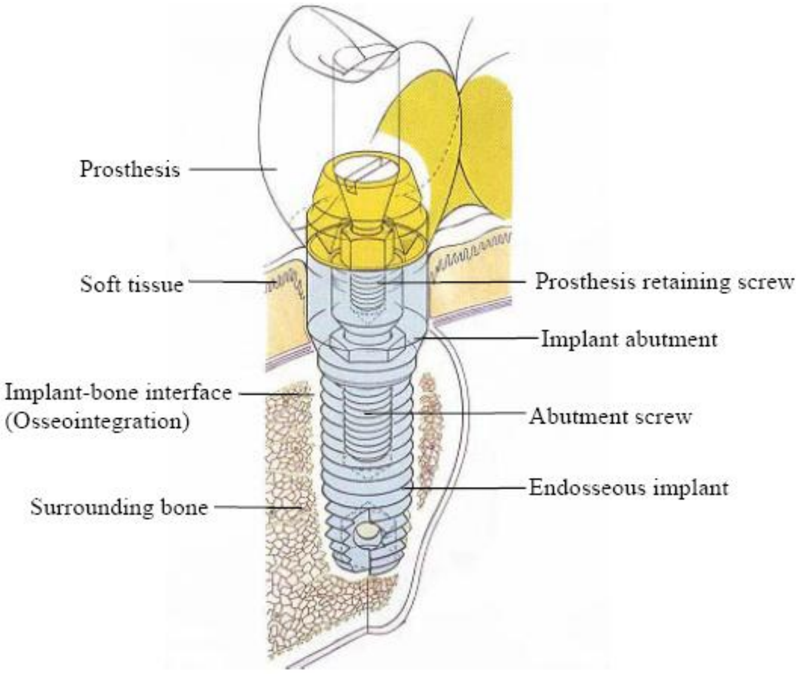
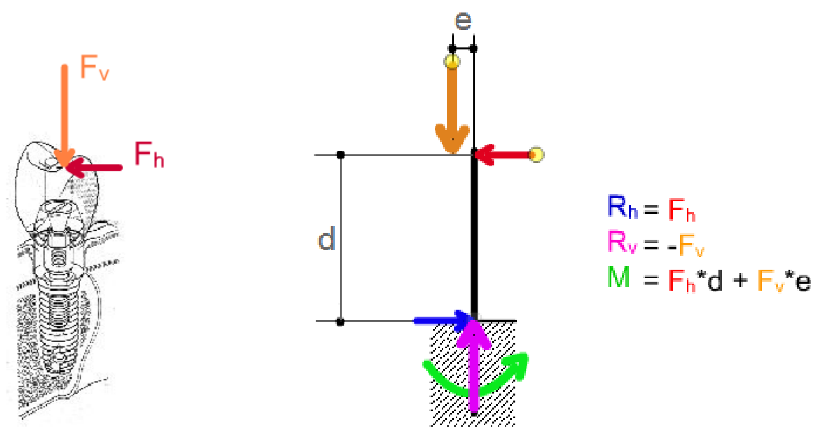


Figure 2.9 A dental implant system (Bränemark implant) integrated with bone and supporting a prosthetic tooth [73]

It is important to point out that the forces on dental implant systems are vector quantities that create reaction forces and moments at the side of fixation in bone. As represented in Figure 2.10, two types of loading, namely horizontal (lateral) and vertical (axial, occlusal), may act on a dental implant. The combinations of these forces form oblique loading which simulate a more realistic loading condition. It is reported that oblique loading could result in highest local stresses in cortical bone [73]. It is difficult to define the forces and investigate the mechanical behavior of such complex systems by analytical methods, especially when the implant is a part of bridge restoration. Therefore, researchers have commonly benefited from the complex problem solving capabilities of finite element (FE) analysis technique to analyze dental implant systems.

The resultant forces of the stresses within the load-bearing components of the implant-supported restorations are counterbalanced with the reaction stresses that occur within the surrounding bone. The reaction stresses create a resultant reaction force with the same total magnitude as the external load but in the opposite direction. As a result, the static equilibrium is maintained.



(a) **(b)**

Figure 2.10 Forces acting on an implant-supported tooth: (a) Forces on a 3-D model, (b) Action and reaction forces and moments on a simplified model [74]

The forces in vertical direction are shown to be better tolerated [75] and it is reported that the magnitude of vertical forces are up to ten times the magnitudes of horizontal forces [76]. On the other hand, the location of application of load is important that it affects the magnitude of axial component. In the molar region the bite forces are higher than four times the values in the incisor region [27].

Dental implants can be used for both single and multiple tooth replacements. In the case of multiple tooth loss, it is a widespread method that a dental bridge supported by both implant and natural tooth are formed in order to fill the toothless region. In such a restoration, the loads acting on the prosthesis during biting or chewing are shared among the implant(s) and natural tooth supporting the bridge, but not with uniform distribution. El-Wakad, by carrying out a 2-D FEM analysis, showed that about 70% of the total vertical load of 100 N and 72% of total horizontal load of 25 N acting on such a dental bridge was countervailed by the implant support and remaining load was acting on the natural tooth support [77]. This load concentration on the implant can be attributed to the rigid structure of the implant compared with the structure of a natural tooth support involving an elastic behavior due to the periodontal tissue at the connection interface with bone.

Regarding implant-supported bridges, partial or complete dentures, the number and location of implant supports are important. Eskitascioglu et al. revealed that for a fixed partial denture under vertical loading, higher stresses occur within bone and implant system when a single implant is used rather than multiple implant supports [78]. Therefore, it is advisable to evenly distribute the total load at multiple locations by using more than one implant support for such bridge or dentures.

Dental implants are composed of rigid parts and due to absence of a resilient layer like periodontal ligament at the interface, implant-bone connection is also rigid compared to the natural teeth. As a result of rigidity of the overall structure, the deflection of a dental implant under loading is determined only by the deformation of alveolar bone. It is revealed that the deflection of an implant-supported artificial tooth is 10-100 times less than the deflection of a natural tooth. As a consequence, at high load levels, bone resorption leading to loosening of the osseointegration occurs since the rigid structure is subjected to greater stresses when functioning in concurrence with natural teeth during mastication [27].

The tractions induced at the implant-bone interface may be compressive, tensile or shear. The compressive tractions within the physiological limits contribute to the maintenance of the integrity of the implant-bone interface, whereas combination of tensile and shear tractions tend to separate the connection at the interface. Actually compressive tractions occur due to vertical loads, while shear tractions are caused by the tensile forces arising from horizontal loading. The risk of failure increases as the shear component of the traction becomes larger [27]. That is why a dental implant can withstand a higher level of vertical load than horizontal load. What puts the destructive effect of shear forces forward is the absence of periodontal ligament on the surface of dental implants. Because it is the periodontal layer distinguishing natural teeth from dental implants with regard to response under shear effects by acting as an effective shear transfer layer and minimizing the stress concentrations in the bone [79].

2.2.2.2 Design of the Implant System

The essential design objective is to obtain a dental implant that is able to manage biomechanical loads and has biocompatible geometric, surface and material properties so as to provide long-term survival rate. Implant failures are attributable to either loss of biomechanical bond at the implant-bone interface or micro fractures within the components of implant system. The former one can be caused by two reasons: (1) insufficient osseointegration due to inadequate initial stability [80] or poor biocompatibility in the short term after surgical placement of the implant; (2) getting harmed due to lack of oral hygiene or overloading which causes bone resorption, loosening of the implant and relative movement at the interface. The latter one, on the other hand, may occur due to inaccurate design of the implant components [81].

The functional design objective can be achieved by incorporating a variety of factors via providing optimum requirements for each. These factors can be grouped as geometric, surface and material properties. The materials used for dental implants and their properties are presented in Section 1.3.2.1, and the effects of surface texture are described in Section 1.3.2.3. Therefore, in this section the subject of geometric design of dental implants is handled thoroughly.

The geometric properties of a dental implant have important effects upon both endurance of the implant against biomechanical loads and primary implant stability which is necessary for osseointegration.

The principles suggested by Vidyasager et al. for the geometric design of a dental implant are: (a) to achieve initial stability that would keep the micromotions between implant and bone within the tolerated range (below 100 μm [68]) and to minimize the required period of waiting to begin loading the implant; (b) to reduce the effect of shear tractions at the interface so as to preserve surrounding bone and the biomechanical bond between implant and bone; (c) to stimulate bone formation via remodeling process [80].

The chief design criteria are about the shape, size, thread profile, neck and abutment design of the implant. The effects of each criterion are explained separately as follows.

2.2.2.2.1 Implant Shape

Concerning the shape design, dental implants are generally categorized into cylindrical, conical, stepped, screw and hollow-cylindrical implant geometries. Siegele and Soltész conducted a study to compare the dental implants having these different types of geometries in terms of stress distribution generated in the surrounding jawbone by using FE analysis. Since only the influence of external geometry but not the effect of the size was investigated, implant samples having same diameter and length were utilized. According to the results, it was stated that different implant shapes lead to prominent variations in the stress distribution within the bone but substantial differences in maximum stress values were not observed. Particularly, it was reported that small radii of curvature (conical type) and geometric discontinuities (stepped or hollow-cylindrical type) on implant surfaces resulted in significantly higher stresses than smoother geometries (cylindrical or screw type) [82].

Tada et al. showed that maximum von Mises stress in trabecular bone was lower for the screw type implant when compared with non-threaded cylindrical implant under occlusal loading [83]. Furthermore, Holmgren et al. stated that stresses within stepped cylindrical implants were more evenly distributed as compared to straight type implants [84].

The initial stability of a dental implant is significantly less in low density regions of jawbone, which increases the risk of failure. In such cases the geometry of the implant becomes more important as it should adapt to the bone state [81]. The studies investigating the effect of tapered design revealed that tapered geometry enhances implant's primary stability by compressing the jawbone during and after implantation. Astrand et al. studied the survival rates of implants with different designs in soft-quality bone and concluded that tapered type had improved survival rate when compared with earlier results of Bränemark implants [85]. Besides, in the study comparing straight cylindrical implants with tapered ones, Friberg et al. showed that tapered implant design resulted in increased primary stability in the regions of bone having low density [86]. Moreover, tapered structure facilitates placement of dental implants into the cavities immediately after tooth extraction and to the regions of bone where neighboring teeth have convergent roots [81].

Contrary to the advantages regarding primary stability, conical shaped implants do not show desirable outcomes when tractions at the interface with bone are considered. It is reported that cylindrical implants produce a more uniform stress profile with less localized stress concentrations [87], lower stress levels [88] and lower strains [89] in bone when compared with conical type implants.

In short, it can be concluded that for the regions of jawbone having good bone quality, screw or stepped type cylindrical implants (Figure 2.11a) will give the most favorable results, whereas for the regions of low bone quality conical type (Figure 2.11b) should be preferred considering primary stability issue.

Figure 2.11 Different dental implant shapes: (a) Cylindrical type, (b) Conical type [90]

2.2.2.2.2 Implant Size

The implant size specified by length and diameter significantly influences the stress distribution in the surrounding jawbone, thereby, the success of the dental implant [81]. In general, wide and/or short implants present higher failure rates, defining a 'wide implant' as a device with a diameter of 4.5 mm or more and a 'short implant' as one with a length of 8 mm or less [91].

Dental implants with wide bodies pose an increased risk of failure in the long-term assessment. Shin et al. documented the 5-year cumulative survival rate (CSR) of implants with different diameters in posterior jaws. In this study statistically significant differences were observed such that implants having wide bodies (5 mm diameter) showed a CSR of 80.9%, whereas regular-diameter implants (3.75-4 mm) had a CSR of 96.8% [92].

Although wide implants result in higher failure rates, for the implants with diameters below the limit of 4.5 mm, as implant diameter increases, the stress levels in bone tend to diminish. Baggi et al. carried out a study to analyze the influence of implant diameter and length on stress distribution within surrounding bone under both axial and vertical load by applying FE analysis, using implants having diameters of 3.3 mm to 4.5 mm and lengths of 7.5 mm to 12 mm. According to the results, stress values and concentration areas decreased for cortical bone when implant diameter increased, while more effective stress distributions for cancellous bone were experienced as implant length was elevated. [93].

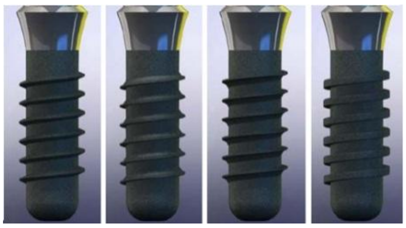
The impact of implant size is dependent on surrounding bone quality. Kong et al. evaluated the effects of the implant diameter and length on the maximum equivalent stress in jawbone of high quality with the aid of FE analysis. Using implants with diameters ranging from 3.0 mm to 5.0 mm and lengths ranging from 6.0 mm to 16.0 mm, it was obtained that under axial load maximum equivalent stresses in cortical and cancellous bones tended to decrease by 77.4% and 68.4% with increasing diameter and length, respectively. It was concluded that implant diameter was more effective on the stress distribution in dense bone and suggested that implant diameter exceeding 3.9 mm and implant length exceeding 9.5 mm were optimal combination of dimensions for a cylindrical dental implant in type II bone (second highest quality of bone), which involves a thick layer of cortical bone surrounding dense cancellous bone [94]. Via a similar study, Kongo et al. further investigated the effect of diameter and length of implant in jawbone of low quality and stated that implant length was predominantly influential, this time, in reducing bone stress and enhancing stability in type IV bone (least quality of bone), which has a low density. For this case, implant diameter exceeding 4.0 mm and implant length exceeding 9.0 mm were reported as optimal combination of implant dimensions [95].

In conclusion, short implants can possibly be used when there is less available bone height, but high density. In other words, low quantity but high quality bone condition is appropriate for short implants. On the other hand, for poor quality bone, within the bounds that bone height allows, longer implants should be preferred. Finally, for both low and high quality bone cases, remaining below the upper limit of 4.5 mm, higher the diameter, better the survival rate.

2.2.2.2.3 Thread Profile

Owing to the fact that dental implants do not involve a cushioning element like periodontal ligament to buffer occlusal loads, forces are transmitted directly to the surrounding bones. Possible consequences of this are microfractures at the interface with bone, bone resorption, fracture of the implant and loosening of components in the implant system [96]. To overcome these problems researchers have attempted to enlarge the contact area of implant with bone. In addition to increasing diameter and/or length of the implant, threads of special forms are incorporated on the surface of dental implants in order to maximize contact area with bone, which improves initial stability, distributes interfacial tractions more uniformly and protect bone from excessive local stresses and strains [80, 97].

The parameters handled in the biomechanical optimization of thread design for a dental implant include thread shape, pitch, width and depth. As the thread shape for dental implants, basically four types of threads have been proposed: V-shape, buttress, reverse buttress and square shape threads (Figure 2.12). Lee et al. compared symmetrical, square and buttress thread types with the same pitch to see the effect of shape on the induced stress on surrounding bone. According to the results, the greatest stress in the bone was concentrated at the root radii of the first thread below neck region for all types as obtained in several previous studies [98, 99, 100]; but the square thread had the highest contact area and lowest maximum stress [99]. Similarly, Steigenga et al. reached to the conclusion that square thread design had significantly more bone-to-implant contact and greater reverse-torque measurements compared with the V-shaped and reverse buttress thread designs [101]. Moreover, Chun et al. obtained that among various thread types the square-thread filleted with a small radius resulted in more evenly distributed stress within surrounding bone [100].



(a) **(b)** **(c)** **(d)**

Figure 2.12 Different thread profiles for dental implants: (a) V-thread, (b) buttress, (c) reverse buttress, (d) square threads [98]

The study of Chun et al. [100] also presented important conclusions about thread pitch, width and depth for dental implants. It was stated that stress distribution was most effective in the case when the width and the height of thread were $p/2$ and $0.46p$, respectively (where p is the screw pitch), whereas the related effects between the width and height were negligible. Furthermore, the maximum effective stress generated in jawbone tended to decrease as the screw pitch is reduced gradually; however the influence of decrease in pitch became insignificant after further decreasing the pitch below 0.9 mm [100]. Kong et al. compared the effects of different implant thread pitch values in the range of 0.5 mm to

1.6 mm and suggested that thread pitch exceeding 0.8 mm was optimal for a cylindrical implant in bones of type II and oversized pitch should be avoided [97].

It can be concluded that although most of the researchers agree on the superior results of square shaped threads, different results on thread pitch size, instead of standardized values have been presented. Therefore, optimization of specific thread designs should be performed in terms of mentioned thread parameters when a new dental implant thread profile is to be designed.

2.2.2.2.4 Neck Region

It is reported that the highest stresses in bone occur in the region surrounding the implant neck and in consistent with that bone resorption initiates around the implant neck [80]. Based on the aim of providing the cortical bone free of load, it was suggested to include a smooth neck portion at the top part of the implant remaining inside bone [3]. However, later studies revealed that significant loss of bone (up to 1-2 mm [102]) occurs in crestal region of bone around polished necks of dental implants and bone remodeling materializes as required from bottom only to the level where the screw threads and/or the rough surface texture ends. Therefore, it is suggested to incorporate a rough neck region that remains inside crestal bone in order to provide positive stress stimulus to bone for remodeling and to decrease bone loss, while the portion remaining above bone level should have a smooth surface since it is in contact with gingiva [80]. Circumferential grooves in the form of micro-threads, as shown in Figure 2.13, have been demonstrated to be quite effective to maintain cortical bone in the crestal region [80, 103]. In addition to micro-threads, micro surface roughness obtained by acid etching or other methods provide enhanced bone retention and adherence to the implant [102].

Sun et al. discussed the effect of the height of smooth region of the neck, in other words, the transgingival height of a dental implant on the maximum equivalent stress in jawbones by using FE analysis. Ranging the transgingival height from 1.0 mm to 4.0 mm, it was obtained that the transgingival height played a more important role under horizontal loads than under an axial load; and the optimal range of that height was 1.7-2.8 mm, leading to minimum stresses in the jawbone [96].

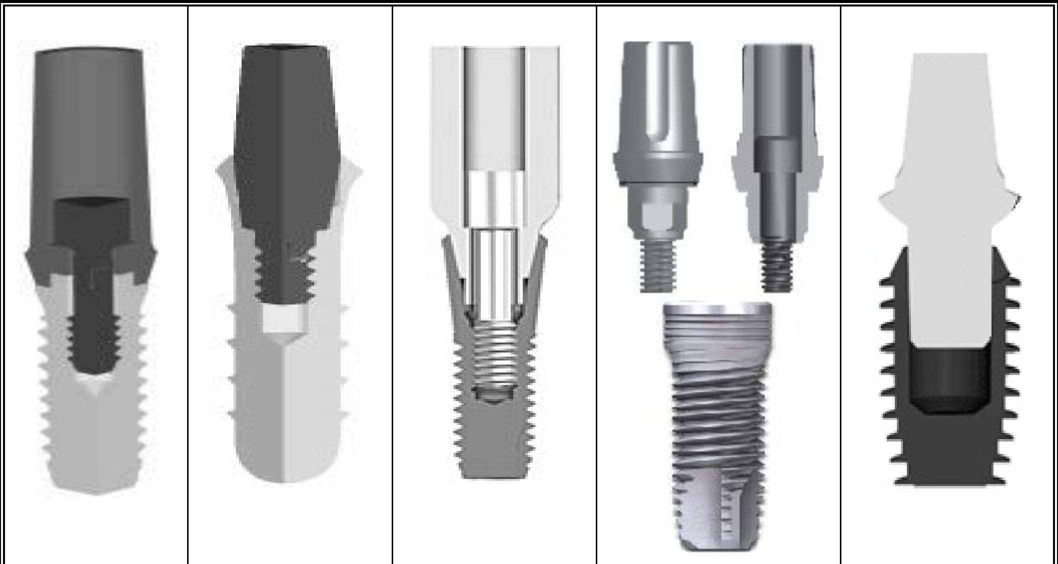
Figure 2.13 Micro and macro threads on a dental implant

2.2.2.2.5 Abutment-Implant Interface

A dental implant system generally consists of three main components, namely, implant, abutment, and internal screw that is used to fasten the abutment to the implant. The mechanical design of these components and interfaces between them substantially affect the mechanical integrity of the system, mechanical response within the system under loads, stability and strength of the system and long-term survival of the dental implant treatment [73].

In general, the presently commercial dental implants involve two main types of configurations for implant-abutment connection, namely, external and internal connections. The implants with external type connection include a polygon-shaped (usually hexagon) protruding part at the top, where the abutment is placed on and fastened with an internal screw. On the other hand, the implants involving internal connection show more variety. The mostly used abutment designs with internal type connection are one-piece conical type screwed abutment, two-piece conical type abutment with a separate internal screw, internal hexagonal abutment with a separate retention screw and conical interference fit abutment. These abutment-implant connection types are demonstrated in Figure 2.14.

Several investigators studied these different designs from the mechanical point of view. For the external hexagonal implant system (Figure 2.14a), by using FE analysis, Segundo et al. found that the highest stress concentration occurred at sharp angled corners of the interface between the implant platform and abutment, and at the midpoint of internal diameter of the first thread of retaining screw, which was stated to be more likely to fail earlier than other components. It was suggested to include less number of sharp angles at the neck, an additional washer to evenly distribute stresses and thicker implant walls to improve resistance to plastic failure [104]. For the internal one-piece conical abutment (Figure 2.14b), it was reported that the neck of the implant posed the highest risk to be a potential zone for fracture in case subjected to high bending forces [105]. Norton evaluated the strength of an internal conical interface compared to an external hexagonal connection and reported that the internal conical design had a dramatically enhanced ability to resist bending forces, which would promise a better clinical integrity for implant-supported prosthesis [106].



(a) External hexagonal-connection (Nobel Biocare) [107]	(b) Internal one-piece conical abutment (ITI) [107]	(c) Internal two-piece conical abutment (Astra) [108]	(d) Internal hexagonal-abutment (Neodent) [110, 111]	(e) Conical interference fit abutment(Bicon) [109]

Figure 2.14 Different types of implant-abutment connection configurations

Kim et al. [112] experimentally compared the fracture and fatigue strength of external hexagonal, one-piece internal conical and two-piece internal conical implant-abutment joint designs. The mean fracture strengths were 1153 N, 1261 N and 1110 N, whereas the mean fatigue strengths were 300 N, 600 N and 453 N for the external hexagonal, one-piece internal conical and two-piece internal conical joints, respectively. The study indicated that one-piece internal conical joint had the strongest connection. Likewise, Ribeiro et al. [113] compared the efficacy of external hexagonal, internal hexagonal and internal conical implant-abutment interfaces with respect to the fatigue resistance and mode of failure by carrying out an experimental study. According to the results, fatigue failure occurred in the internal screw. Besides, superior fatigue resistance of external hexagonal connection was observed in contrast to the study of Kim et al., while there was no significant difference between internal conical and internal hexagonal interfaces.

Saidin et al. [114] conducted a study to analyze the stress distribution on mating surfaces of internal conical and internal hexagonal implant-abutment connections under axial load by using FE analysis. The former one could transfer occlusal force uniformly through the implant, while the latter one produced stress concentration at its vertices, increasing the possibility to be fractured. On the other hand, the hexagonal connection provided the abutment to be locked to the implant, which resulted in reduced micromotion and higher stability. Similarly, Coppedê et al. [111] investigated fracture resistance of internal conical and internal hexagonal connections for dental implants under static oblique compressive forces by implementing an experimental procedure. The results showed that the friction-locking mechanics and solid design of the one-piece abutments of the internal conical connection system provided greater deformation and fracture resistance to the implant-abutment assembly. These findings agreed with the results of Saidin et al. [114] regarding tendency to fracture.

As for the bone-interfacial traction it was reported that internal cone designs of abutment connections provided lower values as compared with flat-top designs. The amount of marginal bone resorption around internal-cone and hex-connection type implants was almost the same according to in vivo observations, while the conical abutment configuration incorporated with retention elements at the implant neck experienced diminishing in the peak bone stresses [73].

One of the major complications encountered in dental implant applications is the loosening of the abutment screw under orthodontic loading. Loosening occurs when the preload clamping the abutment and the implant together vanishes due to applied forces [115]. Inadequate preload, misfit of the mating components and rotational characteristics of the screws are also responsible for abutment screw loosening or fracture [116]. In order to generate a specific preload on the abutment screw, corresponding tightening torque is to be applied to induce compressive forces across the abutment-implant interface and maintain the integrity of the joint. The tightening torque recommendations of manufacturers vary but typically range from 10 to 35 N.cm, depending on the design parameters [117].

The tapered implant-abutment interface outperforms the other connection types with respect to the resistance against loosening, therefore, has become more popular recently. Bozkaya and Müftü investigated the mechanical reliability and stability of tapered implant-abutment interface integrated with a screw (TIS) at the bottom of the abutment. In the study, analytical formulas were developed to predict tightening and loosening torques of the abutment. It was stated that the geometric properties of the screw, friction coefficient between contacting surfaces, the taper angle and contact length of the conical part of the interface and the elastic properties of the materials affected the mechanics of the system. It was shown that the loosening torque was smaller than the tightening torque, both correlated with the screw pretension. A definition of abutment efficiency, the ratio of loosening torque to tightening torque, was made and evaluated to be varying between 0.85-1.37 as the design parameters were changed. The efficiency values greater than 1 were obtained for specific values of taper angle and friction coefficient. Furthermore, almost all of the tightening load (86%) and loosening load (98%) were carried by the tapered section of the abutment, that is, the behavior of the tapered connection integrated with screw was governed by the tapered part. In fact, with certain combinations of the parameters the pretension in the screw might become zero. The retention in the tapered section was shown to rely on the large contact pressure and resulting frictional forces in the mating region. In short, conical interface was verified to successfully resist loosening at the implant-abutment interface [106].

Finally, the concept of tapered interference fit (TIF) abutment with no screw at all was implemented in the commercial product of Bicon as shown in Figure 2.14e. In this type of connection, the engagement of the abutment with the implant is provided by an impact force acting along the longitudinal axis of the abutment. Similar to the tapered connections integrated with screw, in this configuration tapered surface of the abutment creates a large frictional resistance area, and interference fit provides the necessary forces for frictional retention. Very low prosthetic complication rates (0.74%) were reported for TIF connections as compared to TIS connections (3.6-5.3%) and external hexagonal-type connections (40%) [118]. In tapered interference fit abutments, loosening problem is of less significance. The biting forces act in the direction of abutment insertion; thus, contribute to secure the connection in contrast to the implants involving only screw connection where the biting force lowers the pretension in the screw. It is loosening torque or pull-out force that could cause the tapered interference fit to become loose [119].

2.3 The Concept of an Elastic Interface for Dental Implants

The anchorage of dental implants in the jawbone is different from that of natural teeth. The periodontal ligament, existing on the root-bone interface of a natural tooth, provides the tooth to adapt itself to various loading conditions by enabling a range of elastic deformation due to its resiliency [65]. In the dental implant case, on the other hand, implant is surgically positioned in the bone and it maintains a direct contact with bone as an artificial root without an elastic layer at the interface [27]. As a result, natural teeth and dental implants differ in force transmission characteristics and mobility under applied loads.

The periodontal ligament around natural teeth keeps the strain gradients in bone within physiologic limits, by uniformly distributing tractions and damping the occlusal forces, which reduces peak stresses, prevents bone resorption. Hekimoglu et al. compared the strains induced around a natural tooth and an opposing dental implant by conducting in vitro experimental study. It was found that the magnitude of functional strain was lower in the bone around natural tooth than opposing implant. Moreover, the natural tooth experienced a slight intrusion, whereas the same amount of vertical or lateral displacement was not observed for the dental implant. Lower strain and higher mobility in the natural tooth case were attributed to the physioelasticity and cushioning effect of the PDL, which was lacked in the implant system [120].

2.3.1 General Approaches

It is not a new concept to incorporate, somehow, flexibility to dental implant designs to compensate the absence of the PDL and mimic its mechanical behavior. Several researchers have made investigations on this issue and various design ideas have been proposed. Among these are four main approaches.

1. Instead of a rigid implant material, using materials with lower stiffness like polymers: Meijer et al. asserted that flexible implant materials may transfer tractions to the surrounding bone more favorably and supported the hypothesis with experimental findings which indicated that bone around a flexible polyactive implant experienced less decrease in density than a rigid implant [121]. However, the insufficient mechanical properties and biological compatibility have limited application of such materials for dental implants [34].

2. Inserting the dental implant directly into the cavity of lost tooth in bone by retaining the former PDL tissue or generation of new PDL tissue on dental implants: Some researchers achieved to form PDL around dental implants [122]; but also added that implementation of forming new PDL had not been suitable for a potential clinical application yet since the degree of PDL generation remained very limited due to the competition between osseointegration and PDL generation [123]. Besides, it was underscored that maintenance of original periodontal tissue most likely prevents osseointegration of implants, adversely affecting the survival rate [124].

3. Coating the external surface of implant contacting bone with a soft material: Choi et al. introduced a new concept of coating an implant's surface with a natural polymer membrane, namely chitosan, in order to provide the viscoelastic characteristic of the PDL to implant. By FE analysis, it was obtained that such a coating could reproduce the PDL's function to absorb the shock of impact loads reducing the stress convergence generated in the surrounding bone [125].

4. Using an internal component made of a material with elastic mechanical properties in the implant-abutment system: This is the sole major approach that has been implemented in the design of a commercial dental implant that has undergone clinical application and has been used in practice. The IMZ (Intra Mobil Zylinder) Implant System, involving a protective, force-dampening, shock absorbing intermobile element made of polyoxymethylene (POM) and designed to simulate the PDL, was developed in Germany by Dr. Axel Kirsch in the early 1970's [126]. The intermobile elastic element was aimed to absorb impact loads and thus preserve the surrounding bone.

2.3.2 The IMZ Implant System

Two different abutment designs involving elastic components made of a resin of polyoxymethylene were used as given in Figure 2.15. The success of the IMZ implant system have been investigated via several studies in terms of both survival rate of the implant osseointegrated with bone and complications related with the abutment part. Meijer et al. compared the survival rate of the IMZ implant and Bränemark implant after 5- and 10-year periods of clinical trial of implant-retained overdentures. Both the 5-year and 10-year survival rates were reported to be 93% and 86% for the IMZ and Bränemark implants, respectively [127, 128]. At the end of 10-year period, the survival rate did not change, because all of the implant losses occurred in the first five years. Moreover, Meijer et al. conducted another comparative study on three different implant systems (IMZ, Bränemark and TMI implants) for a 6-year follow-up period and reported the implant survival rates as 97.5%, 97.1% and 72.0% for the IMZ, Bränemark and TMI implants, respectively [129]. Kirsch and Ackermann also reported the overall rate of success of the IMZ implants, placed and routinely followed over a 10-year period, to be 97.8% [126].

Figure 2.15 Abutments designs of IMZ implant system: (a) IME(Intramobile Element), (b) IMC (Intramobile Connector) abutments [131]

Although IMZ implants were quite popular in mid 1980s, the clinicians' interest on the product decreased in the following years in spite of the satisfactory survival rates of the implants in bone [130]. The reason why IMZ implants lost popularity was complications arising at the abutment part of the implants. Behr et al. conducted a study to compare the complication rate of two implant systems with two different abutments involving rigid and resilient structures (ITI and IMZ implants). The complications were divided into two groups. In the first group, where problems related with the overdenture such as failure of framework, bar devices or other retentive elements were included, the IMZ implants appeared to be more successful. This was possibly due to stress absorption characteristics of the IMZ abutments. However, in the second group considering the complications with implant components such as internal screw loosening due to loss of preload, fracture of screw and/or abutment components, the IMZ components leaded to a considerably higher complication rate (71%) compared to ITI components (13.5%). This was attributed to the inadequate mechanical properties of elastic intramobile elements (IME) and connectors (IMC) in the IMZ system [132]. As another adverse consequence, IMZ system was reported to cause soft tissue complications. This is because the elastic part contacts the gingiva and due to its mobility it might irritate this soft tissue during loading [126]. Furthermore, being exposed to the oral environment, moisture contamination of the elastic part is inevitable, requiring maintenance at certain time intervals [133].

The mechanical response of the IMZ implant system in terms of displacement and stress-strain characteristics has also been investigated by several researchers. Haganman et al. carried out a study to compare the stress distribution in the components of the implant system such as intramobile elastic element and the retaining screw for three different abutment designs of IMZ implants. As illustrated in Figure 2.16, the IME and IMC abutment configurations involved elastic components, whereas the ABC abutment model was totally rigid. Employing finite element method, three-dimensional models for each were generated and axisymmetric analysis was carried out. Axial and oblique loads (at 45º) with magnitudes of 100 N and 500 N were applied on each model separately [134]. The results showed that deflections and stress concentrations within the IMC and IME abutments were in the same range, but much greater than the ABC abutment. The deflection of the crown ranged from 0.0016 mm (for ABC model under 100 N vertical load) to 0.8016 mm (for IME model under 500 N oblique load). On the other hand, maximum effective stresses in the retaining screw ranged from 129 MPa (for ABC model under 100 N vertical load) to 1315 MPa (for IMC model under 500 N oblique load). Oblique load of 500 N caused stresses exceeding the yield strength of screw material(titanium) for the IME and IMC models, but the limit was never reached with the ABC abutment. These stress concentrations occurred at the head of the screw and near the initial screw threads, which were stated as potential sites of failure. Considering the stresses in the elastic components, axial loading slightly incremented the stress caused by the initial preload, whereas oblique loading caused a dramatic increase in the peak von Mises stresses. As a matter of fact, clinical experiences has indicated that it was not the fastening screw, but the elastic part in IMZ abutments, which often failed. Thus the use of such abutments required high maintenance [133].

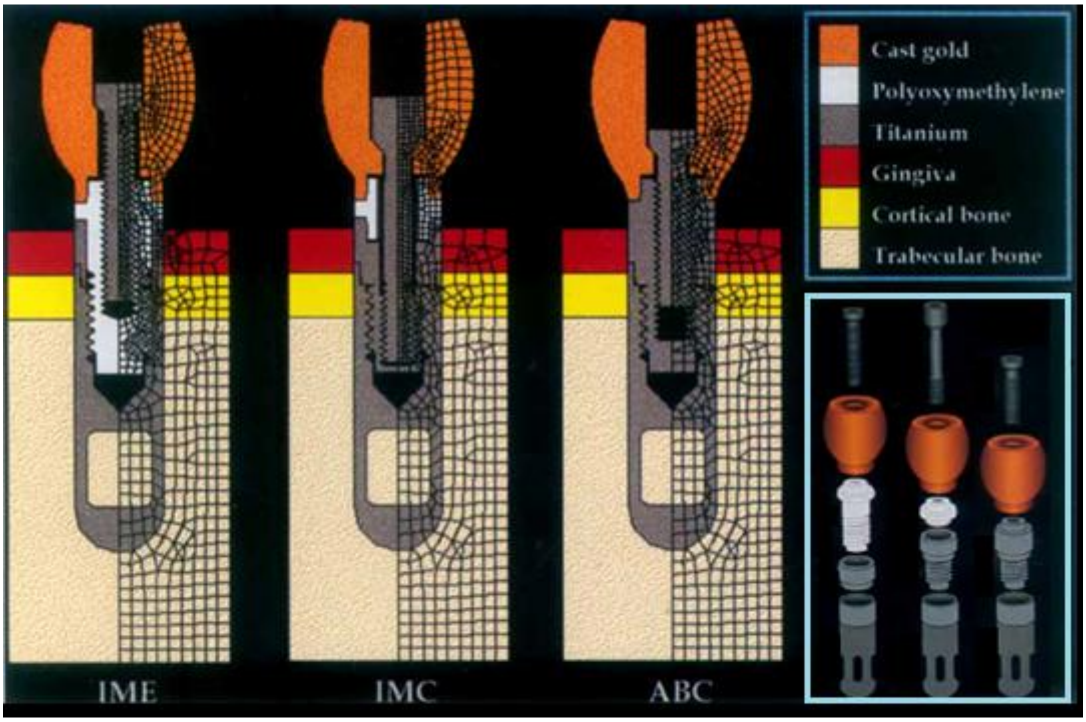


Figure 2.16 FEM analysis of three different abutment designs (IME, IMC and ABC abutments) for IMZ implant system [133]

Oka et al. measured mobility of teeth supported by IMZ implants. The results of the study demonstrated that IMZ implants provided greater mobility than that of a natural tooth in horizontal direction, whereas in axial direction the values were smaller compared with natural case. The IMZ abutments were concluded to function as a stress breaker to horizontal loads and a buffer for the osseointegration of the implants [135]. Akpınar et al. investigated the stresses formed around the implant and the antagonist tooth under occlusal bite force in order to compare a rigid and resilient abutment for IMZ implant in such a configuration by using FE analysis (Figure 2.17). The results of the study demonstrated that a bite force of 143 N parallel to the long axis of the implant resulted in higher compressive stresses around the roots of the tooth opposing the restoration with rigid type abutment. Besides, it was stated that concentration of these high compressive stresses might contribute to intrusion of the tooth. Hence, it was suggested to prefer an implant with a resilient abutment. It was also speculated that in the resilient implant, stresses were concentrated in the intra-mobile element and changeability of this element could make it more advantageous [136]. However, currently, frequent requirement for replacement and maintenance is highly undesirable in clinical dental applications.

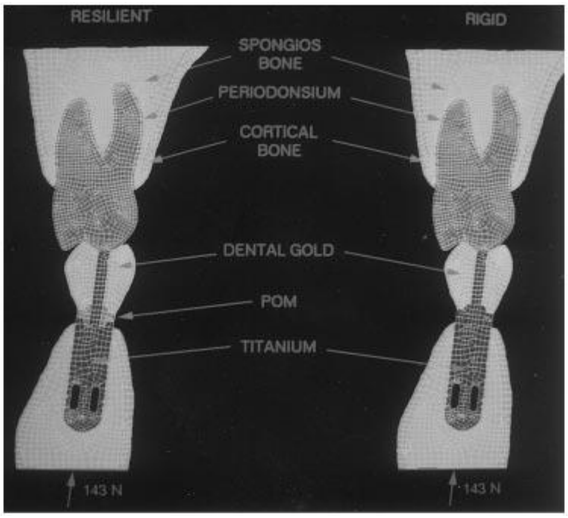


Figure 2.17 Comparison of resilient and rigid dental implants opposing a natural tooth [136]

Dental implants, even if rigid, occluding against natural teeth take the advantage of shock absorbing effect of the opposing natural tooth/teeth. However, implants occluding against implants have a very low shock absorbing effect which is determined only by the deformation of the surrounding alveolar bone. As a result, occluding implants may experience higher strains than those around a natural tooth opposing an implant if the implants are completely rigid [120].

2.3.3 Other Dental Implant Designs Proposed to Simulate PDL

Many researchers tried to come up with the problems arising during the clinical use of the IMZ implant system. Several designs of dental implants involving elastic internal components have been proposed to simulate the PDL.

A patented design of a dental implant with shock absorbent cushioned interface, shown in Figure 2.18a, was claimed to maintain the resilience and cushioning properties of a living tooth. It was speculated that the presented design would protect the jawbone from fracture by distributing and blunting impact forces, increase longevity of the implant system, prevent the penetration of bacteria into the implant, eliminate the need for replacement later in life and provide the patients with increased comfort and the feel of a live tooth [137]. However, when examined in detail, the system is obviously too complex in terms of manufacturability. It seems to be just a conceptual design not proper for practical implementation.

Gaggl and Schultes introduced a new design including maintenance free shock absorbing elements, as given in Figure 2.18b. The aim of using these elements was to decrease the magnitude of induced stresses in case of impact loads and prevent micro-fractures of overloaded bone. Employing a mechanical testing unit, the mobility of the implant under static loading and the long-term stability of the system under cyclic loading in both axial and horizontal directions were investigated. A progressive shock absorption was registered. In the axial direction movements were measured in the range from 0.005 to 0.04 mm with forces from 1 to 90 N. Maximum axial movement was 0.06 mm with a force of 1600 N. In the horizontal direction, maximum movement of 0.16 mm was already achieved with a force of 30 N. In this design, the shock absorbing unit is completely closed within the implant by using a semi-elastic titanium ring connected to the implant by laser welding, which prevents soft tissue problems faced in some other designs arising from the contact of elastic parts to gingiva. After cyclic loading no signs of fractures were observed in the critical welding region in the neck [138]. However, no information was supplied about material damage of elastic components, leaving it uncertain. Although the authors suggested this design to be well suited for superstructure treatment with combined teeth and implant loading, the complexity of the design makes it impracticable for mass production.

Daas et al. evaluated the influence of the retention mechanism on the behavior of an implant-retained overdenture by comparing rigid and resilient attachment configurations (Figure 2.18c) using FE analysis. It was concluded that resilient attachments reduced the load transmission through implants and stresses in the bone surrounding the implant during mastication process. Moreover, rocking motions of overdenture in rigid case was replaced with moderate tilting motions accompanied by a vertical translation in the resilient case. Finally, the resilient attachment allowed for a better load distribution between the dental implants and the denture bearing surface [139]. The limitation of this design is that it is only applicable for supporting removable complete dentures in complete tooth loss cases.

Genna et al. proposed a non-standard design for a fixed dental implant, incorporating a soft layer to simulate the presence of the PDL as shown in Figure 2.18d. Employing the continuum approach, the internal layer was modeled using the compressible hyperelastic model of Storakers with isotropic assumption. FE analysis results indicated a profound redistribution effect on the stress field in the surrounding bone upon loading. In addition, incorporation of such an internal layer into the implant provided the implant-supported prosthesis with a very similar mobility as that of a natural tooth. The authors underscored that a certain existing material was not specified for the internal layer, instead a material whose mechanical behavior is analogous to that experimentally obtained for the PDL was assumed to be available [140]. On the other hand, the design proposed in this study is just conceptual and not proper for practical application since the mechanical integrity of the system can not be maintained in case a force having a tensile component is applied. Moreover, from the biological perspective, the internal layer is in contact with gingiva and exposed to intraoral agents, posing the risk of gingiva inflammation and bacterial penetration.

Mensor et al. evaluated the damping ability of the Compliant Keeper abutment which consisted of silicone O-rings (Figure 2.18e). Cyclic loading and compression tests on the system demonstrated that the Compliant Keeper system might mechanically replicate the measured damping function of a natural tooth when natural teeth and implant bodies are connected to support an overdenture. The O-rings in the abutment allowed the system to serve as a selectively controlled universal joint that functioned like an analog of the periodontal ligament, providing damping and progressive loading for implant-supported prostheses. One major limitation is that it can not be used for freestanding implants, but only for supporting overdentures [141]. Further, the connecting screw of the abutment may probably loosen under cyclic loading.

Unlike the aforementioned designs, Achour et al. studied the concept of interposing a bio-elastomer between the abutment and the crown (Figure 2.18f) in order to damp the occlusive shocks and to attenuate the stress concentrated at the implant-bone interface. Performing an FE analysis of two different geometries, conventional model and new model with elastomer, it was concluded that both geometries presented quite similar qualitative stress distribution; lower stresses occured in the implant sytem and at the interface with bone for the one involving elastomer due to the stress shielding effect of the elastomeric stress barrier; in both geometries stress concentration occurred at one side of the neck; the sytem involving elastomer was capable to diminish or to delay the loads transmitted to implants and to the bone [142]. Carvalho et al. obtained very similar results to that of Achour et al., by redesigning the conventional Bränemark dental implant system and incorporating a bio-inert stress barrier (elastomer) between the implant and the ceramic crown [143]. The numerical model used within both studies suffered from limitations related to elastomer characterization. Besides, for a free standing implant such a design seems to be unreliable in terms of mechanical integrity.

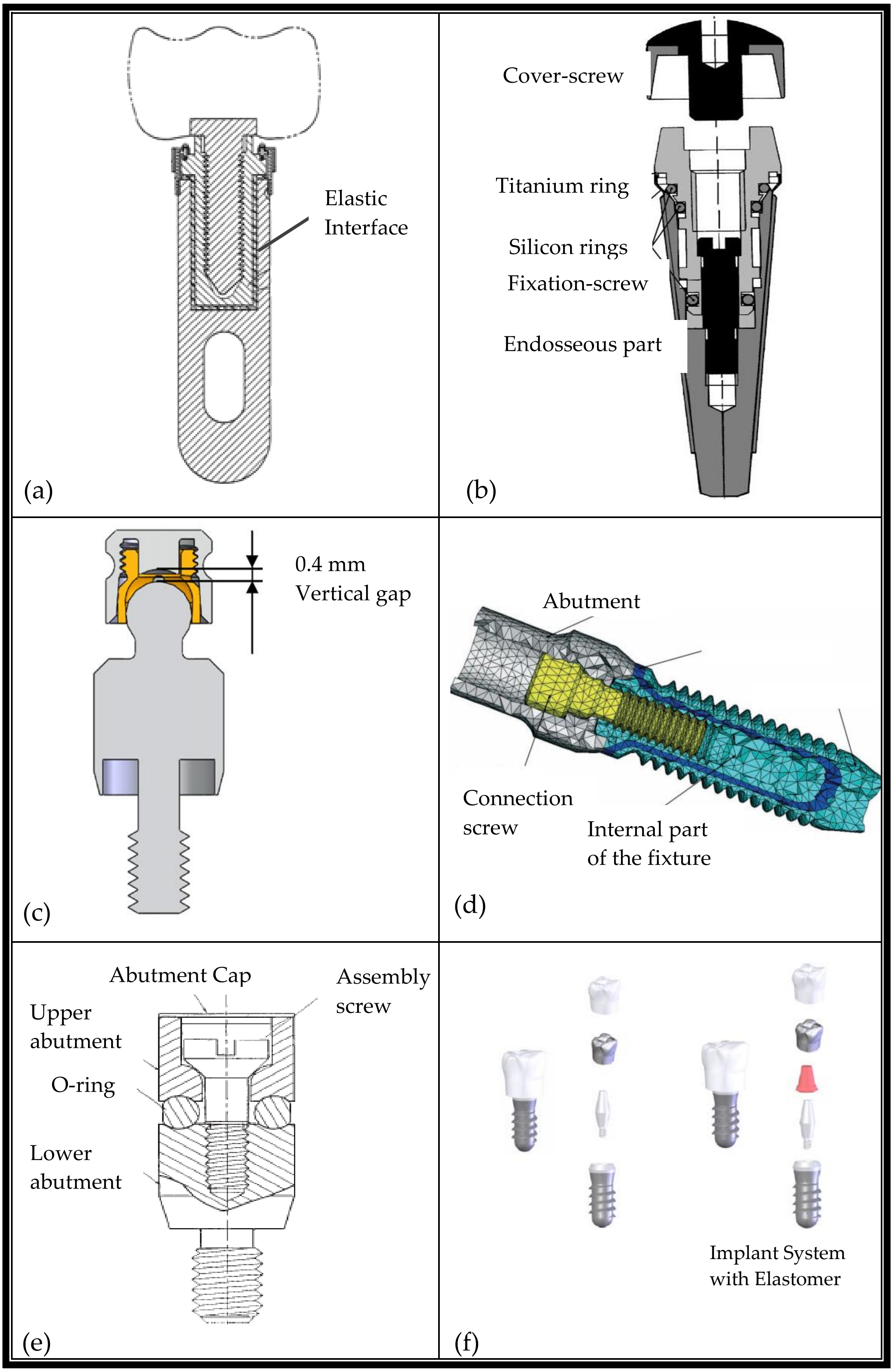


Figure 2.18 Different designs of dental implants involving elastic interface to simulate mechanical behavior of periodontal ligament

2.3.4 Other Important Considerations on Elastic Interface Concept

The use of a resilient dental implant instead of a rigid one has been proven to lead to superior biomechanical performance considering immediate loading of dental implants and construction of implant-supported overdentures.

2.3.4.1 Immediate Loading of Dental Implants

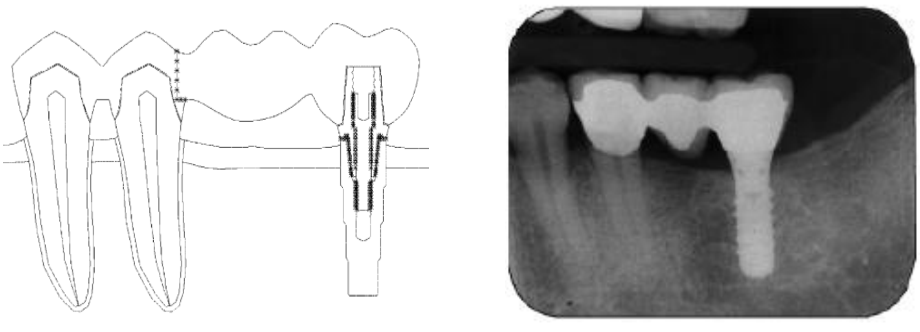
In general, clinical dental implant treatments are carried out using two different surgical techniques, two-stage and one-stage surgery. The time interval between implant placement and prosthetic loading underlies the distinction between these methods [144]. In the former technique, a waiting period of 3-6 months follows the surgical placement of the implant into the bone in order to provide the implant with sufficient time to firmly osseointegrate with the surrounding bone. After this period, the artificial crown that will be subjected to functional loads is placed on to the implant system [145]. The latter technique, on the other hand, referred as immediate loading, involves no waiting period before prosthesis restoration on the implant with full functional occlusal loading of the implant within 2 weeks of placement of the implant [146]. The timeframe of the restorative phase of the implant surgery have been debatable until a consensus was reached in 2006, in the International Congress of Oral Implantologists, on the definition that immediate loading is a technique in which the implant supported restoration is placed into functional occlusal loading within 48 hours of implant insertion [145, 147].

Although, the two-stage surgical technique has been mostly used due to its documented success rates since 1960 [145], the concept of immediate loading has recently become popular due to less trauma, reduced overall treatment time, decreased patient's anxiety and discomfort, high patient acceptance and better function and esthetics [148]. Nevertheless, immediate loading of a just placed implant may result in failure of the implant osseointegration [149], which is related with several factors such as nature of loading, initial stability, bone quality and quantity and design of the implant and abutment. Among these factors, the establishment of primary stability is the most important variable concerning the success of immediately loaded implants [145], since any relative motion between implant and bone can produce a fibrous tissue at the interface and cause abrasion of the bone, leading to crestal bone loss, progressive loosening of the implant and finally failure of osseointegration [65, 145]. It is reported that relative micromotions should be maintained in the range of 50 to 150 μm not to cause failure of osseointegration according to immediate loading protocol [149, 150].

The use of a resilient implant instead of a rigid one may help avert relative micromotions that are out of admissible range. The mechanical behavior of rigid and resilient dental implants with respect to immediate loading was compared via FE analysis by del Palomar et al. The micromotions of the resilient implant were lower than those of the rigid one, while keeping the stresses in the implant under allowable maximum values. It was concluded that the resilient implant consisting of a silicone gasket was efficient in reducing the micromotion between bone and implant under early load, keeping it inside the bounds of the immediate load protocol [150].

2.3.4.2 Construction of Implant-Tooth Co-Supported Dentures

It is a common clinical application to construct a fixed partial or complete denture that is supported by implant(s) together with natural tooth/teeth as can be seen in Figure 2.19. In such a combined restoration, the presence of the periodontal ligament around natural tooth roots provides a certain dental mobility, whereas implant has a rigid connection with bone [151]. The resulting differential mobility throughout the prosthesis causes uneven load sharing between the supports of the system. A larger portion of the masticatory load is transferred to the most rigid supporting member that is the implant in a combined restoration. In order to relieve stress to prevent implant overload, nonrigid internal attachments are recommended to be included in the system [152].



(a) **(b)**

Figure 2.19 Combined implant and tooth support for dentures: (a) 2-D model [153], (b) Radiograph of a combined prosthesis [154]

Menicucci et al. investigated the stresses at implant-bone interface that occurred during loading of a tooth which was rigidly connected to a distally placed implant as illustrated in Figure 2.19a. FE analysis was carried out by applying vertical static load (for 10 s) and impact load (for 5 ms) separately. According to the results, load duration had a greater influence than load intensity on the stress distribution. Under static loading, higher stresses were created in the bone around the implant than the bone around the tooth. This was attributed to the progressive deformation of the periodontal ligament, causing the bridge to act as a cantilever on the implant. As a result, stress concentration occurred in the bone around the implant neck [151].

In order to distribute stresses evenly and eliminate the risk of overloading of the implants in fixed dentures combining natural teeth and implants, the mobility of the implant must be similar to that of the teeth. Nevertheless, teeth and implants display different patterns of mobility under physiological loads. A natural tooth exhibits a two-stage displacement behavior. The first stage, being rapid within the confines of the PDL, is followed by a linear second stage related to the deformation of bone. The mobility of implants depends solely on bone deformation and display a linear behavior proportional to the applied load [155].

Boldt et al. measured the axial and lateral mobility of natural teeth and implant-supported teeth under physiological loading by conducting an in vivo experimental study on a series of volunteers using an optical system attached to the teeth/implants to be measured and a light source attached to a reference point. Axial displacement of teeth displayed a time dependent viscoelastic behavior due to damping effect, which was insubstantial for lateral displacement. Whereas increasing the rise time raises the order of displacements, for very short rise times, axial displacement could disappear almost completely. On the other hand, for dental implants, without measurable influence of the load rise time, elastic deflection was observed in both axial and lateral directions. From Figure 2.20a it can be observed that for a maximum axial loading of 80 N, the axial displacements of natural teeth reached to 100-140 μm, whereas that of dental implants remained in the range of 60 to 80 μm.

According to Figure 2.20b, a lateral load of only 30 N created a displacement about 160 μm for natural teeth, whereas same load caused a displacement of just 30 μm for dental implants [156].

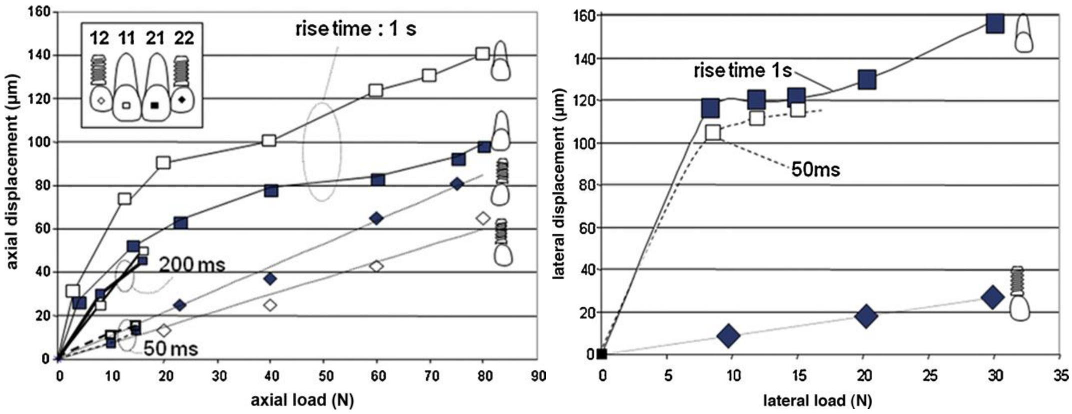


Figure 2.20 Experimentally obtained force versus displacement curves for (a) axial and (b) lateral loading [156]

As a remedy for compensating the dissimilar mobility of natural teeth and dental implants, it has been recommended to employ stress absorbing elements in the dental implant systems, especially when utilized in combination with natural teeth in a fixed prosthesis. Rossen et al. used FE analysis to examine the stress distribution in bone around implants with or without elastic stress absorbing elements for a free standing implant and an implant connected with a natural tooth. For the free standing implant, the elastic component acted as a damping element under dynamic load, but not as a stress distributer under static loading. The change in the shape or elastic modulus of the element slightly manipulated the stress distribution in the bone, the total force remaining constant with static loading. For an implant connected to a natural tooth, however, elastic component provided more uniform distribution of load in the bone around the implant, decreased the height of the peak stresses in the bone around the natural tooth. More uniform stress around the implant was explained by the fact that displacement of the bridge was more translational than rotational, as opposed to the case with a rigid implant system. In short, for an implant connected with a natural tooth, the elastic element was effective both as a stress distributer and stress absorber [157].

It can be concluded that if the mobility of the implant can be achieved to display the same behavior as that of the natural tooth connected to it by a proper mechanical design of a dental implant system including resilient element(s) and appropriate material selection, the best mechanical performance would be attained from the dental prosthesis that is co-supported by a dental implant and a natural tooth.

CHAPTER 3

MATERIALS AND METHODS

3.1 Introduction

Dental implant, the most functional and efficient tool as a choice for treatment of tooth losses, has been a subject of intensive research with regard to both mechanical properties and biocompatibility characteristics. Being beyond the scope of this study, the biocompatibility of dental implants is related with the material selection and specifications of the surface that contacts with jawbone. The mechanical properties of a dental implant, on the other hand, directly depend on the mechanical design of the system.

As explained comprehensively in the previous chapter, all of the dental implants, those are presently in practical implementation, are composed of rigid components constituting a totally rigid structure. On the other hand, the concept of incorporating somehow elasticity to a dental implant system has been verified to provide crucial benefits in terms of mechanical response by several experimental and theoretical studies. Although there have been many attempts to come up with a functional design of an elastic dental implant, each of them lacked some important requirements for dental implants to be used clinically as a treatment for tooth loss.

In this chapter, a novel design will be proposed for an elastic dental implant that would mimic natural tooth with a periodontal ligament. The procedure followed during the design of the dental implant system and optimization of the design by using finite element method is explained in detail.

3.2 Design of the Dental Implant System

The design objective is to achieve an optimized, workable dental implant model that is capable of withstanding biomechanical loads within the constraint of using biocompatible material, surface and geometric properties in order maintain long-term survival rates. The critical contribution will be the simulation of the mechanical behavior of the periodontal ligament of a natural tooth. In accordance with this objective, the optimization of several parameters in a combined manner is required to be made. Being very critical, the subject of surface texture is not dealt within the context of this study. The geometric design, in this section, and material modeling, in the finite element modeling section, are handled in detail in this study.

A generic dental implant system consists of two major components, namely, the implant and the abutment. Implant is the basic part that is surgically placed and remains inside the jawbone. The abutment is the intermediate part between the implant and artificial crown. It is fastened to the hollow interior of the implant and usually bonded to the crown at the top. In addition to the design parameters of the implant and abutment, the mechanical design of the interface between the implant and abutment is of the essence with respect to the mechanical integrity of the system, mechanical response within the system under loads, stability and strength of the system and long-term survival of the dental implant treatment.

3.2.1 Design of the Implant

The important design criteria about the implant geometry are about the shape, size, thread profile and neck region of the implant. The shape of the implant significantly affects the stress distribution at the interface between bone and implant, but not the maximum stress values. A ***cylindrical geometry*** is preferred (Figure 3.1), since it produces a more uniform stress profile with less localized stress concentrations in bone when compared with conical type implants [87]. For the regions of low bone quality, it can be switched to conical type, since conical external geometry results in increased primary stability by compressing the soft bone during and after implantation [85, 86]. On the other hand, instead of a straight or stepped surface type, a ***threaded screw profile*** is used, since it has been verified to reduce stress levels in trabecular bone [83].

The size of the implant is another important factor that influences the stress distribution in the surrounding jawbone. Although the selection of size, specified by length (in the range of 6.0-16.0 mm) and diameter (in the range of 3.0-6.5 mm), depends on the patient’s bone state and the location of application on the jawbone, the optimum size values have been reported. For the diameter, remaining below the upper limit of 4.5 mm, survival rate of the implant has been reported to increase as diameter increases. Moreover, implant length exceeding 9.0 mm has been proven to reduce bone stress and enhance implant stability [93, 94, 95]. Therefore, for the models used in this study a combination of ***4.2 mm external diameter and 9.0 mm length*** is determined to be used.

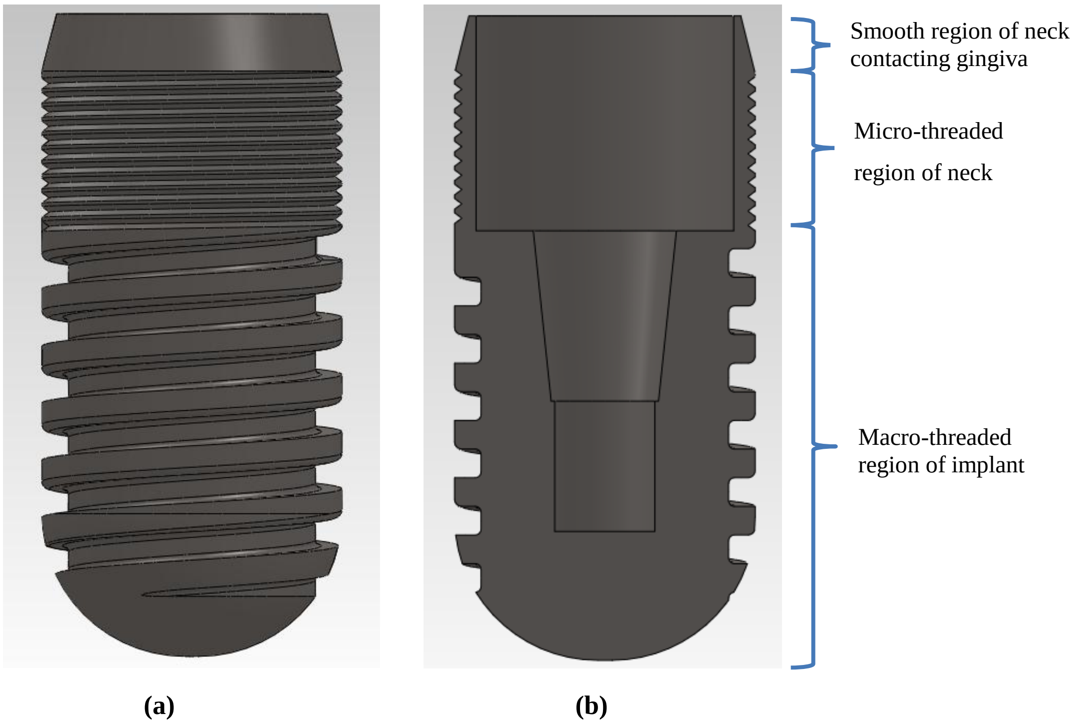


Figure 3.1 The design of the implant geometry: (a) Implant external geometry, (b) Sectional view of the geometry

Enlarging the area of contact surface of implant is another important design criterion, since maximization of the contact area with bone improves initial stability, distributes interfacial stresses more evenly and preserves bone from excessive local stresses [80, 97]. In addition to increasing the size of the implant, choosing an appropriate thread profile on the external surface of the implant is critical for increasing the contact area. Among the alternative thread profiles like square shape, V-shape, buttress and reverse buttress threads, the square shape profile has been verified to provide the highest contact area and lowest maximum interfacial stresses, which promotes osseointegration and initial stability [99].

Square threads have about 10 times less shear component of axial loads reducing the risk of overload, compared to other thread types [158]. Therefore, for the model proposed in this study, a ***square-type thread*** is employed (Figure 3.2). The upper and lower edges at root radius and lower edge at outer radius of the threads (shown in Figure 3.2) are filleted with a small radius to avoid stress concentrations that might possibly induce micro-cracks under high loads, by more evenly distributing the stress [100]. Moreover, it has been reported that the thread pitch(p) values between 0.8-0.9 mm is optimal for cylindrical implants [97], with best stress distribution when width(W) and height(H) of the thread are $p/2$ and $0.46p$, respectively [100]. Hence, for the present model the values for ***pitch, width and height of the thread*** are selected as ***0.8 mm, 0.4 mm and 0.368 mm***, respectively.

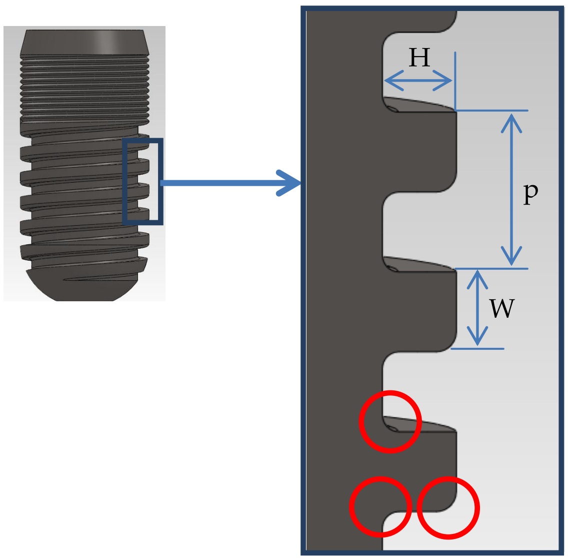


Figure 3.2 The thread profile of the proposed implant geometry

Under masticatory loads and bite forces, the highest stresses in bone occur and bone resorption initiates in the region surrounding the neck portion of the implant [80]. The cortical bone, the thin hard layer of bone surrounding the softer trabecular bone, is in contact with the neck region of the implant. In order to prevent bone loss and maintain the cortical bone, circumferential grooves in the form of micro-threads, as shown in Figure 3.1b are incorporated at the neck region along a length of 2.2 mm, which is the average cortical bone thickness [59], since applying micro-threads is reported to provide positive stress stimulus to bone for remodeling and to decrease bone loss [80, 103]. Finally, above the upper level of micro-threads a smooth region of neck is left, since this portion of the neck, called transgingival region, will contact to the gingiva which is a soft tissue.

The details of the design of the interior geometry of the implant are left to be explained in Section 3.2.3 (Design of the Implant-Abutment Interface).

3.2.2 Design of the Abutment

Abutment is the part of a dental implant system which holds the artificial crown bonded onto the surface of it. It has the function of transmitting the occlusal forces to the implant. The most critical part in a dental implant system is the abutment or the internal screw which is used to fasten the abutment to the implant. Except from some of the abutments that could be used only for removable complete dentures (like the one in Figure 2.18c), all of the present commercial products involve totally rigid abutment designs. In Figure 3.4, a rigid type dental implant abutment is illustrated. This design involves a one-piece conical abutment. Besides, there also exist rigid type dental implant systems including internal fastening screws connecting the abutment to the implant (Figure 2.14a, Figure 2.14c and Figure 2.14d).

Although the significant advantages of involving elasticity in a dental implant system have been verified both theoretically and experimentally, no workable design has been achieved yet. In accordance with the aim of this study, the suggested approach to this problem will be introduced at this step of the design procedure. In order to overcome the problem, the last approach among the four alternatives explained in Section 2.3.1 was utilized. More explicitly stating, in order to mimic the mechanical behavior of the periodontal ligament, the method of using internal components with elastic material properties within the upper structure (abutment) of the system is preferred due to the reasons explained in the literature review section.

Prior to taking a step to the conceptual design phase, the problems with the previous designs, listed in Section 1.6 were scrutinized and determined as design criteria to be fulfilled. After that, different concepts for abutments involving elastic intermediate elements were generated and rough assessment was carried out according to the design criteria. The notable concepts among several alternatives generated at this step are presented in Figure 3.3, in an arrangement that indicates sequential progress in terms of improvement predicating on the design criteria.

The model shown in Figure 3.3a was the starting point to concept generation. In this model, the abutment unit consisted of an abutment housing enclosing two elastic internal elements, closed with a cap at the top and connected to the implant with two separate components. One is the part that guides the elastic parts and the abutment housing, whereas the other is the internal connecting screw. Although the model was a candidate to provide certain mobility to the dental implant system at the upper structure, it required improvement for two reasons. The first was that outer diameter of the abutment exceeded the implant diameter and seemed to be oversized to place a crown on it. The second negative aspect was the unnecessary multiple parts to make connection with the implant, increasing the complexity of the system.

The mentioned problems were eliminated by proceeding to the design illustrated in Figure 3.3b. In this design, the connecting parts were combined into a single bridging part, the upper portion of which served to guide the mobile abutment housing and to support the elastic parts, whereas the lower portion provided a screwed connection into the implant. Yet, the drawback that leaded to further elimination of this concept was the gap existing between bottom of the abutment and top of the implant neck, which was already a problem in the preceding design. This cavity was predicted to pose the risk of being a potential zone for bacteria accumulation.

The next step involved a prominent altering in the design (Figure 3.3c). The gap that existed in the previous designs was eliminated by moving and confining the abutment inside the interior hole at the top of the implant so that the internal surface of the implant would serve as a guide for the movement of the abutment during loading. Another noteworthy improvement in this design underlies the locations of the elastic elements. While in the previous designs the load acting on the system was carried solely by the upper elastic element and completely transmitted through this element to the internal screw, in this design the load would be shared on two elastic parts. Moreover, only the portion of load that acted on the upper elastic element would be transmitted to the internal screw, remaining load being directly transmitted to the implant. This would significantly decrease the stress levels induced on the internal screw and elastic parts, which have been the most critical components in such dental implant systems that are previously proposed as indicated in the literature review. As a result, the load carrying capacity of the whole system was considerably improved. In spite of these favorable consequences of changing the position of lower elastic part, an important function that was included in the previous designs was excluded by this alteration. It was the role of the lower elastic element in the two previous designs to provide a smooth continuous unloading behavior when applied forces are removed and to bear tensile shock forces that may act on implants supporting fixed overdentures.

In order to incorporate that function again, a third elastic component is placed at the prior position of the lower elastic part in the first two designs (Figure 3.3d). Moreover, the lower elastic element was transferred to the gap between abutment and implant in the form of a sealer at the neck. The abutment was still remained to be confined in the internal hole of the implant. This design resolved the last mentioned problem, however, at the same time arising a new downside. The elastic element at the neck would apparently bulge towards the gingiva, causing irritation and inflammation of this soft tissue. This was also a common problem in some of the previously proposed designs available in the literature.

A further improved design, given in Figure 3.3e, tackled with the last complication by narrowing the abutment housing and other components so that abutment could again be confined inside the implant. By this means, the lower elastic element was again located inside the implant below the abutment.

This step was the end of the conceptual design phase, since all the rough design considerations were seemed to be achieved, requiring further optimization by analytical or finite element techniques in order to satisfy specific objectives described in Section 1.7. Actually, the model presented in Figure 3.3f is the final optimized design that was obtained at the end of the study through implementation of a number of analytical procedures and finite element analysis techniques. It is included in this section just for completeness and enabling easier comparative assessment for the reader.

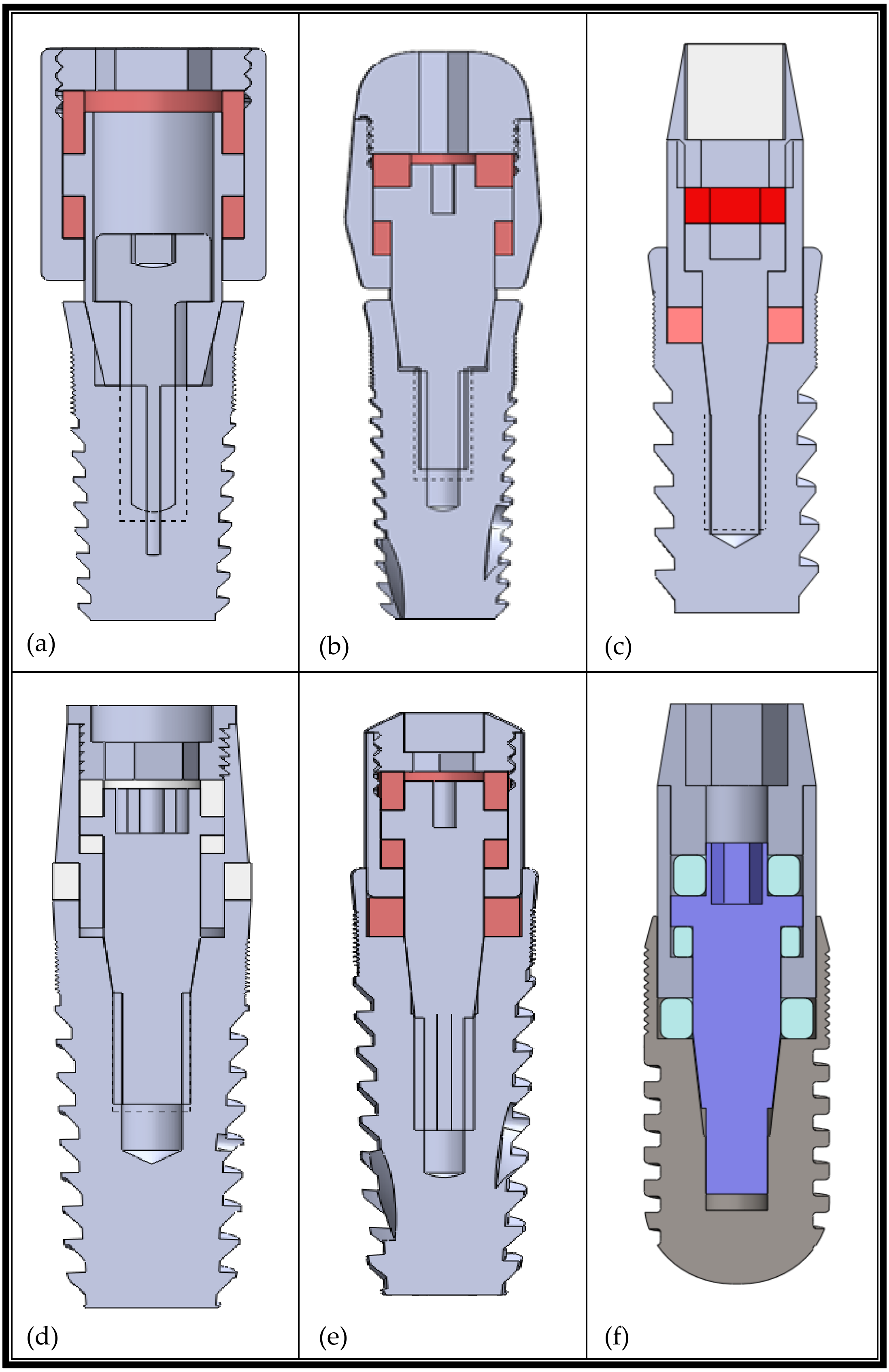


Figure 3.3 The conceptual designs for the elastic abutment

In Figure 3.5, the structure of the optimized elastic dental implant system is demonstrated more definitively and it can be compared with the structure of a rigid system given in Figure 3.4.

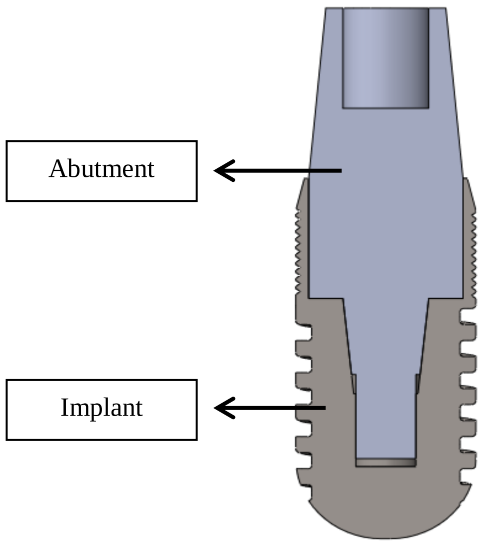


Figure 3.4 The components of a rigid type implant system (sectional view)

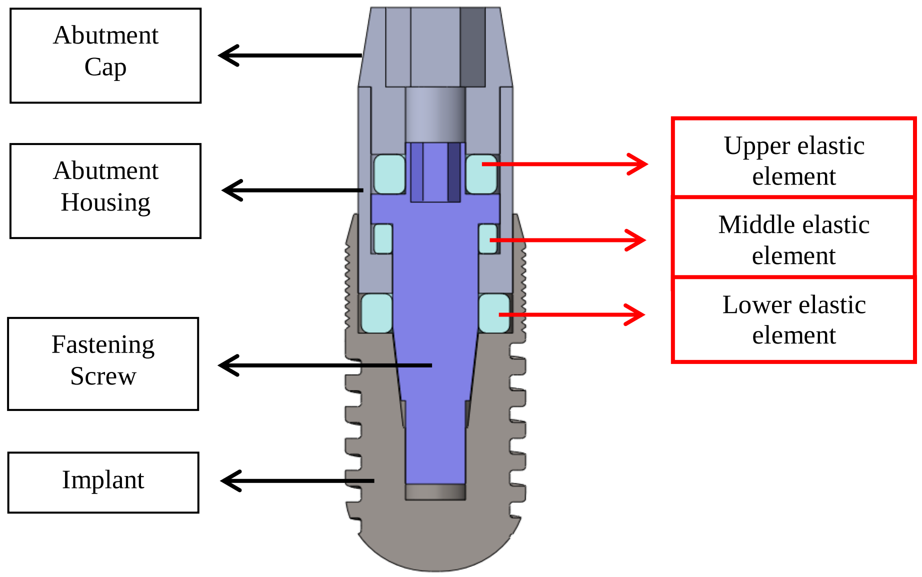


Figure 3.5 The components constituting the proposed resilient dental implant system

3.2.3 Design of the Implant-Abutment Interface

The design of interface between the implant and the abutment is of very high concern in addition to design of these individual components, with respect to the biomechanical properties of the dental implant system. The mechanical response within the system under loads, mechanical integrity, stability, strength, and thereby, long-term survival rate of the whole system depends to a great extent on the nature of this interface.

One of the foregoing complications in implant-supported restorations is loosening of the internal screw that is responsible for sustaining the connection between the implant and abutment. Although it has been stated as a problem confronted in elastic dental implant designs, even the rigid implants carry the risk of screw loosening under functional loading. The resistance to loosening is determined by the preload that is created by applying a tightening torque to the head of the screw in order to clamp the abutment to the implant by the action of compressive and frictional forces at the contacting surfaces. Screw loosening occurs when the initially applied preload becomes absent under loading.

The factors affecting screw loosening are the amount of applied preload, machining tolerance of the mating surfaces, the geometry, material properties and surface characteristics of the components. It is apparent that higher the initial screw preload, greater the resistance to loosening. However, the magnitude of preload is limited by the maximum tightening torque that is allowable to be applied. The tightening torque value has to be lower than the reversal torque of the osseointegrated implant by a safe difference. Otherwise, applying a tightening torque approaching the value of average reversal torque of the implant might harm the biomechanical bond between implant and jawbone and cause loosening of the implant, which will result in failure of the implant treatment. The reported values for implant removal torque and screw tightening torque are in the range of 41.7-139.7 N.cm [159, 160] and 10-35 N.cm [117], respectively. The removal torque for implant indicates the strength of the bond between implant and bone. It may change depending on the state of bone of the patient and the specifications of the implant like size, shape and surface characteristics. The tightening torque of the abutment screw, on the other hand, is specified by the manufacturer and may vary between different products according to design specifications.

As described in Section 2.2.2.2.5, rigid dental implant systems involve two main types of configurations for implant-abutment connection, namely, external and internal connections. The external type connections involve hexagon-mediated butt joints, whereas the internal type connections can be achieved by conical or hexagonal joints with one-piece or two-piece systems. In comparison to other connection types, the internal conical connection, with its solid design, poses a dramatically improved ability to resist bending forces and provides better clinical integrity. In addition, it leads to more uniform force transmission, less stress concentration and lower bone-implant interfacial tractions. However, the most striking superiority of internal conical interface is the enhanced capability to resist against loosening at implant-abutment interface, which has made this type of connection more popular to be implemented in most of the commercial rigid-type dental implant systems.

As distinct from all of the previous attempts to design an elastic dental implant, it is decided to incorporate an internal conical connection combined with a screw into the design owing to the prominent advantages of this type of connection. The detailed design of the implant-abutment interface will be carried out by adapting an analytical approach (the concept of taper integrated screw for abutments) explained in a previous study (conducted by Bozkaya and Müftü [106]) to this design in order to cope with the problem of loosening. In fact, the analytical procedure implemented in that study was utilized to examine and compare the implant-abutment interfaces of certain commercial rigid type implants involving ITI implant (shown in Figure 2.14b). Nevertheless, the same approach is easily applicable to the presently proposed design of elastic dental implant system in this study. The geometric parameters involved in the conical interference connection between implant and abutment is represented in Figure 3.6.

Against loosening, the internal conical connection takes the advantage of friction-locking mechanics at the region of mating conical surfaces more than the preload of the fastening screw. When a tightening torque in the allowable range is applied to the abutment screw, the screw at the bottom advances inside the bore of the implant. This causes the threads on the abutment and implant to engage with each other which creates a tensile force, called preload, in the screw. In addition to the generated preload in the screw, the conical region above the threads also advances inwards of the tapered hole of the implant. This results in interference fit of the conical mating surfaces, which dominates the resistance of the connection against loosening. During tightening, the micro-roughness of the contacting surfaces flattens, decreasing the radial distance between surfaces. This process is defined as settling. At the end of settling process, some of the clamping force both on the thread surfaces and conical surfaces decrease. Hence, the torque required to loosen the abutment is always lower than the initially applied torque to tighten the abutment. Besides, based on the same reason, repeated tightening and loosening of the abutment screw provokes the reduction of the loosening torque values [115].

The geometric parameters, friction coefficient between contacting surfaces and the elastic properties of the materials affect the mechanics of the system. The closed-form analytical relations for screw preload, tightening torque and loosening torque depending on variables associated with these properties can be derived by examining the balance of forces acting on each part using free body diagrams.

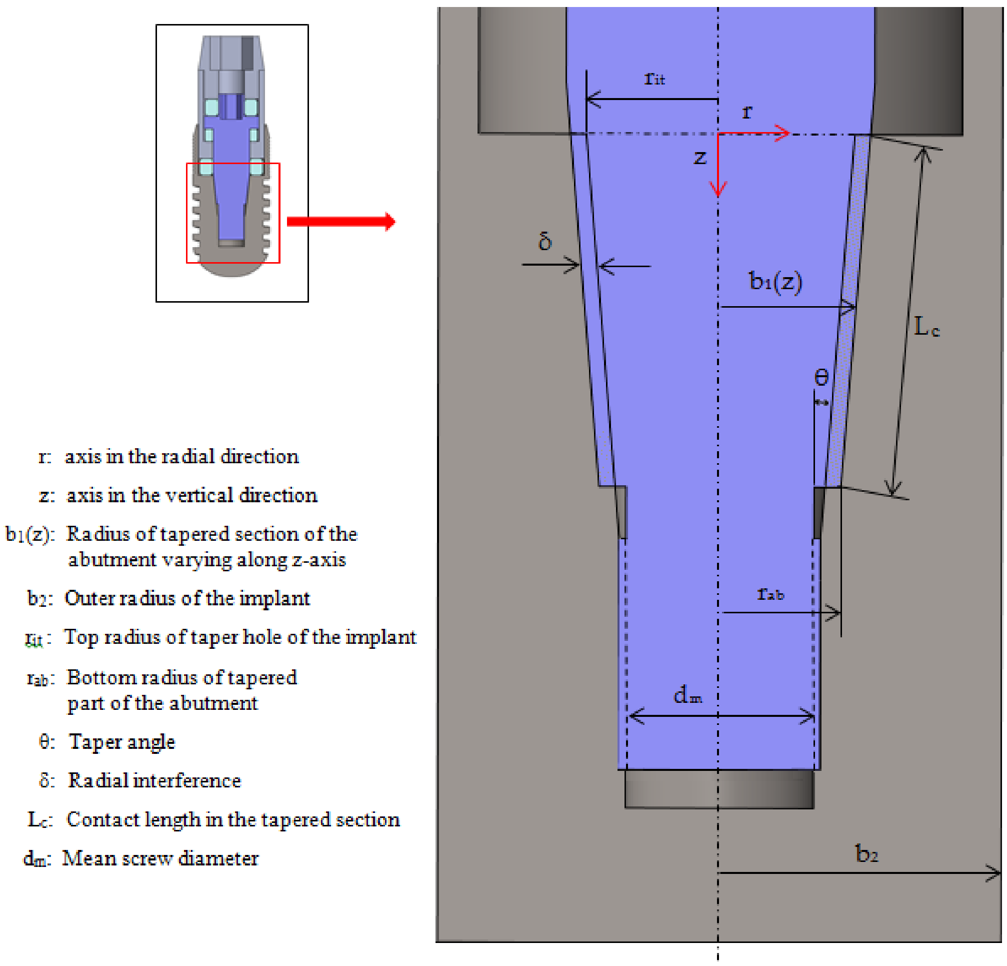


Figure 3.6 The representation of the geometric parameters involved in conical interference connection between implant and abutment

Before proceeding to force and torque equations, some geometrical relations as functions of the parameters shown in Figure 3.6 should be defined. The interference in the radial direction is a function of axial displacement (Δz) and taper angle (θ),

$$\delta = \Delta z \tan\theta \tag{3.1}$$

The contact length (L_c) of at the tapered surfaces can be defined as

$$L_c = \frac{(r_{it} - r_{ab})}{\sin\theta} + \frac{\Delta z}{\cos\theta} \tag{3.2}$$

where r_{it} is the top radius of the tapered hole of the implant and r_{ab} is the bottom radius of the tapered part of the abutment. The radius of the tapered section of the abutment varying along the main axis of the abutment is

$$b_1(z) = r_{ab} + (L_c \cos\theta - z)\tan\theta \tag{3.3}$$

The friction force in the tapered part develops along a helical path which depends on the lead of the screw as demonstrated in Figure 3.7c. Therefore, the helix angle of this path is equal to the lead angle of the screw (λ) which is defined as

$$\lambda = a\tan^{-1}\left(\frac{l}{\pi d_m}\right) \tag{3.4}$$

where a is a constant, d_m is the mean screw diameter

$$d_m = d_{major} - 0.649519\,p \tag{3.5}$$

and d_{major} is the outer diameter of the screw thread, p is the thread pitch and l is the lead of the screw,

$$l = np \tag{3.6}$$

while n indicates whether the screw has a single (n=1) or multiple thread (n= 2, 3, …) [161].

3.2.3.1 Mechanics of Tightening

During tightening the abutment screw into the implant, the tensile preload developed in the screw is counterbalanced by the resisting force which is generated in the tapered region. Figure 3.7a illustrates the equilibrium of forces on the free body diagram of the tapered part of the abutment during tightening.

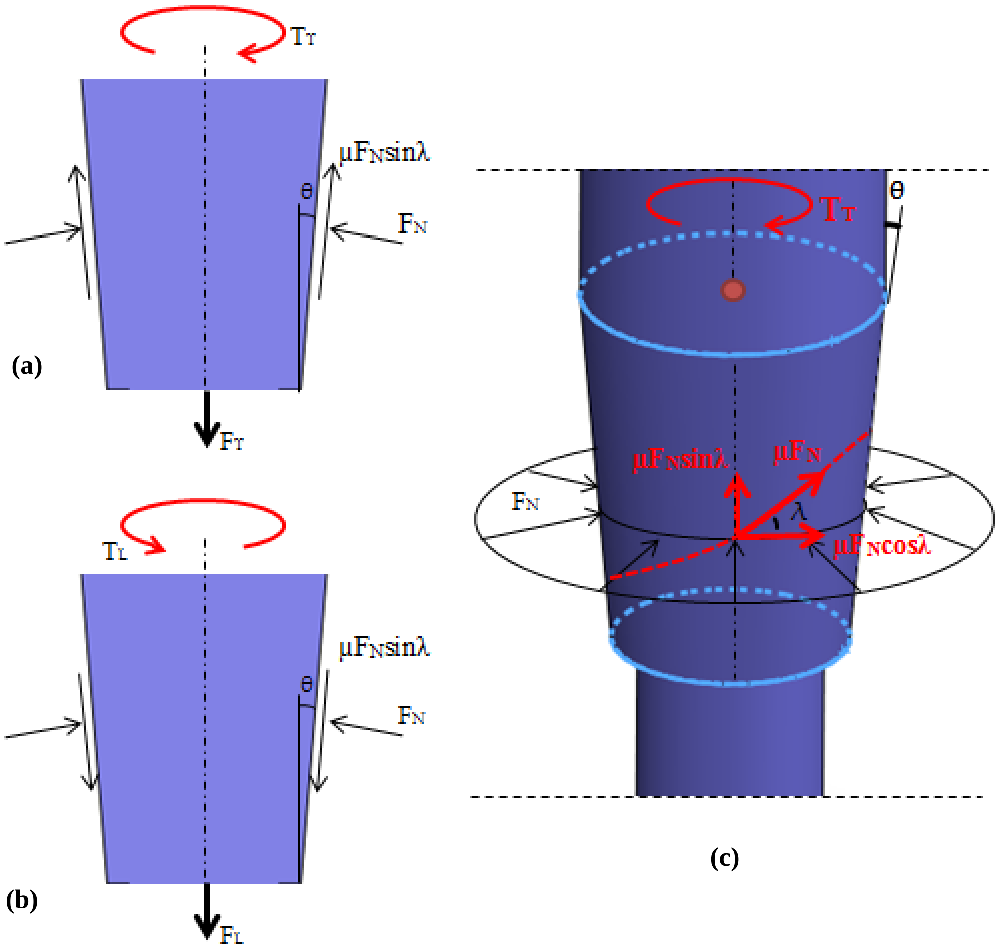


Figure 3.7 The free body diagrams of the conical part of the abutment screw (a) during tightening and (b) during loosening; (c) The representation of frictional force along helical path during tightening

The resisting force, F_T, is the resultant of the components of the frictional and normal forces on the tapered surface acting along the main axis (z-axis) of the abutment. The magnitude of the axial resistive force is equal to the screw preload. It can be calculated from

$$F_T = F_N \sin\theta + \mu_k F_N \sin\lambda \cos\theta \tag{3.7}$$

where μ_k is the kinetic friction coefficient. As seen in Equation 3.7, the resistive force, therefore, screw preload is related to the normal force (F_N) acting on the tapered contact surfaces. The resultant contact force due to interference fit on the tapered surface can be obtained by integrating the contact pressure $P_c(z)$ over the contact area along the contact length [119],

$$F_N = 2\pi \int_0^{L_c \cos\theta} b_1(z) P_c(z) dz \tag{3.8}$$

The contact pressure due to interference fit of the tapered part is a function of axial position,

$$P_c(z) = \frac{E\delta\left(b_2{}^2 - b_1{}^2(z)\right)\cos\theta}{2b_1(z)b_2{}^2} \tag{3.9}$$

where E is the elastic modulus of the implant and abutment screw, b_2 is the outer radius of the implant. Carrying out the integration, the following relation can be obtained for the normal contact force

$$F_N = \frac{\pi E \Delta z L_c \sin 2\theta}{6b_2{}^2}\left[3\left(b_2{}^2 - r_{ab}{}^2\right) - L_c \sin\theta\left(3r_{ab} + L_c \sin\theta\right)\right] \tag{3.10}$$

The *total tightening torque,* T_T*,* for the abutment is the sum of the resistive torques in the tapered part and in the threaded region,

$$T_T = T_{T_{Screw}} + T_{T_{Taper}} \tag{3.11}$$

The resistive torque in the screw threads can be calculated from the power screw formula for raising a load [161]

$$T_{T_{Screw}} = \frac{F_T d_m}{2}\left(\frac{l + \pi\mu_k d_m \sec\alpha}{\pi d_m - \mu_k l \sec\alpha}\right) \tag{3.12}$$

where α is the thread angle of the screw.

In the tapered region, since the horizontal component of the friction force ($\mu F_N cos\lambda$) resists the tightening torque, the required torque to overcome this friction can be obtained by the following equation [119].

$$T_{T_{Taper}} = \left[2\pi\mu_k \int_0^{L_c\cos\theta} b_1^{\,2}(z)P_c(z)dz\right]\cos\lambda \tag{3.13}$$

Then, the total tightening torque can be obtained by inserting Equation 3.12 and Equation 3.13 into Equation 3.11. Note that in the equations for tightening torque and resistive force, kinetic friction coefficient is utilized since tightening is a dynamic process.

3.2.3.2 Mechanics of Loosening

The equations for the case of loosening can be obtained in a similar way to tightening. The distinction is that during loosening, the friction force acts in the opposite direction in comparison to tightening. Figure 3.7b illustrates the equilibrium of forces on the free body diagram of the tapered part of the abutment during loosening.

The *total loosening torque* for the abutment, T_L, is the sum of the resistive torques in the tapered part and in the threaded region,

$$T_L = T_{L_{Screw}} + T_{L_{Taper}} \tag{3.14}$$

The preload of the screw which resists loosening, at the beginning of loosening process, can be obtained from

$$F_L = F_N \sin\theta - \mu_s F_N \sin\lambda\cos\theta \tag{3.15}$$

where the static friction coefficient μ_s is used since loosening starts from a static equilibrium state. The negative sign in this equation indicates that for certain combinations of taper angle, θ, screw lead angle, λ, and friction coefficient, μ_s, the screw preload might become zero or negative. In such a case, the threaded section displays no contribution to the resistance against loosening, leaving it totally to the conical section. Therefore, when the value of F_L is negative, the resistive torque in the screw threads becomes zero. Hence, the resistive torque in the screw threads due to preload can be calculated by including a Heaviside step function $\Phi(F_L)$ into the power screw formula for lowering a load

$$T_{L_{Screw}} = \frac{F_L d_m}{2}\left(\frac{\pi\mu_s d_m \sec\alpha - l}{\pi d_m + \mu_s l \sec\alpha}\right)\Phi(F_L) \tag{3.16}$$

where the Heaviside step function is

$$\begin{aligned} \Phi(F_L) &= 1 \quad \text{if} \quad F_L > 0 \\ \Phi(F_L) &= 0 \quad \text{if} \quad F_L < 0 \end{aligned} \tag{3.17}$$

The resistive torque against loosening in tapered section can be obtained by substituting static friction coefficient in place of the kinetic friction coefficient in Equation 3.13

$$T_{T_{Taper}} = \left[2\pi\mu_s \int_0^{L_c\cos\theta} b_1^2(z)P_c(z)dz\right]\cos\lambda \tag{3.18}$$

Then, the total resistive torque against loosening can be found by inserting Equation 3.16 and Equation 3.18 into Equation 3.14. In the study of Bozkaya and Müftü, a definition of efficiency for taper integrated screw connection was made as the ratio of the total loosening torque to the total tightening torque

$$\eta = \frac{T_L}{T_T} \tag{3.19}$$

This efficiency definition is very helpful for assessing the success of the design of the interface between implant and abutment.

3.2.3.3 Mechanics During Functional Loading

After the abutment screw is tightened to the implant with a certain tightening torque, the artificial crown is placed onto the abutment. Then, implant supported restoration becomes ready for patient use. During mastication or biting, the implant is subjected to functional loading. Under functional loading, unlike the tightening and loosening cases, the friction force at the tapered region is not along the helical path. The screw preload, when a biting force F_b is applied on the system, can be determined from the free body diagram given in Figure 3.8 as follows,

$$F_{FL} = \left(F_N + F_{Nb}\right)\left(\mu_s \cos\theta + \sin\theta\right) - F_b \tag{3.20}$$

where F_{FL} is the screw preload under functional loading, F_N is the resultant normal contact force due to initial interference during tightening, F_{Nb} is the additional normal force due to the biting force F_b.

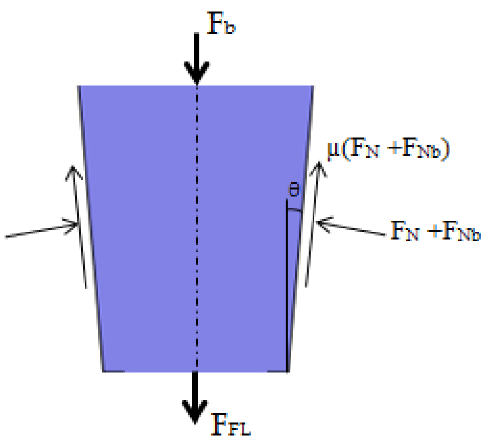


Figure 3.8 The free body diagram of the conical section during functional loading

For the implants involving screw connection only, the compressive biting forces lower the pretension in the screw possibly leading to loosening of the screw. In contrast, in tapered interference fit abutments, loosening problem is less significant. The biting forces acting in the direction of abutment insertion increase the degree of engagement in the conical region, thus, contribute to secure the connection instead of causing loosening. Only application of a loosening torque may cause the tapered interference fit connection to loosen. As a result, it can be concluded that in a taper integrated screw connection, the tapered section dominates the resistive behavior against loosening.

The relations given above are used to analyze the implant-abutment interface for the elastic dental implant model proposed in this study and the numerical findings are supplied in Chapter 4. The results are compared with values obtained for ITI implant and discussed in detail.

3.3 Finite Element Analysis of the Elastic Implant System

Having completed the preliminary geometric design of the proposed dental implant system, a mechanical analysis of the entire model is to be carried out in order to make an optimization on the design in accordance with the objectives of the study specifically stated in Section 1.7. In the previous section, the implant-abutment connection was modeled by analytical relations derived in terms of geometrical, material and surface properties of the components. In other words, the problem of loosening at the implant-abutment interface, one of the main problems aimed to be overcome in this study, was handled by an analytical approach. However, when the whole system, especially the upper structure (abutment) of the proposed design, is considered, it can be observed that a multi-component structure with mechanically interacting parts is involved such that it is difficult to model and achieve a solution by an analytical procedure. Hence, a numerical approach is required to be implemented.

Finite element analysis (FE analysis) is an effective numerical method for obtaining solutions to a myriad of engineering problems in many fields, with the powerful complex problem solving capability involved. Dental biomechanics is one of the fields, investigators on which have commonly utilized and benefited from FE analysis technique as a computational tool to solve complicated structural mechanics problems.

For the present study, FE analysis was implemented in order to

1. Optimize the mechanical design parameters of the proposed dental implant system so that it would

a. Yield a very similar axial mobility behavior under loading as that of a natural tooth by simulating the mechanical behavior of the periodontal ligament,
b. Withstand functional loading at component level without mechanical failure, especially, considering the internal screw and elastic parts,

2. Compare the stresses and displacements within the proposed elastic system with those in a totally rigid counterpart by modeling both systems separately as supporting a dental bridge in combination with a natural tooth.

Prior to beginning the analyses using FEM, the axial mobility behavior of natural teeth as the major criteria for optimization should be defined precisely. Based on the experimental data obtained by Boldt et al. [156], force-displacement relationships under axial loading for two upper incisors and two implant-supported teeth, are presented in Figure 3.9 and Figure 3.10, respectively.

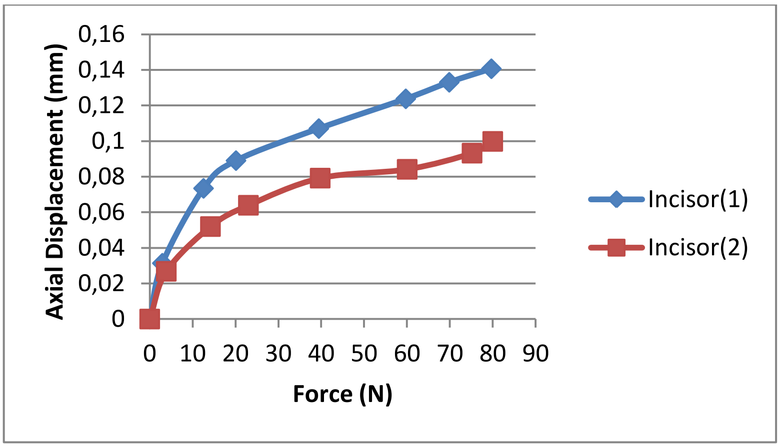


Figure 3.9 The force-displacement relationship for upper incisors under axial loading [156]

It is apparent that the natural teeth display a greater mobility under axial loading when compared to implant-supported teeth. The differential mobility originates from the rigid structure of dental implants and their interface with bone in contrast to the natural teeth which involve an elastic tissue, called the periodontal ligament (PDL), at tooth-bone interface. Under a maximum axial static load of 80 N, the displacements of the natural incisors are in the range of 0.10-0.14 mm, whereas the displacements of the teeth supported by rigid dental implants do not exceed 0.08 mm. It should be noted that the movement of the implant-supported teeth arise from the deformation of the surrounding bone, whereas the movement of the natural teeth depends on both the deformation of the PDL and that of surrounding bone. Therefore, the displacement data for incisors, shown in Figure 3.9, comprise the displacements due to PDL deformation appended on the values caused by bone deformation.

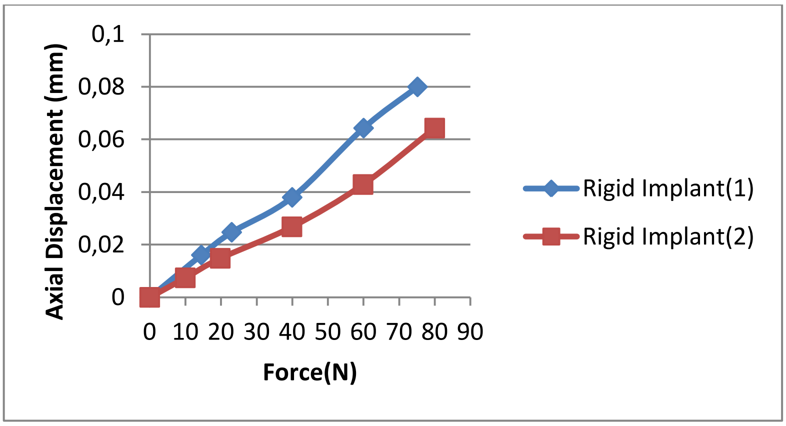


Figure 3.10 The force-displacement relationship for two crowns supported by rigid implants in the upper front jaw [156]

In order to simulate the natural case by implementing an elastic dental implant system that would compensate the differential mobility, the force-displacement relationship of a natural tooth depending purely on the deformation of the PDL was to be determined by eliminating the effects of bone deformation. By this way, the required mobility to be possessed within the elastic implant system being designed could be derived. For this purpose, a new force-displacement curve, referred as reduced displacement curve for natural teeth, was obtained by subtracting the amount of displacement due to bone deformation from the total deformation at each data point on the curve for incisor(1) in Figure 3.9. The values due to bone deformation at each data point were calculated by multiplying the force values shown in horizontal axis with the slope of the fitted curve to the data for rigid implant(2) in Figure 3.10. By fitting a curve to the resulting data, the reduced displacement curve for an incisor was obtained as shown in Figure 3.11.

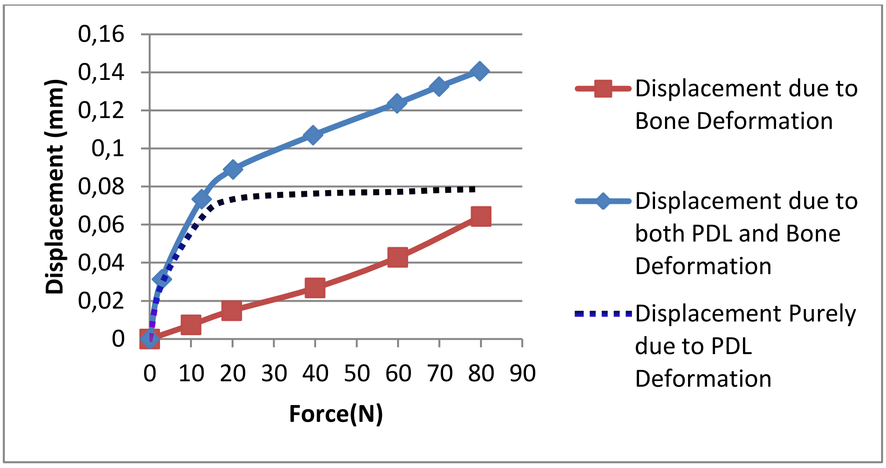


Figure 3.11 The reduced force-displacement curve for an incisor (dashed line)

Considering the reduced curve, it can be observed that the displacement due to PDL deformation exhibits a rapid increase up to a force level of about 20 N, after which it very slightly contributes to the total deformation. In fact, the total displacement curve involves a biphasic nature. In the first phase the effect of the PDL dominates, whereas in the second phase the predominant linear behavior is governed by the deformation of bone.

The data manipulated above are valid for the incisor region of the jaw. For the molar region, however, tooth displacement values are smaller. The force-displacement relationship for a molar tooth compared with an incisor is supplied in Figure 3.12, by employing the experimental data present in the literature [55]. For low forces with magnitudes under 5 N the displacement of an incisor could exceed 0.04 mm, whereas for a molar tooth stiffness increases after reaching a displacement of 0.02 mm. Both of the incisor and molar teeth display a nonlinear force-displacement behavior due to nonlinear mechanical characteristics of the PDL. Therefore, in order to mimic the mechanical behavior of the PDL, a nonlinear modeling should be implemented during the design of an elastic dental implant. In this study, the behavior of the incisor based on the reduced curve in Figure 3.11 will be considered as the target to be achieved throughout the design optimization procedure from this point on.

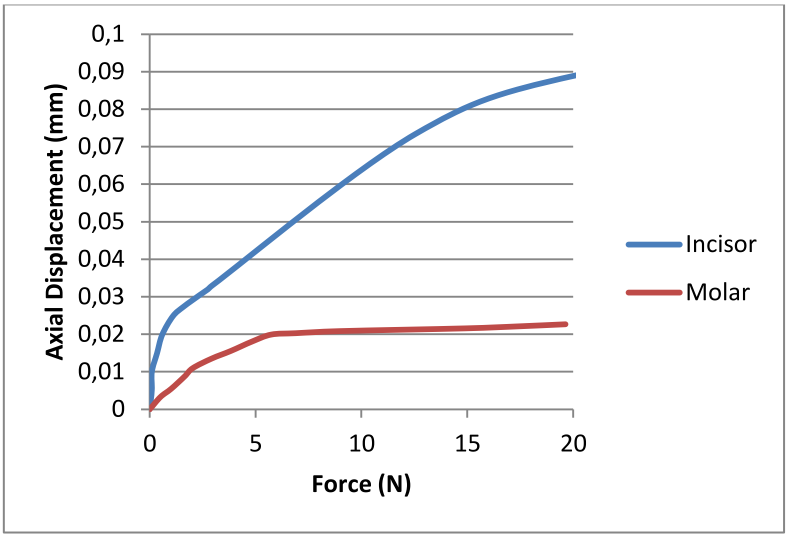


Figure 3.12 The force-displacement relationship of a molar in comparison with an incisor

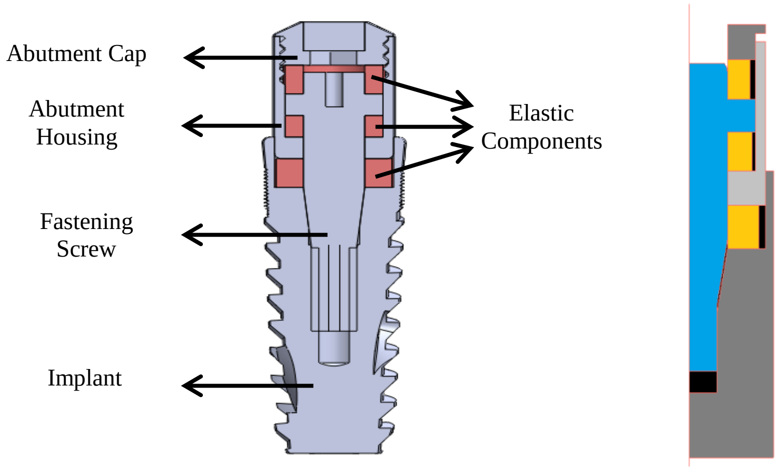
The FE analyses were carried out by following a two-stage procedure. Firstly, a series of two-dimensional axisymmetric analyses was conducted to iteratively update the starting geometric model, which was the output of the conceptual design phase, presented in Figure 3.3e. In this first stage, it was aimed to benefit from the advantages of axisymmetric analysis like simplicity, easy managing and fast solution, which helped save analysis resources like time and memory, significantly. Secondly, the final geometric model attained as a resulted of the first stage of analyses was utilized to carry out three-dimensional analyses involving more realistic modeling conditions. Besides, the comparative analyses between elastic and rigid systems were accomplished in the three-dimensional modeling and analysis stage.

3.3.1 Axisymmetric Analysis

In general, axisymmetric analysis is a simplified method for finite element analysis (FEA) appropriate when the geometric model, load and boundary conditions show symmetry around a central axis. The geometric model that is to be used in the axisymmetric analysis is a two-dimensional section which creates the original three-dimensional model when revolved around the central axis. Screw type cylindrical dental implants, under axial loading conditions, perfectly conform to the requirements of an axisymmetric analysis. In this part of the study, the design parameters of proposed dental implant system were tuned so as to result in the desired mobility behavior represented by the reduced curve in Figure 3.11, by the aid of axisymmetric FEA carried out in a commercial FEA software, MSC Marc Mentat 2010.

3.3.1.1 Geometric Modeling

The first step was to create the geometric model with axisymmetric properties. From the conceptual design, the section view of which is shown in Figure 3.13a, the axisymmetric geometric model was created as presented in Figure 3.13b. The implant in the geometric model had a diameter of 4.2 mm and a length of 9.0 mm, as stated in the design of the implant in Section 3.2.1.



(a) **(b)**

Figure 3.13 The representation of the axisymmetric geometric model obtained from the conceptual design: (a) The conceptual design, (b) The axisymmetric geometric model

Note that in the axisymmetric geometric model the screw threads were not included just to further simplify the problem. Actually the threads on the implant surface would not affect results of the analysis, since implant-bone interface was not modeled and not considered throughout the analyses. The focus in all of the analyses was on the mechanical response within the dental implant system. Furthermore, as can be seen from Figure 3.13b, there exists a gap between the abutment cap and abutment housing in addition to that between conical surfaces of fastening screw and implant. The given geometric model illustrates the positions of the components prior to application of load and boundary conditions. The mentioned gaps in the model serve for applying preload to the elastic components during fastening the assembly before functional loading. The details about the application of preload will be explained later in Section 3.3.1.4.

3.3.1.2 Material Modeling

The components included in the proposed dental implant system can be divided into two groups in terms of material properties: (1) components with resilient behavior and (2) relatively rigid components.

3.3.1.2.1 Relatively Rigid Components

The implant, abutment housing, abutment screw and abutment cap are relatively rigid components of the proposed dental implant system. The material for these parts was defined to be Ti-6Al-4V (annealed), a titanium alloy that has been proven to display perfect biocompatibility and excellent mechanical properties like fracture toughness, yield strength and fatigue strength. Therefore, it has been commonly used in biomedical applications, especially as a dental implant material by commercial dental implant manufacturers. Some of the important mechanical properties of Ti-6Al-4V (annealed) are presented in Table 3.1 [162].

The relatively rigid components were assumed to have simply isotropic and linear elastic behavior involving small strain and small deformation.

Table 3.1 Mechanical properties of Ti-6Al-4V (annealed)

Property	Value
Elastic Modulus (GPa)	100-114
Yield Strength (MPa)	825-869
Ultimate Tensile Strength (MPa)	895-930
Ultimate Tensile Strain (%)	6-10
Poisson’s Ratio	0.342
Brinell Hardness	326

3.3.1.2.2 Resilient Components

For the resilient components simulating complex nonlinear elastic behavior of the periodontal ligament, a more complicated nonlinear material model appropriate for large strain and large deformation must be used and a material with high elasticity should be selected.

a. Constitutive Material Model

The problem was simplified by assuming the material of the elastic components to be isotropic, homogenous, continuous and incompressible. Note that the material has isotropic properties only in the undeformed configuration, whereas in the deformed configuration it becomes anisotropic due to unsymmetrical relative displacements of the points inside the material under arbitrary loading conditions. Moreover, a material that is not of porous type can be perfectly considered as continuous and homogeneous. On the other hand, if the volumetric changes are small compared to the initial volume, the material can also be assumed to be incompressible.

For the desired properties, an appropriate choice might be a rubber-like material whose stress-strain relationship could be defined as non-linearly elastic, isotropic, incompressible and generally independent of strain rate. In order to model the stress-strain behavior of such materials, generally, hyperelasticity concept is used. Hyperelastic idealization is a common method for modeling filled elastomers and biological tissues. The stress-strain relationship of a hyperelastic material is characterized by a strain energy density function (W) [163].

In the analysis of such an elastomeric structure, one important consideration is nonlinearity which can be categorized as material, boundary and geometric nonlinearities. Rubber materials usually behave in a nonlinear manner (material nonlinearity). Contact surfaces between elastomeric components usually cause boundary changes during analysis. This results in boundary nonlinearities. Moreover, when the magnitude of displacement affects the response of the structure, geometric nonlinearity occurs.

An efficient and appropriate mathematical model was required to determine the material behavior in terms of stress-strain relationship to be used in finite element modeling. The constitutive model would be composed of constants fitting to experimental data and measures of deformation. Hyperelastic materials store energy under applied loads and this energy is called strain energy. If viscous effects are involved, mechanical energy is also dissipated under the action of forces. The strain energy density function depends both on measures of deformation and material constants that are experimentally determined.

As a nonlinear, isotropic hyperelastic material model for the desired behavior of the elastic part, the Mooney-Rivlin material model was utilized. It has been experimentally verified that this material model works well with strains up to 200% [164], which is higher than the strains expected to be experienced within the elastic parts of the proposed dental implant system under functional loading. Therefore, the choice of the material model conformed well to the present problem handled in this study. In Mooney-Rivlin model the strain energy is expressed as a linear combination of invariants of Finger tensor (or inverse of the left Cauchy-Green deformation tensor) B, which is defined as

$$B = FF^T \quad \text{or} \quad B_{ij} = \frac{\partial x_i}{\partial X_k}\frac{\partial x_j}{\partial X_k} \tag{3.21}$$

where $F_{ik} = \frac{\partial x_i}{\partial X_k}$ is the spatial deformation gradient tensor [165]. The strain energy density function, in Mooney-Rivlin material model [166], is based on polynomials

$$W = \sum_{i+j=1}^{m} C_{ij} (I_1 - 3)^i (I_2 - 3)^j + \frac{1}{2}\kappa(I_3 - 1) \quad (3.22)$$

where C_{ij} denotes the material constants calculated from experimentally obtained data, m is the order of model and κ is the bulk modulus. I_1, I_2 and I_3 are the invariants of the left Cauchy-Green deformation tensor B, defined as

$$I_1 = \lambda_1^2 + \lambda_2^2 + \lambda_3^2 \quad (3.23)$$

$$I_2 = \lambda_1^2\lambda_2^2 + \lambda_2^2\lambda_3^2 + \lambda_3^2\lambda_1^2 \quad (3.24)$$

$$I_3 = \lambda_1^2\lambda_2^2\lambda_3^2 = (\frac{V}{V_0})^2 \quad (3.25)$$

with V and V_0 being volumes after and before deformation, respectively. The terms λ_i in the expressions of invariants are the principal stretches for i= 1, 2, 3.

$$\lambda_i = \frac{l_i}{l_{i,0}} \quad (3.26)$$

where l_i is the deformed length and $l_{i,0}$ is the undeformed initial length in the direction i.

Equation 3.22 defines the strain energy in a generalized form of Mooney-Rivlin material model, considering compressibility of the material by the last term involving bulk modulus. For incompressible assumption, the third invariant is equal to 1 since the volume is assumed to be unchanged after deformation. Thus, the last term in the strain energy density function vanishes, reducing Equation 3.21 to the following form:

$$W = \sum_{i+j=1}^{m} C_{ij} (I_1 - 3)^i (I_2 - 3)^j \quad (3.27)$$

In general, two-, three-, five-, and nine-term Mooney-Rivlin models are used in commercial software packages. More terms serve to capture more inflection points in the engineering stress-strain curve. However, using higher order terms can cause large oscillations outside the experimental data range without any physical meaning [164]. In particular, in the present study, for modeling the materials used in the finite element analyses two- and three-term Mooney-Rivlin models were used. The expressions of strain energy can be written as

$$W = C_{10}(I_1 - 3) + C_{01}(I_2 - 3) \tag{3.28}$$

$$W = C_{10}(I_1 - 3) + C_{01}(I_2 - 3) + C_{11}(I_1 - 3)(I_2 - 3) \tag{3.29}$$

for two- and three-term Mooney-Rivlin models, respectively. After determining appropriate model and deriving material coefficients from experimental data, the principal stresses are calculated from the equation

$$\sigma_i = -p + \lambda_i \frac{\partial W}{\partial \lambda_i} \tag{3.30}$$

where p is the hydrostatic pressure.

Upon selecting the material model and defining the experimental test data, the finite element analysis software automatically calculates the material coefficients C_{ij}. Furthermore, based on the strain energy density function, it determines the stresses and strains within the geometric model by performing the analysis.

b. Material Selection

Instead of assuming the availability of a material that best conforms to the requirements for the mechanical behavior expected from the elastic parts included in the dental implant system, the methodology in this study was to choose among already available materials in the market, to check their convenience via finite element analysis and to decide the best material yielding the elastic behavior which approaches most to the desired one explained above.

In many engineering applications including the cases that require high compressive load bearing capacity, engineering polymers like polyamide, polyurethane, polyoxymethylene, polyetherketone and polyethylene are frequently used. Among these alternatives, due to its

- high flexibility with good tensile and compressive strength,
- very high load bearing capacity,

– outstanding abrasion, tear and flex crack resistance,
– enhanced fatigue durability,
– extra toughness,
– very good impact resistance [167, 168] and
– most importantly favorable biocompatibility [169],

polyurethane is a good candidate for the elastic components of the proposed dental implant system. In the market, polyurethane is available in two forms with respect to the chemical structure, namely, ester- and ether- based polyurethanes. Ester-based polyurethanes (polyesters) are damaged by microbiological attack when subjected to heat and humidity, whereas ether-based polyurethanes (polyethers) are resistant to such conditions [170]. Hence, ether-based polyurethanes should be selected so that the elastic components which will be subjected to heat and humidity in the oral environment can resist against microbiological attack.

Engineering polyurethanes are classified with respect to hardness as demonstrated in Figure 3.14. In the hardness scale, polyurethanes remain in the range between 20 Shore A to 80 Shore D Durometer. In this study, polyurethanes with 80, 85, 90 and 95 Shore A Durometer hardness are employed and compared throughout the finite element analyses.

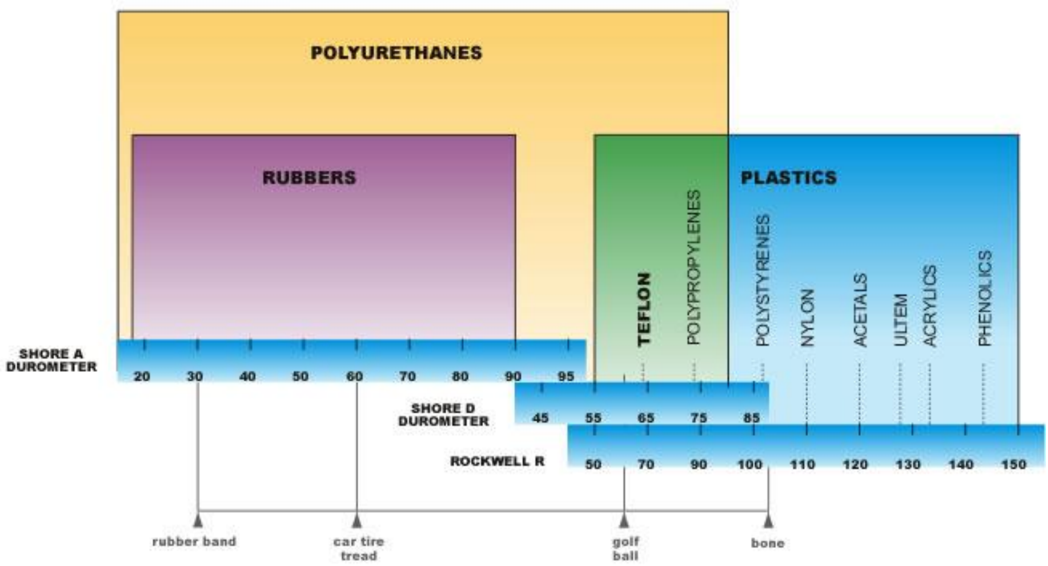


Figure 3.14 The hardness scale for polyurethanes, rubbers and plastics [171]

Another important consideration substantially affecting the compression-deflection relationship of polyurethanes is the concept of shape factor. Shape factor is defined as the ratio of the area of one loaded surface to the total area of the unloaded surfaces that are free to bulge. For an elastomeric component having two parallel load faces, the shape factor can be defined and has critical impact on deformation behavior under compressive loads. Higher the shape factor, lower the amount of deflection and higher the load bearing capacity. In other words, as shape factor increases, the load required to induce a certain strain also increases. On the other hand, for the parts made from same material and having identical shape factors, the behavior under compression is the same irrespective of the actual geometric shape or actual size. In addition, parts with shape factors below 0.25 are prone to buckling under compression [172, 173].

In spite of the fact that no mathematical correlations associating the shape factor to the compressive modulus have been derived, several polyurethane manufacturers supply the empirical data corresponding to their products. Figure 3.15 illustrates the effect of shape factor on the stress-strain behavior of polyurethane with different Shore A Durometer hardness values [172]. It can be observed from these experimental data that as shape factor increases, the compressive stress corresponding to the same compressive strain becomes higher.

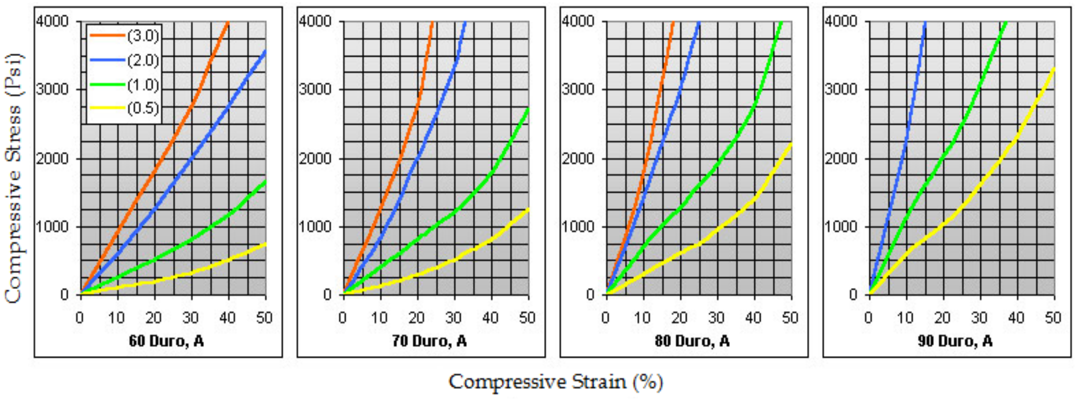


Figure 3.15 The effect of shape factor on the compressive stress-strain relationship of polyurethanes with different hardness values (each color indicate a distinct shape factor)

The elastic parts in the proposed dental implant system have a hollow cylindrical shape (Figure 3.16). The loaded surfaces are parallel to each other providing the appropriate condition to implement the shape factor concept. The shape factor (A_1/A_2) for three elastic components in the system remain in the range of 0.5-0.6. Thus, a shape factor of 0.56, for which experimental data is available, was selected.

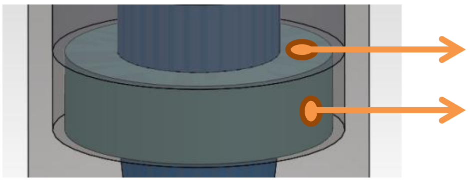


Figure 3.16 The loaded and free surfaces of the elastic components included in the proposed dental implant system

The tensile strength and compression properties of polyurethane with the shape factor of 0.56, varying with hardness are presented in Table 3.2. The tabulated data supplied by the manufacturer is stated to be obtained following standard test methods for rubber in compression (ASTM D575) and tension (ASTM D412).

Table 3.2 Compression properties (with shape factor 0.56) and tensile strength of polyurethane varying with hardness [174]

Compression/Tensile Property	Durometer Hardness		
	85A	90A	95A
Comp. Stress (MPa) @ 5% Deflection	1.24	2.14	3.10
Comp. Stress (MPa) @ 10% Deflection	2.69	4.31	6.14
Comp. Stress (MPa) @ 15% Deflection	3.86	6.03	8.07
Comp. Stress (MPa) @ 20% Deflection	4.76	7.76	9.65
Comp. Stress (MPa) @ 25% Deflection	5.52	9.31	9.65
Tensile Strength (MPa)	41.37	31.03	37.92

The compression strength for polyurethane is not provided by manufacturers. However, it is known that polymers are approximately 20% stronger under compression compared to tension [175]. In addition to the compression data for 85, 90 and 95 Shore A Durometer given in Table 3.2, for 80 Shore A Durometer the compression curve for 0.5 shape factor (in Figure 3.15) was digitized by a software (GetData Graph Digitizer, version 2.24) and the discrete stress-strain values was input in the form of tabulated data to the FEA software during defining material properties. The Mooney-Rivlin coefficients based on the input experimental data was calculated by the software for each of the materials as shown in Table 3.3.

Table 3.3 The Mooney-Rivlin coefficients calculated by the FEA software

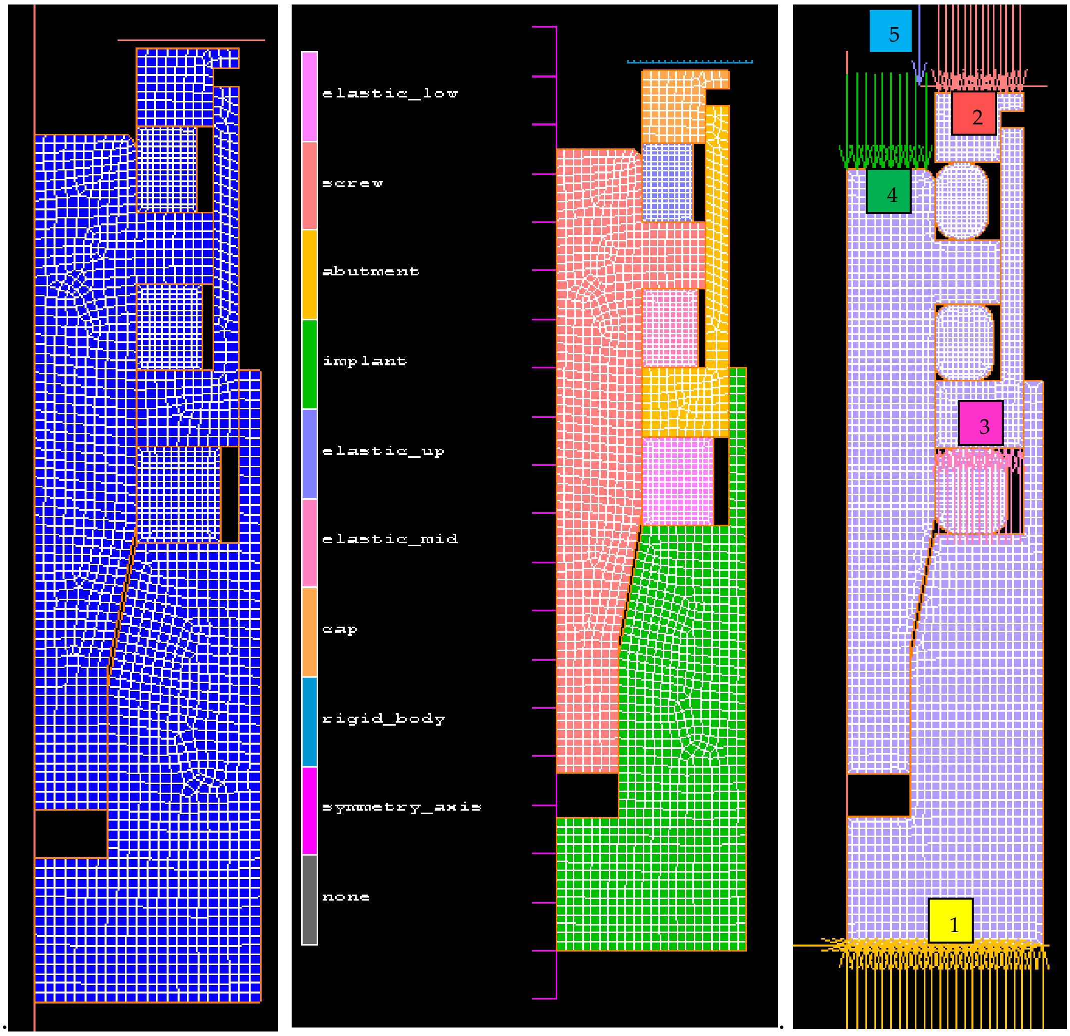
Material	Model Order	C_{10} (MPa)	C_{01} (MPa)	C_{11} (MPa)
80A	3	7.701	-4.069	0.615
85A	2	9.420	-5.026	-
90A	2	15.375	-8.176	-
95A	2	24.790	-14.389	-

3.3.1.3 Mesh Generation and Contact Body Definitions

For the axisymmetric analysis, two-dimensional planar meshing with QUAD(4) elements was utilized. The mesh generated on the starting geometric model (Figure 3.17a) contained totally 3088 elements and 3447 nodes. For the elastic parts, the meshes were considerably refined in order to obtain more accurate results since within these components large deformation gradients occurred during the load increments. Moreover, for all of the elements, axisymmetric solid property type was defined as a geometric property.

All of the components of the dental implant system were defined as deformable contact bodies. As a simplification, no friction was defined between the contact bodies. The blue line above the abutment cap that can be seen in Figure 3.17b represents the rigid body that was defined to simulate the crown. This rigid body was controlled by load with an approach velocity of 5 mm/s. Finally, from the symmetry option of the contact menu, the central symmetry axis was also defined as a contact body.

The appropriate contact definitions between the bodies are made using the contact table option. As seen from Figure 3.18, all of the contact types are selected to be ‘touching’.



(a) (b) (c)

Figure 3.17 (a) The meshes generated on the initial model, (b) the contact body definitions on the initial model, (c) load and boundary conditions on an updated model

Figure 3.18 The contact table for the components in the dental implant system

3.3.1.4 Loads and Boundary Conditions

The load and boundary conditions applied on the model are listed below and illustrated in Figure 3.17c.

1. Fixed displacement condition on implant bottom edge,
2. Edge load on top edge of the abutment cap for prestressing,
3. Edge load on bottom edge of the abutment housing,
4. Fixed displacement for the abutment screw,
5. Vertical point load on the rigid body (simulating the crown)

The application of load and boundary conditions simulated three actual situations which are explained below.

i. *Fixation of the implant inside jawbone*

In most of the previous studies a fixed bond was assumed between the implant and jawbone. Since the aim, in this study, was to approximate the reduced displacement curve based on PDL deformation regardless of bone deformation, there was no need to model the complex behavior of the jawbone and the bone-implant interface. Hence, as a displacement boundary condition

- the bottom edge of the implant body was simply fixed in all directions.

ii. Application of preload on the elastic components

The elastic components in the system display a spring-like action under loading. Prestressing, also termed as presetting, of a compression spring improves the capability of the spring to withstand stress, increases its load carrying capacity and fatigue resistance [176, 177]. By making analogy to compression springs, prestress was applied to the elastic parts that are required to display high fatigue durability. However, this benefit of prestressing was not verified in this study. For applying prestress to the elastic components:

- Two vertical forces equal in magnitude (3.5 N) but in opposite directions were defined, one acting on top edge of the abutment cap and the other on the bottom edge of the abutment housing. These counter forces moved these components towards each other, compressing the middle and upper elastic parts remaining in between. These load conditions closed the initial gap (0.2 mm) left between the abutment cap and abutment housing, fastening them to each other, which simulated the assembly during manufacture.
- A fixed displacement of 0.2 mm was applied to the abutment screw, which caused prestressing of the lower elastic component in addition to the others prestressed in the previous step. This displacement condition closed the initial gap (0.2 mm) between conical surfaces of the abutment screw and implant, simulating the fastening of the abutment to the implant, which is done by clinicians before construction of prosthesis on the implant.

iii. Application of bite force

- In order to simulate the bite forces on the implant-supported crown, in the axisymmetric analysis, subsequent to application of the forces and displacement boundary conditions that would generate prestress, a vertical force of 80 N was applied to the load-controlled rigid body that was contacting to the top edge of the abutment cap. The mobility of a natural tooth under axial forces below 80 N was tried to be approximated. In addition, a vertical force of 800 N, the maximum bite force reported, was applied to the final optimized model to check whether it could withstand the maximum load at component level without mechanical failure.

After properly defining the aforementioned analysis parameters, a series of axisymmetric analyses were carried out. Throughout the analyses conducted by following an iterative procedure, the geometric model, mesh created on the model and the material definitions were varied subsequently until the desired axial mobility behavior was reasonably approached. The interim findings and final results are provided in Chapter 4.

3.3.2 Three-Dimensional Analysis

Having completed the axisymmetric analysis stage, an optimized model for the proposed dental implant system resulting in desired axial mobility behavior was obtained. At this stage, the model which was the output of the previous stage was utilized directly via converting the axisymmetric model to a three-dimensional geometric model. Basically, the investigations in the three-dimensional analyses were concentrated on three points of consideration, which were

1. Verification of the results obtained in the axisymmetric analysis stage by performing a number of analyses keeping the geometric model unchanged but varying the material defined for the elastic parts,
2. In addition to axial loading, checking the strength and reliability of the proposed dental implant system under lateral loading,
3. Comparing the stress and displacements that occur in a bridge prosthesis supported by a natural tooth in combination with a dental implant for two cases of implant selection: an elastic dental implant and a totally rigid dental implant.

The three-dimensional analyses were carried out by using ANSYS 12.0 (as analysis software) and the details about these analyses are explained below.

3.3.2.1 Analysis of a Dental Implant Supporting a Single Crown

The three-dimensional geometric model (Figure 3.19b) was generated, in SolidWorks 2011, from the final optimal two-dimensional axisymmetric geometric model (Figure 3.19a) which was obtained at the end of axisymmetric analyses. The diameter of the implant was 4.2 mm and the length of the implant was 9.0 mm, as before. The threads on the surface of the implant were not included in 3-D models for the same reason as in the axisymmetric analysis. The gaps between components were left for prestressing.

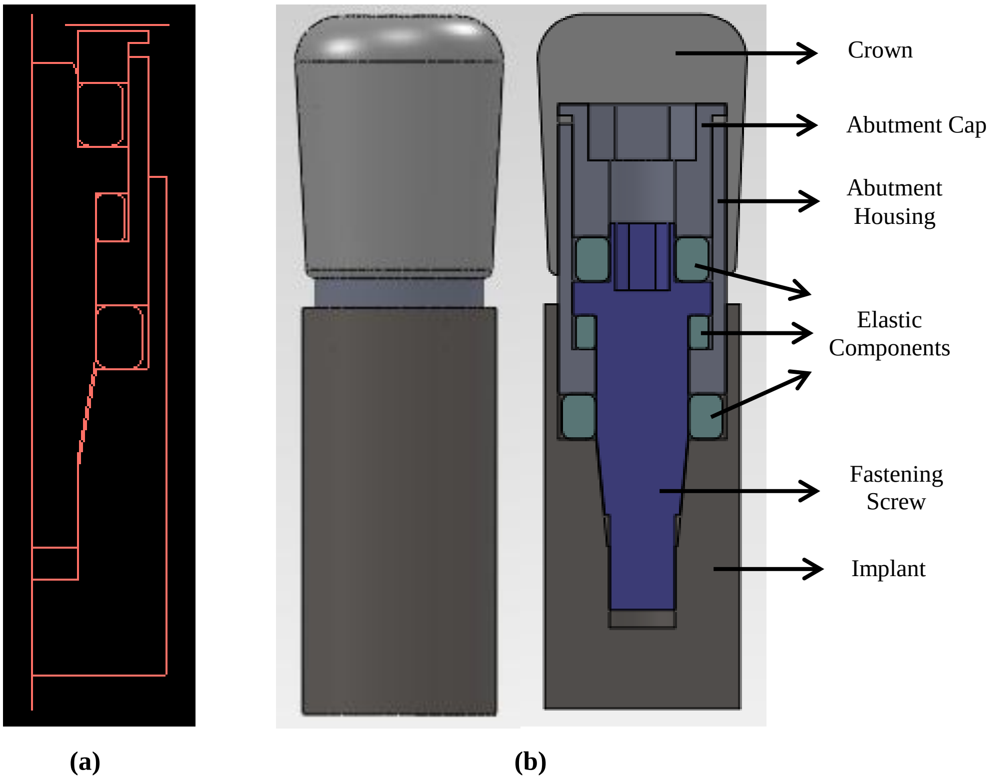
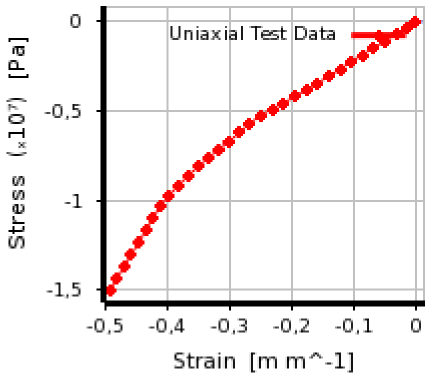
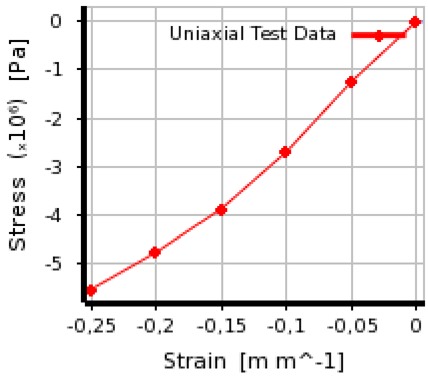


Figure 3.19 The geometric models: (a) optimal axisymmetric model, (b) 3-D model obtained from the axisymmetric model

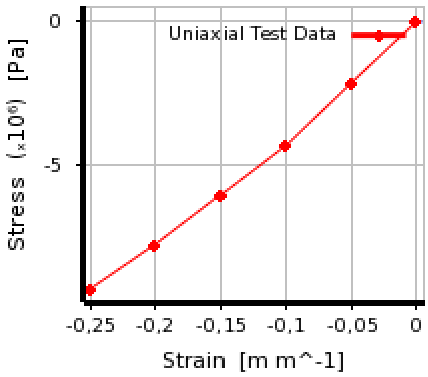
For material modeling, exactly the same procedure and material specifications as in the axisymmetric analyses were used except that the crown was defined as a deformable body with linear elastic, isotropic material properties instead of a rigid body. The Young's modulus and Poisson's ratio of the porcelain crown was defined to be 67.2 GPa and 0.3, respectively. For the material of the elastic components, the resulting effects of defining polyurethane with 80, 85, 90 and 95 Shore A Durometer hardness on the mechanical response of the system were compared. The curve fits of the software to the experimental material data are presented in Figure 3.20. The Mooney-Rivlin coefficients calculated in ANSYS were the same as those in MSC Marc Mentat.



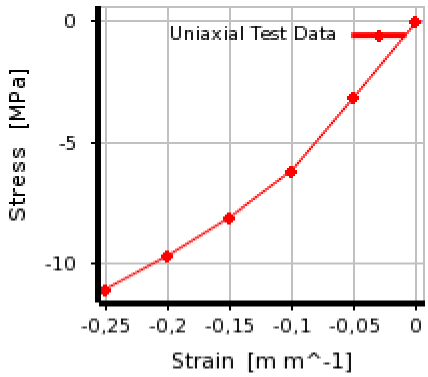
(a) 80A



(b) 85A



(c) 90A



(d) 95A

Figure 3.20 The curve fits to the uniaxial compression test data

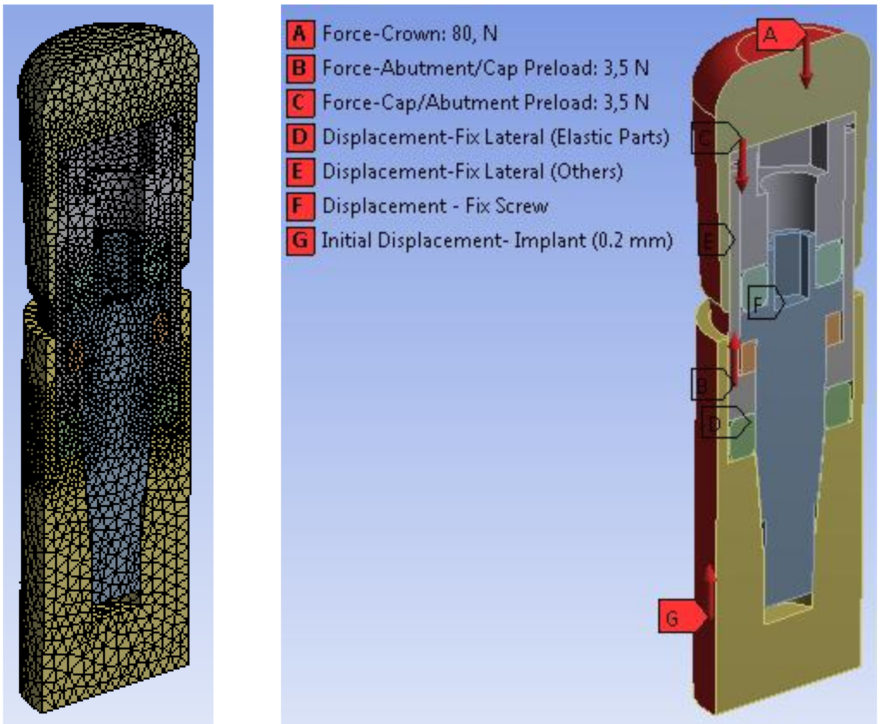
All the contact bodies were defined as deformable bodies. The contact definitions (frictionless contact) were similar to those in the axisymmetric analysis. The difference was that between the crown and abutment cap bonded contact was defined and between the conical surfaces of the abutment screw and implant frictional contact (with a friction coefficient of 0.3) was specified. The frictional contact of the conical surfaces helped to observe the consequences of applying the tapered interference fit concept (explained in Section 3.2.3) on the stresses around this contact region. Besides, for the frictionless contacts between the resilient components and relatively rigid components, 'Augmented Lagrange Formulation' was specified since this formulation works better for sliding conditions at contact regions in spite of requiring additional iterations.

The model was meshed with curvature-based tetrahedral solid mesh using patch conforming method (Figure 3.21a). By using ‘contact sizing’ and ‘body sizing’ options, the element size at the contact regions and in the elastic components which would experience high deformation were defined to be finer (0.16 mm) than that in the other regions (0.35 mm). The total number of nodes, solid elements and contact elements were 154279, 97554 and 28502, respectively. The primary reason for involving a large number of elements and nodes was the existence of multiple deformable contact bodies experiencing large displacements and large deformations.

For the case of axial loading, the definitions of the load and boundary conditions were the same as those defined for the axisymmetric model (Figure 3.21b). One distinction was that the displacement boundary conditions of the implant and the abutment screw were exchanged. In other words, the location of the screw was fixed and a displacement of 0.2 mm towards the screw was defined for the implant, which created the same effect of prestressing the lower elastic component and simulating the fastening of the screw to the implant as before. This exchange of boundary condition facilitated the convergence of the solution. In addition, the relatively rigid components and the inner surface of the elastic components were restrained against moving along both axes (x and y) of the horizontal plane, enabling only vertical movement along z-axis.

For the case of lateral loading, the vertical force acting on the crown was converted to a horizontal force of 100 N (along x-axis), simulating the maximum lateral load on a natural tooth. This magnitude of horizontal load is actually higher than the values reported in previous studies [81]; therefore, it would be safe to check the strength of the system against lateral loading with this load definition. Moreover, the definition of the restriction of the components against moving along the two axes in horizontal plane was changed so as to constrain the movement only along only y-axis, enabling displacement along x-axis on horizontal plane and z-axis on vertical plane.

The loads on the crown were applied incrementally, an iterative solver type was selected and large deflection option was activated as analysis settings before running the analyses. The results obtained are presented in Chapter 4.



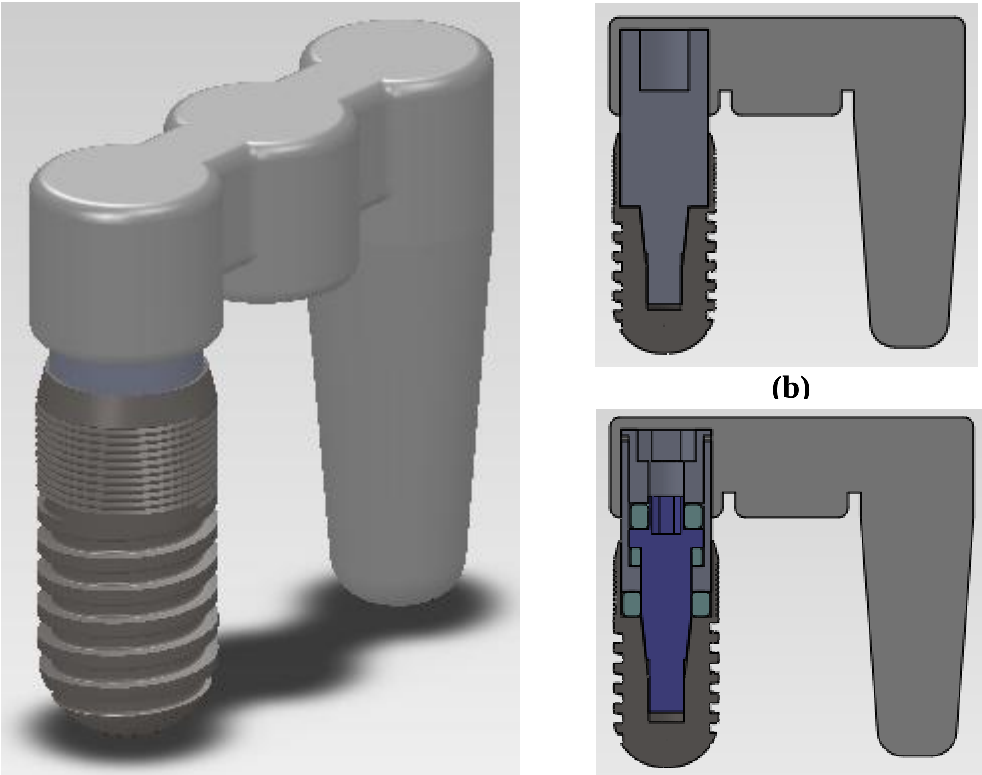
(a) **(b)**

Figure 3.21 (a) Mesh on the model, (b) Load and boundary conditions

3.3.2.2 Analysis of a Dental Bridge Supported by a Dental Implant

It is a common clinical treatment for tooth loss to construct a dental bridge which is supported by one or more dental implants in combination with natural teeth. The simplest case is a dental bridge co-supported by a single dental implant and a single tooth. In this section, a novel methodology is introduced to model and compare a rigid and an elastic dental implant supporting a dental bridge with respect to the stresses and displacements that occur in the bridge prosthesis and in the dental implant system.

The three-dimensional geometric models of the described dental bridges were created separately in SolidWorks 2011 as shown in Figure 3.22. In order to simplify the situation, the bridge prosthesis and supporting natural tooth was modeled in a single piece. Moreover, in the analysis the threads on the implant surface were suppressed for the same purpose. For the material modeling and mesh generation, the same procedure explained in the previous section applies to this problem. A vertical load of 40 N was applied on the top surface of the crown bridge as shown in Figure 3.23.



(b)

(a) (c)

Figure 3.22 The geometric models of dental bridges: (a) full model, (b) sectional view of the bridge with rigid implant, (c) sectional view of bridge with elastic implant

For the elastic implant case, the same load and boundary conditions for prestressing the elastic components were applied and identical contact definitions were made as described in the previous section. For the rigid implant case, the only boundary condition applied on the implant side was the fixation of the implant in all axes. The contact between crown bridge and abutment were defined to be bonded contact, whereas frictional contact was defined between abutment and implant.

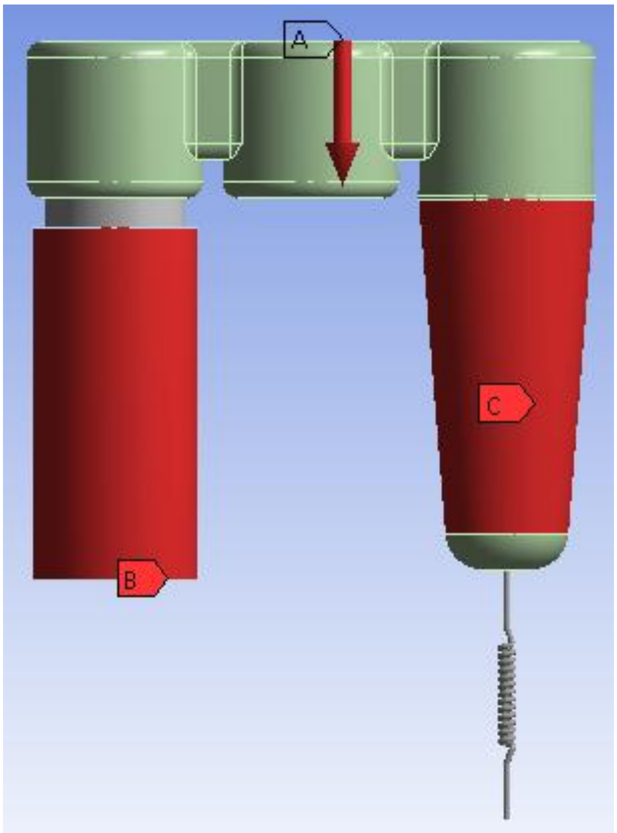


Figure 3.23 The load and boundary conditions on the dental bridge model

For the natural tooth side, the behavior of the periodontal ligament was to be considered. The periodontal ligament surrounding the roots of natural teeth provides a spring-like action under orthodontic loading. In the previous studies investigating the mechanical behavior of the PDL under loading, the PDL and the jawbone has been geometrically modeled in three-dimension. In most of those studies, linear elastic assumption was made and certain elastic modulus values were defined both for the PDL and the surrounding jawbone. However, in the real case, the PDL, jawbone and the contact between them display a complex nonlinear behavior. Therefore, it is difficult to model and obtain accurate results by including the geometric models of them in the finite element analysis.

In this study, in order to simulate the mechanical behavior of the periodontal ligament of the natural tooth subjected to vertical loading, the natural tooth supporting the crown bridge was connected to the ground by placing a nonlinear spring in between the bottom surface of the tooth and the ground, for both rigid and elastic implant cases, as illustrated in

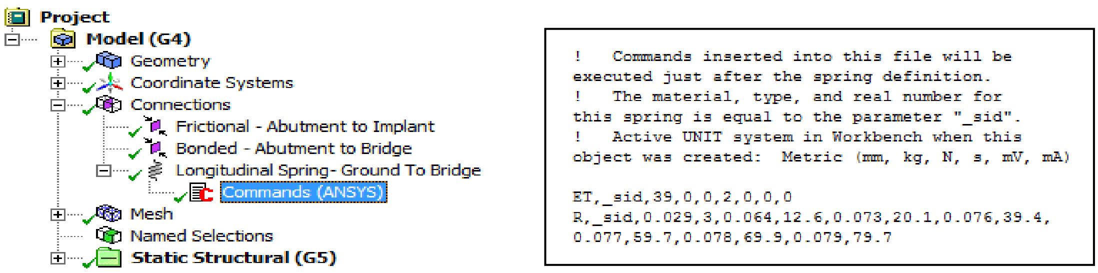
Figure 3.23. The force-displacement behavior of the nonlinear spring was defined based on the reduced curve for a natural tooth which was derived at the beginning of Section 3.3. The reduced curve reflected the force-displacement behavior of a natural tooth depending purely on PDL deformation, excluding the displacements due to bone deformation. As a consequence, there was no need to model the PDL since the behavior of the PDL was replicated by a nonlinear spring. In addition, bone modeling was also not required, since the effects of the bone was eliminated at both implant side that was fixed and natural tooth side that was modeled by a spring excluding the effects of bone deformation. In fact, in this study, by employing en elastic dental implant, it was tried to mimic the displacement of a natural tooth caused purely by PDL deformation since bone deformation occurs around both an implant and a natural tooth root, leaving the only distinction to be the extra displacement originating from the elasticity of the PDL. The exact purpose was to compensate this extra mobility by incorporating elasticity into the dental implant instead of using a completely rigid system which provides almost no mobility. In a dental bridge prosthesis structure, achieving similar mobility at implant side to that at natural tooth side would prevent stress concentrations in the crown bridge and avoid overloading of the dental implant, providing uniform stress distribution on the whole prosthesis structure.

The discrete force-displacement data constituting the reduced curve is presented in Table 3.4, and was used to define the behavior of the nonlinear spring. Based on the data in Table 3.4, a command definition was attached to the spring connection as shown in Figure 3.25.

Table 3.4 The discrete force-displacement data of the reduced curve for a natural tooth

Force (N)	Displacement (mm)
0	0
3.0	0.029
12.6	0.064
20.1	0.073
39.4	0.076
59.7	0.077
69.9	0.078
79.7	0.079

Note that in both cases (rigid and elastic implant co-supports) the tooth side was not supported along the axis parallel to the bridge, but it was constrained not to move along the axis perpendicular to this axis in horizontal plane. Therefore, the tooth was allowed for lateral movements along the axis parallel to the bridge as in the case of a natural tooth in order to observe the expected rotational movement of the crown bridge in the case of rigid implant co-support.



(a) **(b)**

Figure 3.24 Defining the behavior of nonlinear spring: (a) attaching command to the spring definition, (b) the description of the command

After completing all the definitions of analysis parameters, the two models were analyzed separately. The stress distribution and the magnitude of displacements in both of the models were investigated and compared with each other. The results are presented and discussed in Chapter 4.

CHAPTER 4

RESULTS AND DISCUSSION

4.1 Results Related with the Design of the Implant-Abutment Interface

In order to cope with the problem of loosening of the abutment screw, it was proposed to incorporate an internal conical connection combined with a screw into the design of the elastic dental implant system in Section 3.2.3. For this purpose, an analytical approach (the concept of taper integrated screw for abutments) explained in the study of Bozkaya and Müftü [106] was implemented. The success of the design was assessed by comparing the results with those for ITI implant, a commercial rigid dental implant, involving a similar abutment-implant connection.

All the geometrical design parameters (presented in Figure 3.6) optimized for the proposed design and same parameters for the ITI implant are presented in Table 4.1. The details of the optimization procedure are explained below. In addition, other required material and surface related properties (same for the proposed design and ITI implant) are provided in Table 4.2. Based on these optimal parameters, the results obtained from analytical relations are summarized in Table 4.3.

Table 4.1 The geometric design parameters used in the calculations (for the proposed design and the commercial ITI implant)

Implant Type	b_2 (mm)	r_{it} (mm)	r_{ab} (mm)	θ (deg)	Δz (mm)	δ (mm)	L_c (mm)	d_m (mm)	α (deg)	l (mm)
Proposed Design	2.1	1	0.8	6	0.020	0.0021	1.933	1.205	30	0.3
ITI Implant	2.24	1.517	1.42	8	0.036	0.0051	0.731	0.875	30	0.44

Table 4.2 The properties related with material and surface of the parts

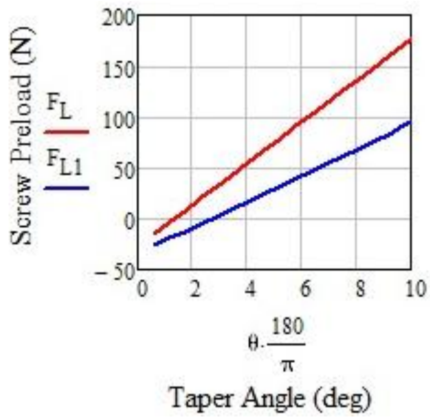
E (GPa)	μ_s	μ_k
113.8	0.3	0.3

Table 4.3 The results of the analytical calculations for the proposed design and ITI implant

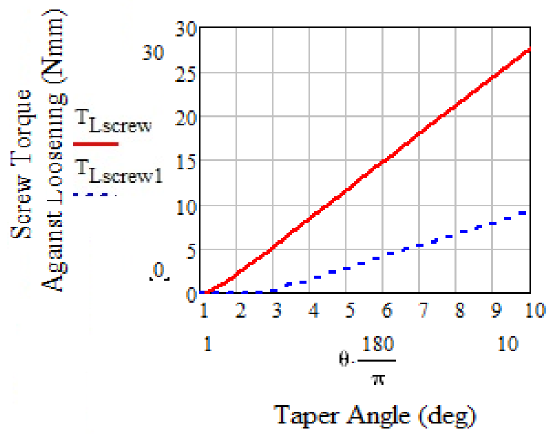
Result Name		Proposed Design	ITI Implant
For Tightening	F_N (Normal force, N)	1171	737
	F_T (Screw preload, N)	150.1	137.3
	$T_{T_{Screw}}$ (Screw resistive torque, N·mm)	39.6	32.2
	$T_{T_{Taper}}$ (Taper resistive torque, N·mm)	315.1	321.1
	T_T (Total tightening torque, N·mm)	354.7	353.3
For Loosening	F_N (Normal force, N)	1171	737
	F_L (Screw preload, N)	94.8	67.9
	$T_{L_{Screw}}$ (Screw resistive torque, N·mm)	14.9	5.2
	$T_{L_{Taper}}$ (Taper resistive torque, N·mm)	315.1	321.1
	T_L (Total loosening torque, N·mm)	329.9	326.3
η (Efficiency of the tapered-with-screw connection, $\frac{T_L}{T_T}$)		0.930	0.924

Before discussing the results in Table 4.3, the procedure for optimization of the geometrical design parameters should be explained. First of all, by using the equations provided in Section 3.2.3, each of the force and torque expressions were parametrically defined, in Mathcad® 14, in terms of the parameters mentioned above. After parametrization, the effect of the important variables on the force and torque values was investigated separately, by selecting a variable, taking the others as constant and repeating the procedure until determining an optimal set of parameters. The calculations carried out in Mathcad® are provided in Appendix A.

To begin with, the taper angle was one of the major determinant parameters. The screw preload resisting against loosening was linearly proportional to the taper angle as shown in Figure 4.1a. Below a certain taper angle, the screw preload became zero or negative, which meant that resistive torque in the screw threads became zero as can be seen in Figure 4.1b. The proposed design provided higher screw preload and resistive torque values compared to the ITI implant. Note that in this figure and in all of the following figures in this section, **the red curves indicate the results for the proposed dental implant system, whereas blue curves indicate the results for the ITI implant**. Moreover, all the graphical representations present the finalized results obtained by taking optimal values of the parameters presented in Table 4.1, except for the variable parameter.



(a)



(b)

Figure 4.1 The effect of taper angle θ on (a) the screw preload (F_L) and (b) screw torque against loosening (T_{LScrew})

The total tightening and total loosening torques also increased with taper angle (Figure 4.2). The tightening torque was always larger than the loosening torque regardless of the taper angle. Both total tightening and loosening torques were higher for the proposed design compared with the ITI implant.

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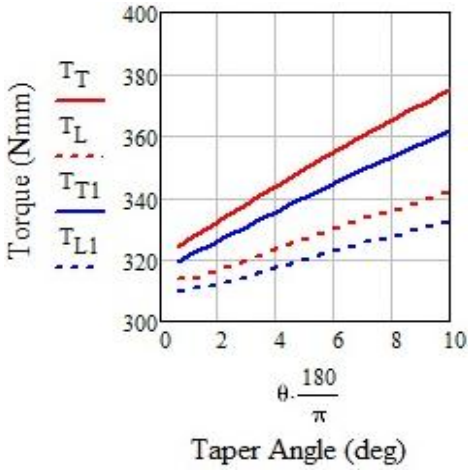


Figure 4.2 Total tightening torque (T_T) and total loosening torque (T_L) varying with taper angle θ

It was stated in Section 3.2.3 that the tapered region in a tapered-with-screw connection dominates the resistive behavior of the connection against loosening due to the interference fit between the conical surfaces. Figure 4.3 justifies this statement. The ratio of the resistive torque at the tapered region to the total resistive torque changed from 1 to 0.84 as taper angle was varied between 0-10º. These high torque ratios demonstrated that a great portion of the total resistive torque was constituted by the torque at the tapered region. Moreover, for low taper angles, the torque ratio converged to 1, since screw preload went to zero as explained above. The torque ratio of the ITI implant was slightly higher.

The efficiency of the connection was defined as the ratio of total loosening torque to the total tightening torque. The efficiency appeared to decrease with increasing taper angles, but remaining above 0.9 for the whole range (0-10º). The efficiency curve for the proposed design was very similar to that of the ITI implant, as illustrated in Figure 4.4.

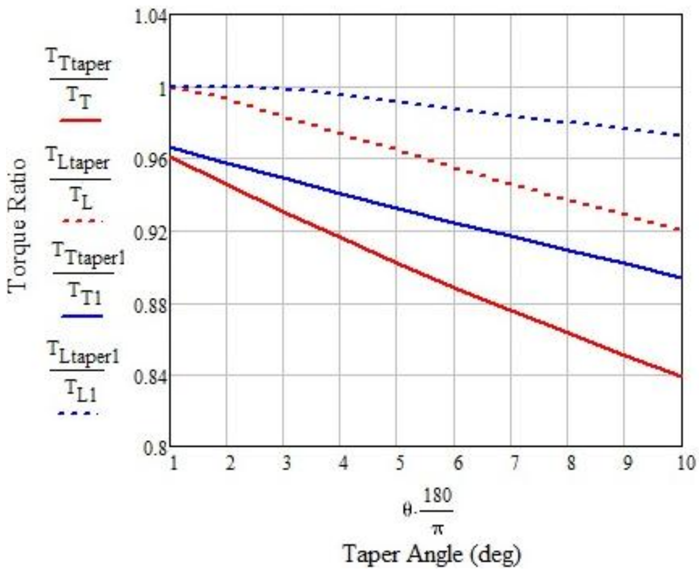


Figure 4.3 The effect of the taper angle θ on the torque ratio (the ratio of the resistive torque at the tapered region to the total resistive torque) for both tightening and loosening

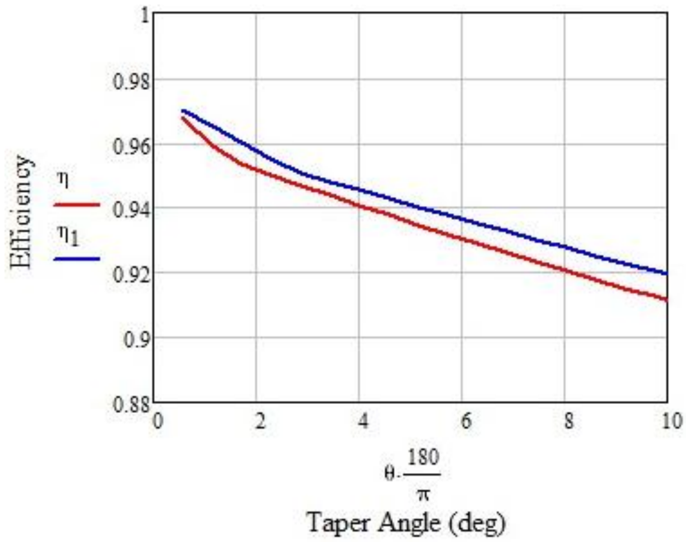
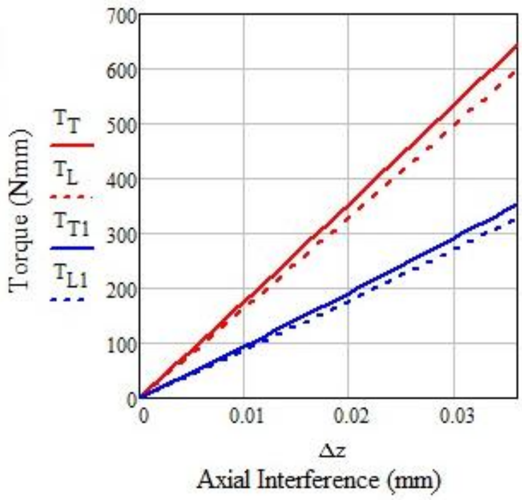
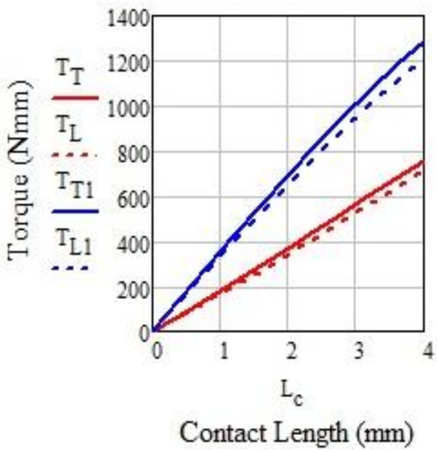


Figure 4.4 The effect of the taper angle θ on the efficiency η of the connection

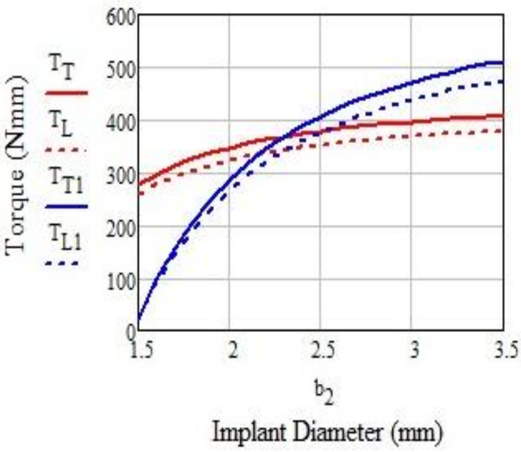
Other than the taper angle, axial interference, contact length and implant diameter were the parameters influential on the results (Figure 4.5). The total tightening and loosening torque values were almost linearly proportional to axial interference and contact length. The torque values increased with both parameters, as expected; because, as the axial interference increases, the contact pressure, therefore, the normal force and frictional forces at the interface also increases. Moreover, increase in the contact length leaded to larger contact area forming larger normal and frictional forces between contacting surfaces. These forces directly affect the resulting torque values. On the other hand, a nonlinear relation was observed between the implant diameter and total torque values. According to Figure 4.4c, higher torque values were required to tighten the screw into the implant for larger implant outer radii.



(a)



(b)



(c)

Figure 4.5 The effect of the (a) axial interference Δz, (b) contact length L_c and (c) implant diameter b_2 on the total tightening torque (T_T) and total loosening torque (T_L)

Finally, the total torque values resisting against loosening and tightening were directly proportional to the preload applied to the screw. The recommended values of tightening torques by the manufacturers of the commercial dental implants is in the range of 100-350 N.mm. Applying a tightening torque in this range generates a screw preload between 40-150 N according to Figure 4.6.

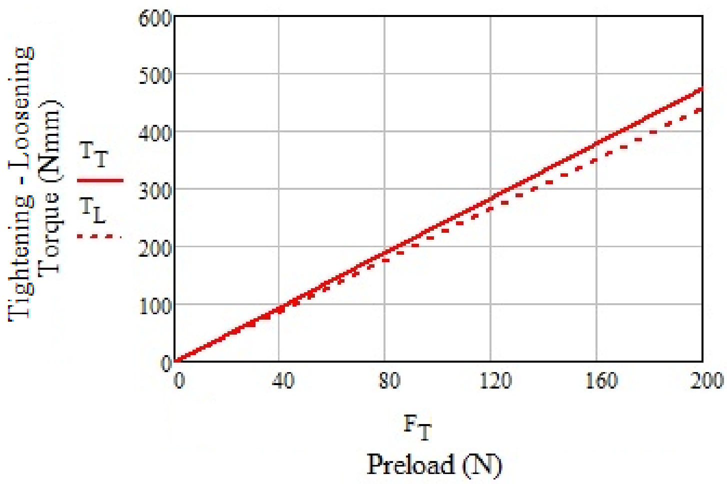


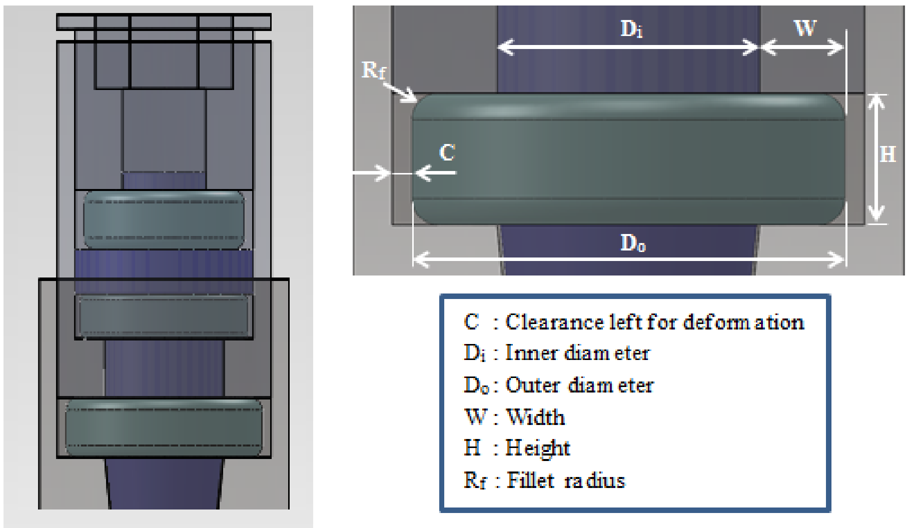
Figure 4.6 The screw preload (F_T) in relation with total tightening (T_T) and loosening torque (T_T)

To sum up, all the aforementioned geometrical parameters were optimized for a tightening torque of about 350 N.mm, by considering relationships presented. In general, the results obtained for the proposed dental implant system using the optimal parameters were quite similar to the results for the ITI implant. After all, it can be concluded that the tapered-with-screw connection that has been implemented in rigid commercial dental implant systems can be easily adapted to the design of the proposed dental implant system. By this way, the problem of abutment screw loosening can be successfully overcome.

4.2 Results of the Axisymmetric Finite Element Analyses

The axisymmetric analysis was the first stage of the finite element analysis procedure followed to assess and optimize the mechanical response of the proposed elastic dental implant system. A systematic methodology was implemented starting with an initial model and updating it until the desired mechanical response, as described in the previous section, was satisfactorily approximated.

The elastic dental implant system contains three resilient components placed at specific positions inside the abutment assembly as shown in Figure 4.7a. At the beginning of each step, throughout the axisymmetric analyses, some geometrical parameters of the elastic components and related dimensions of the relatively rigid parts were modified based on the results obtained at the end of the preceding step. These geometrical parameters are clearly indicated in Figure 4.7b. Actually, alteration of these parameters changed the shape factor (ratio of one loaded area to the total area free to bulge) of each elastic component, thereby, load carrying capacity and deformation characteristics under compressive loading.



(a) **(b)**

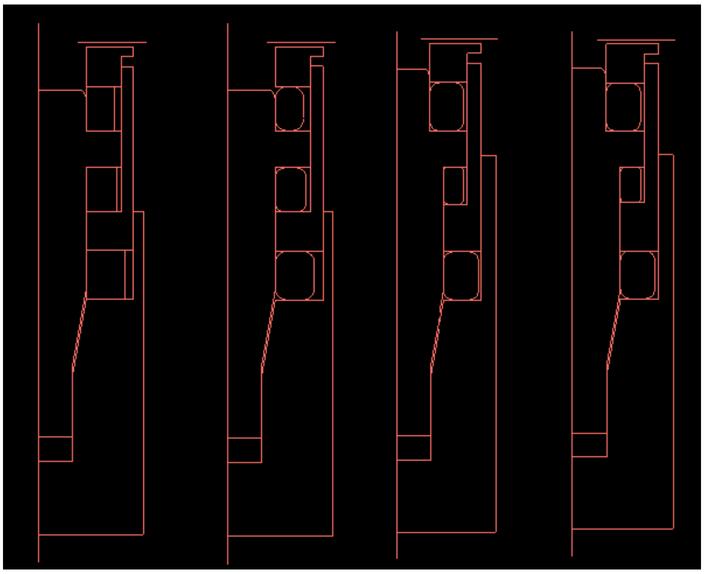
Figure 4.7 (a) The abutment assembly including three elastic components, (b) Geometric design parameters of the elastic components

In Table 4.4, the geometrical parameters belonging to the initial model and the models modified at the end of each step are presented. Model 1 was the initial starting model which was created at the conceptual design stage. In Model 2, four edges of all three elastic components (two edges at upper and lower outer diameter, two edges at upper and lower inner diameter) were filleted with the given fillet radii (R_f). In Model 3, the previous model experienced an extensive change and almost all of the geometrical parameters of the elastic components were modified. Model 4 involved only a slight change in the magnitude of clearance (C) that was left for providing free space required for deformation of the elastic components. However, this slight change was notably influential on the force-displacement characteristics of the system. Model 4 was the last modified design in the axisymmetric analysis stage.

Table 4.4 The values of geometric parameters corresponding to each elastic component in the models modified at each step (the unit of the dimensions is mm)

Axisymmetric Models		C	D_i	D_o	W	H	R_f
Model 1	Elastic-up	0.14	1.94	3.10	0.58	0.9	-
	Elastic-mid	0.09	1.94	3.20	0.63	0.9	-
	Elastic-low	0.17	1.94	3.54	0.80	1	-
Model 2	Elastic-up	0.14	1.94	3.10	0.58	0.9	0.25
	Elastic-mid	0.09	1.94	3.20	0.63	0.9	0.20
	Elastic-low	0.17	1.94	3.54	0.80	1	0.25
Model 3	Elastic-up	0.07	1.4	2.86	0.73	1	0.2
	Elastic-mid	0.07	2	2.86	0.43	0.75	0.1
	Elastic-low	0.07	2	3.46	0.73	1	0.2
Model 4	Elastic-up	0.08	1.4	2.86	0.73	1	0.2
	Elastic-mid	0.08	2	2.86	0.43	0.75	0.1
	Elastic-low	0.08	2	3.46	0.73	1	0.2

In Figure 4.8, all of the four axisymmetric models are shown together. At the first glance, it seems that all the models are very similar to each other. However, the minimal changes applied at each step leaded to critical contributions which approached the mechanical response of the system further towards the desired form.



(a) (b) (c) (d)

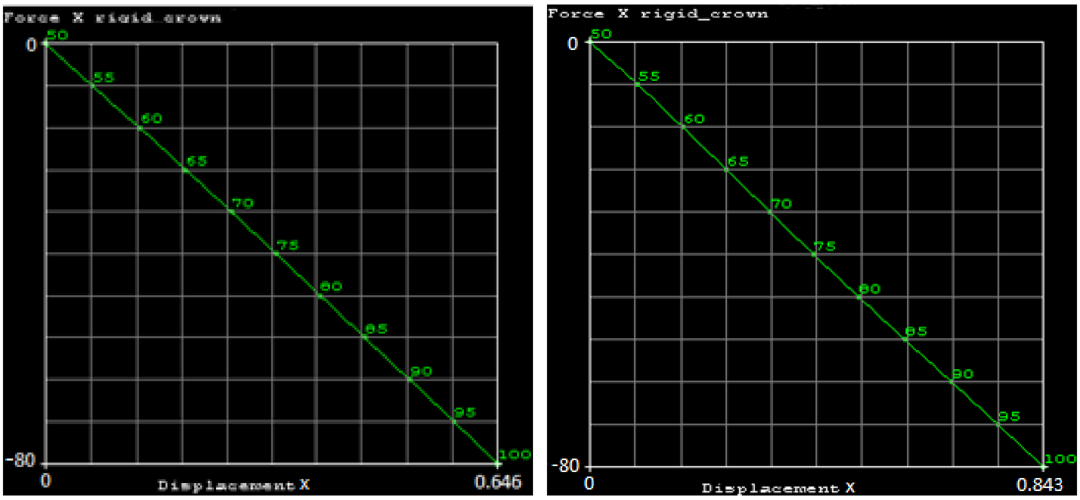
Figure 4.8 The axisymmetric models: (a) Model 1, (b) Model 2, (c) Model 3, (d) Model 4

4.2.1 Iterative Renewal of the Models

Upon updating the geometry of the model in terms of the mentioned parameters, at each step, the four different materials, namely, polyurethane with 95, 90, 85 and 80 Shore A Durometer hardness were defined for the elastic components and separate analyses were carried out for each material specification.

At the outset, Model 1 was analyzed with a material specification of 95 Shore A Durometer hardness. Applying a static load of 80 N resulted that the maximum displacement of the crown were 0.065 mm with a precisely linear behavior (Figure 4.9a) and the maximum von Mises stress appeared to be 17.7 MPa (Figure 4.10a) in the elastic components. When the stress distribution in the elastic components were carefully inspected, it was observed that at the inner upper and lower edges of the elastic components, stress concentration occurred for upper and lower elastic parts, which was an expected outcome since sharp edges are potential sites for stress concentration. Besides, the middle one did not carry load under compression. The stresses occurring in this component were caused by the prestress applied before the functional loading of the system. In addition, the incremental increase of the load caused the prestress of the middle elastic component to decrease gradually. However, no contact separation occurred, as at the top load prestress still remained within the component.

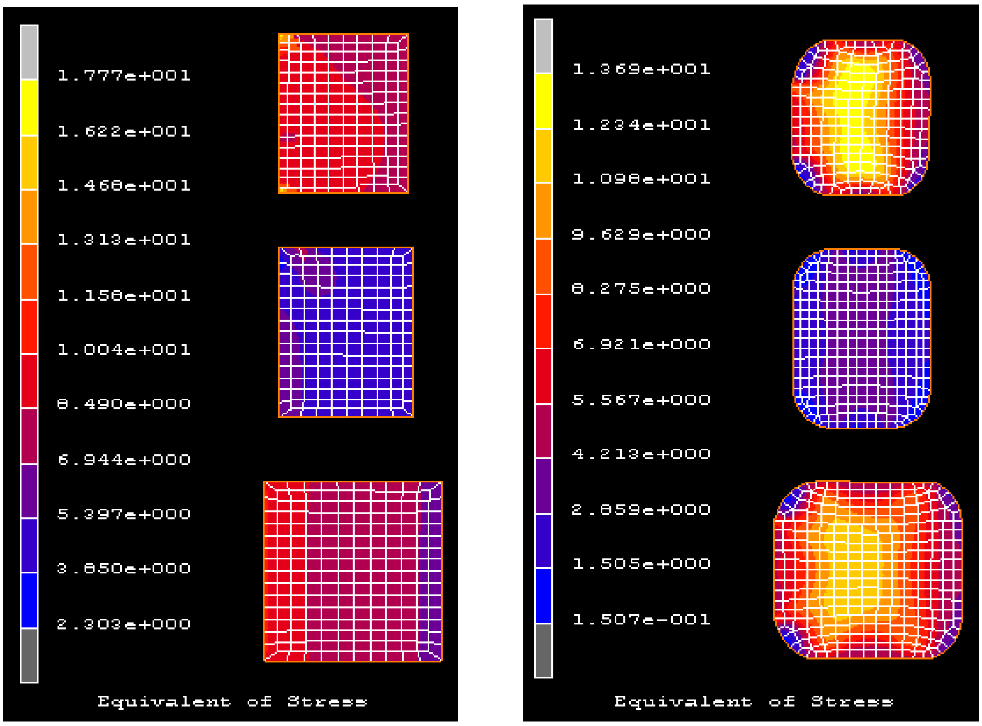
Model 2 was generated to overcome the problem of stress concentration at the sharp edges by creating fillets with a radius of 0.25 mm for upper and lower parts, 0.20 mm for the middle one. Filleting prevented high stresses at the edges and lowered the maximum von Mises stress in the elastic parts to 13.7 MPa. The stress distribution was such that highest stress occurred at the center, decreasing towards the boundaries as shown in Figure 4.10b. The stresses in the upper elastic part was higher than those in the lower one, which was most probably caused by non-uniform load sharing due to different sizes. On the other hand, the force-displacement relation of the crown was again linear but maximum displacement increased to 0.084 mm (Figure 4.9b). Note that the curve remained below the displacement axis in the fourth quadrant since the load is compressive (-80 N).



(a) Model 1 (95A) **(b) Model 2 (95A)**

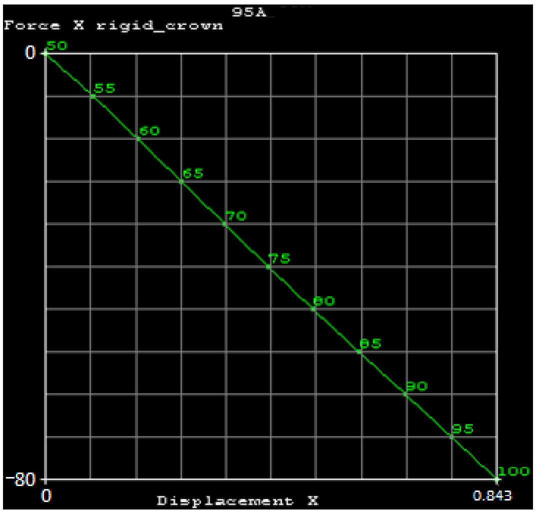
Figure 4.9 The force (N)-displacement (mmx0.1) results for (a) Model 1 and (b) Model 2 with 95 Shore A polyurethane

After examining 95 Shore A polyurethane, the same model was analyzed by defining the three other materials separately for the elastic parts. As shown in Figure 4.11, the maximum displacement of the crown was 0.084 mm, 0.115 mm, 0.180 mm and 0.182 mm corresponding to 95, 90, 85 and 80 Shore A polyurethane, respectively. As the material became compliant, the mobility increased as expected. Moreover, for 85 and 80 Shore A polyurethane, the displacement curve stiffened from a linear path when approaching the point of maximum load. When compared with the target force-displacement curve, which is the reduced curve for a natural tooth as derived in the previous section, the curvature of the path resembled the target behavior but starting at a higher level of load. The displacement values obtained with this model were higher than desired. At the end, this modification provided a better mechanical response regarding both stress distribution in the elastic parts and mobility of the system, however, requiring further enhancement.

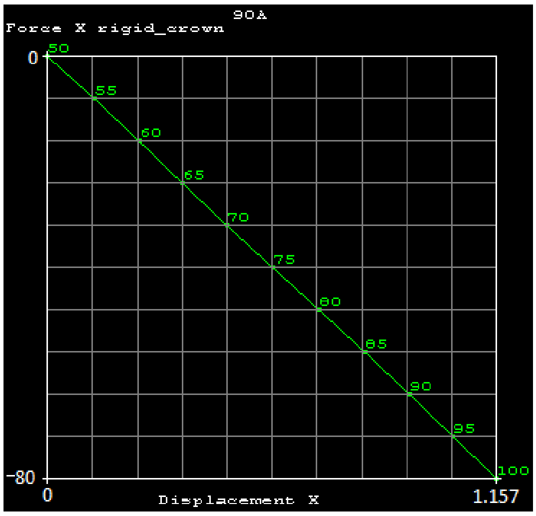


(a) **(b)**

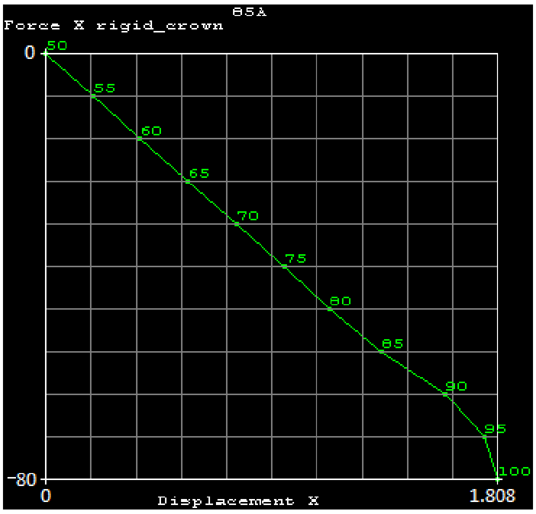
Figure 4.10 The stress distribution (in MPa) in the elastic component in (a) Model 1, (b) Model 2



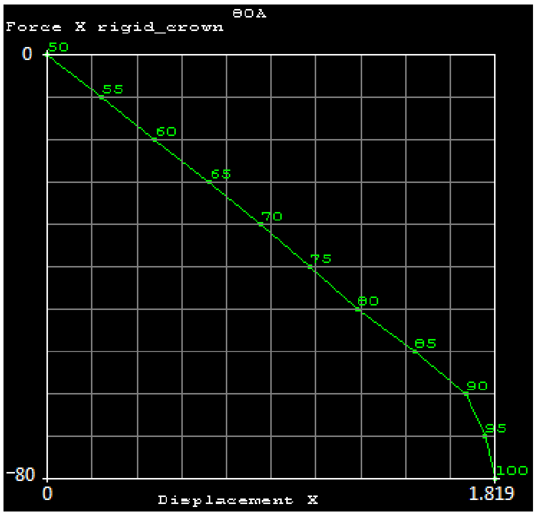
(a) 95A



(b) 90A



(c) 85A

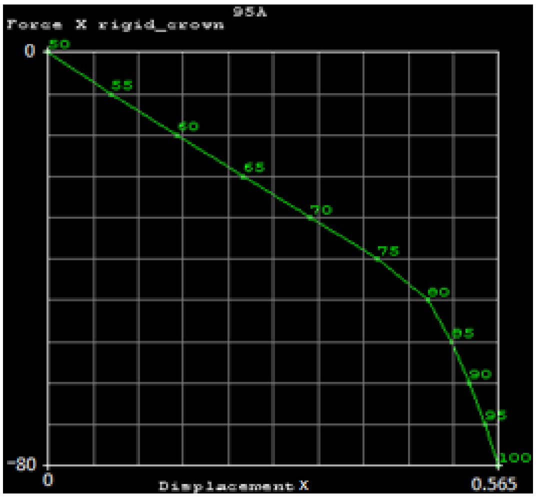


(d) 80A

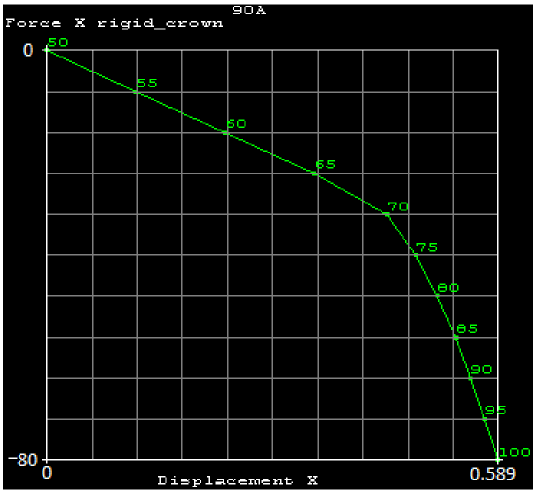
Figure 4.11 The force (N)-displacement (mmx0.1) results for Model 2 with (a) 95, (b) 90, (c) 85 and (d) 80 Shore A polyurethane

Transition from Model 2 to Model 3 was a changeover; because, instead of a single parameter, almost all of the geometrical parameters were modified with the aim of maintaining a more uniform load sharing between the upper and lower elastic components, decreasing the maximum displacement at the previous step and shifting the curvature of the displacement curve to fore points corresponding to lower force values like the case in the reduced curve of a natural tooth. The upper and lower elastic parts in the proposed dental implant system was functioning as two parallel springs bearing the applied total load.

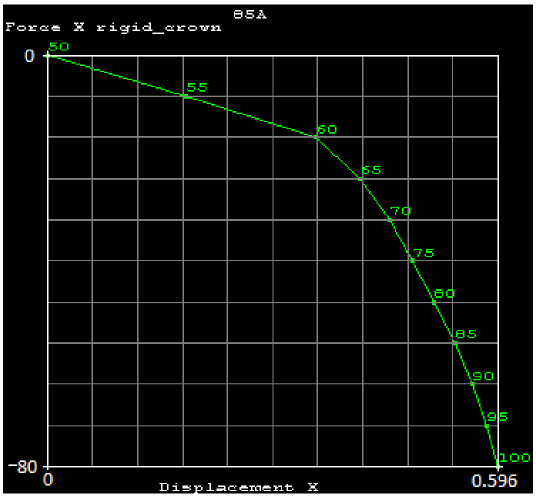
Therefore, at this step the width and height of these components were equalized for sharing the load more evenly. In order to decrease the amount of deformation, the clearance at outer diameters and the radii of the fillets at edges were decreased with certain proportions as seen in Table 4.4. It can be deduced from the force-displacement curves in Figure 4.12 that Model 3 was successful to achieve a similar curvature to the target curve, in other words, the desired nonlinear displacement behavior was approximated to a large extent, except that displacement values at each force increment and the maximum displacement values were remaining below those in the target curve.



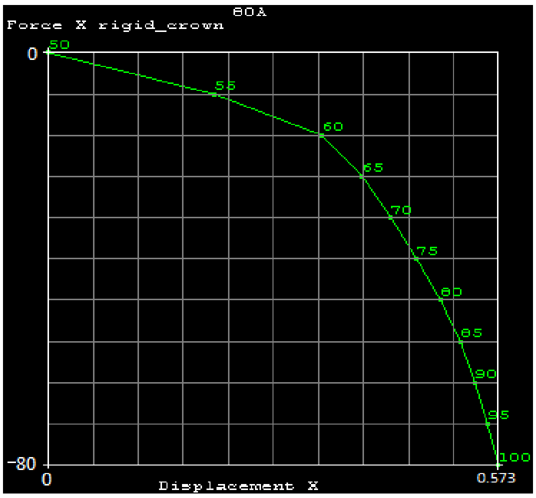
(a) 95A



(b) 90A



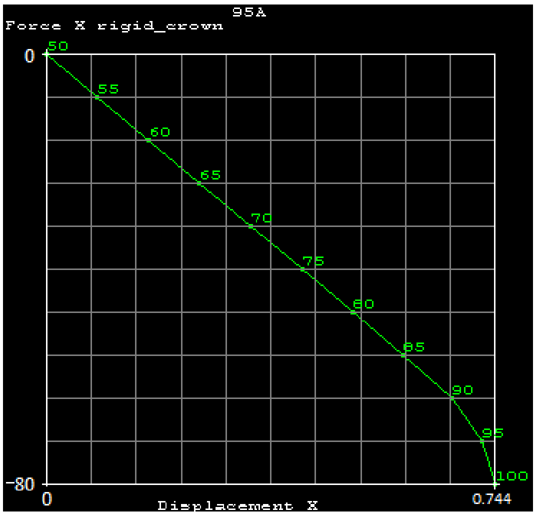
(c) 85A



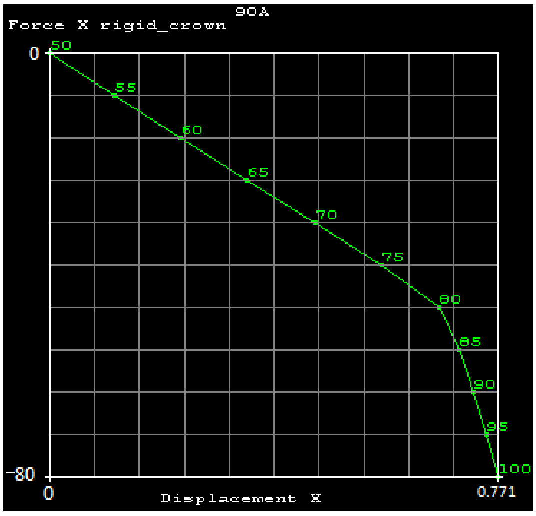
(d) 80A

Figure 4.12 The force (N)-displacement (mmx0.1) results for Model 3 with (a) 95, (b) 90, (c) 85 and (d) 80 Shore A polyurethane

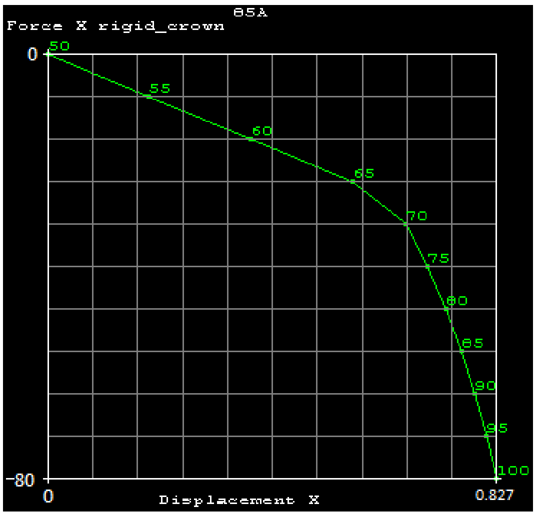
As a single final modification, converting the Model 3 to Model 4, the clearance of all three elastic parts was increased from 0.07 mm to 0.08 mm so as to increase displacements under the same compressive force. In fact, this modification successfully attained the intended purpose as can be perceived by examining the force-displacement curves for Model 4, in Figure 4.13. The displacement values corresponding to each load increment increased when compared to Model 3. This was because the space for deformation was enlarged by the increase in clearance.



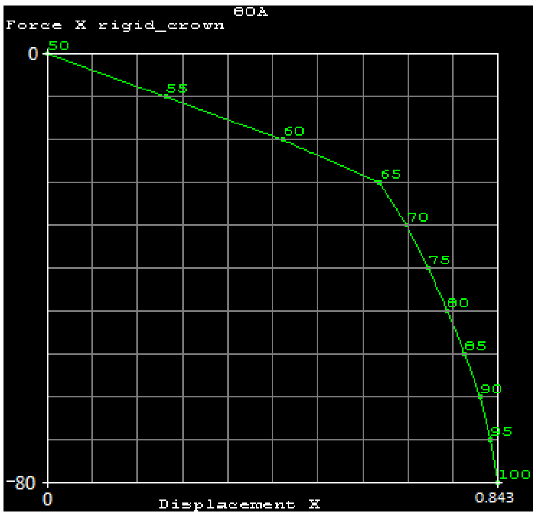
(a) 95A



(b) 90A



(c) 85A



(d) 80A

Figure 4.13 The force (N)-displacement (mmx0.1) results for Model 4 with (a) 95, (b) 90, (c) 85 and (d) 80 Shore A polyurethane

As the material became more resilient from 95 Shore A to 80 Shore A, the curvature tended to shift back more towards the displacement axis and maximum displacement of the crown changed from 0.074 mm to 0.084 mm under the compressive load of 80 N, which was very similar to that in the target curve.

4.2.2 Summary of the Results for All Models

To sum up the results of the axisymmetric analysis, the maximum displacement of the crown and maximum von Mises stress in the elastic components varying with different models and material selection are presented in Table 4.5. By applying the modifications from Model 1 to Model 4, the displacement range of the crown under the given compressive load was tuned to that of the natural case. Moreover, the maximum von Mises stress in the elastic components reduced at each modification of the model in addition to the decreasing pattern corresponding to the decline in the material resiliency from 95 Shore A to 80 Shore A. Figure 4.14 illustrates the progress in achieving reduction of maximum stress.

Table 4.5 The maximum crown displacement and maximum von Mises stress in the elastic components varying with each model and material selection

Models	Max. Displacement of the Crown (mm)				Max. von Mises Stress in the Elastic Components (MPa)			
	95 A	90A	85A	80A	95A	90A	85A	80A
Model 1	0.064	-	-	-	17.8	-	-	-
Model 2	0.084	0.116	0.181	0.182	13.7	11.9	11.2	10.4
Model 3	0.057	0.059	0.060	0.057	10.9	8.9	7.3	6.7
Model 4	0.074	0.077	0.083	0.084	11.4	8.8	7.0	6.8

Considering the entire model, the maximum von Mises stress values and the maximum equivalent elastic strain in the dental implant system also tended to decrease at each step of model updating. However, in contrast to the elastic components, the stress and strain values in the whole model did not decrease with reducing material stiffness of the elastic components. Table 4.6 summarizes the results for related with the entire model.

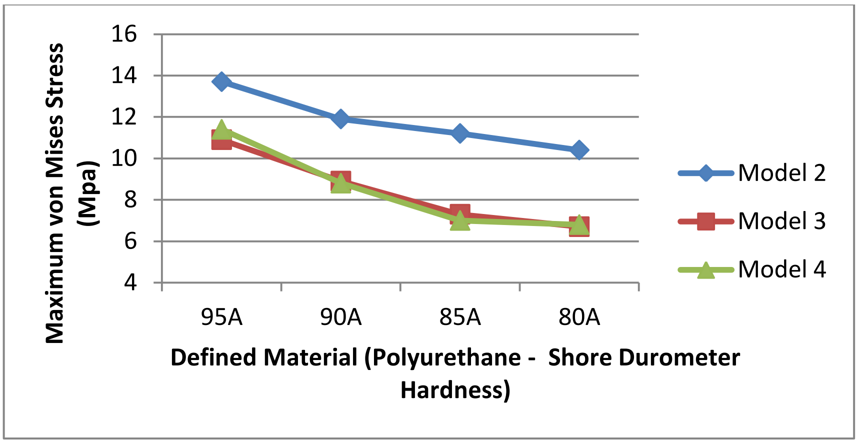


Figure 4.14 The variation of the maximum von Mises stress in the elastic components with respect to material definition

The maximum von Mises stresses in the dental implant system occurred inside the wall of the abutment housing, while maximum elastic strains occurred in the elastic parts. The maximum stresses remained well below the strength limits of both elastic and relatively rigid components for the axial compressive load of 80 N.

Table 4.6 The maximum von Mises stress and the maximum equivalent elastic strain in the overall dental implant system varying with each model and material selection

Models	Max. von Mises Stress in the Dental Implant System (MPa)				Max. Equivalent Elastic Strain in the Dental Implant System (mm/mm)			
	95 A	90A	85A	80A	95A	90A	85A	80A
Model 1	57.4	-	-	-	0.75	-	-	-
Model 2	104.5	111.3	109.3	109.0	1.59	1.64	1.69	1.72
Model 3	69.1	75.5	84.0	86.9	1.02	0.90	0.94	0.96
Model 4	59.7	65.1	69.3	67.6	1.06	0.90	0.82	1.12

4.2.3 Determination of the Optimal Model

Having conducted the axisymmetric analysis of all four models updated by following an iterative procedure, by comparing the curves in Figure 4.9, 4.11, 4.12 and 4.13, the curve in Figure 4.13d was the closest one to the target curve among all the force-displacement curves obtained until this point. It both satisfactorily approximated the nonlinear mobility behavior of a natural tooth and provided a range of displacement under the prescribed load condition almost the same as the natural case. In Figure 4.15, this curve is presented together with the target reduced curve for a natural tooth. The curve for Model 4 captured the general trend in the target curve very closely. The only distinction was that for low level of forces a natural tooth seemed to have higher mobility. One idea could be employing a more resilient material such as 75 Shore A or 65 Shore A polyurethane. However, the strength of polyurethane is lower for the types with lower hardness values, which may pose the risk of failure in case the system becomes subjected to the maximum biting forces (between 690-800 N).

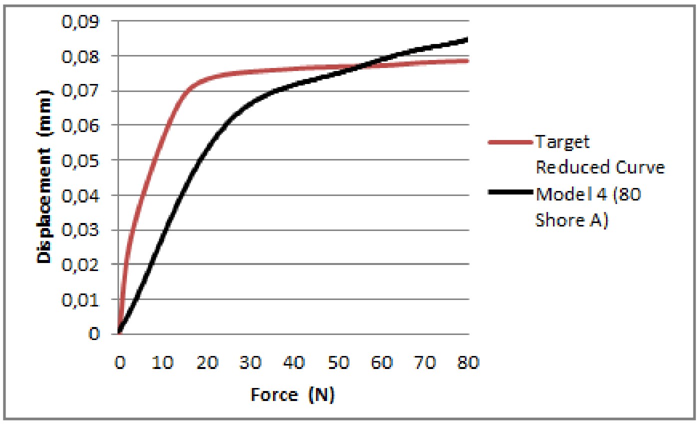
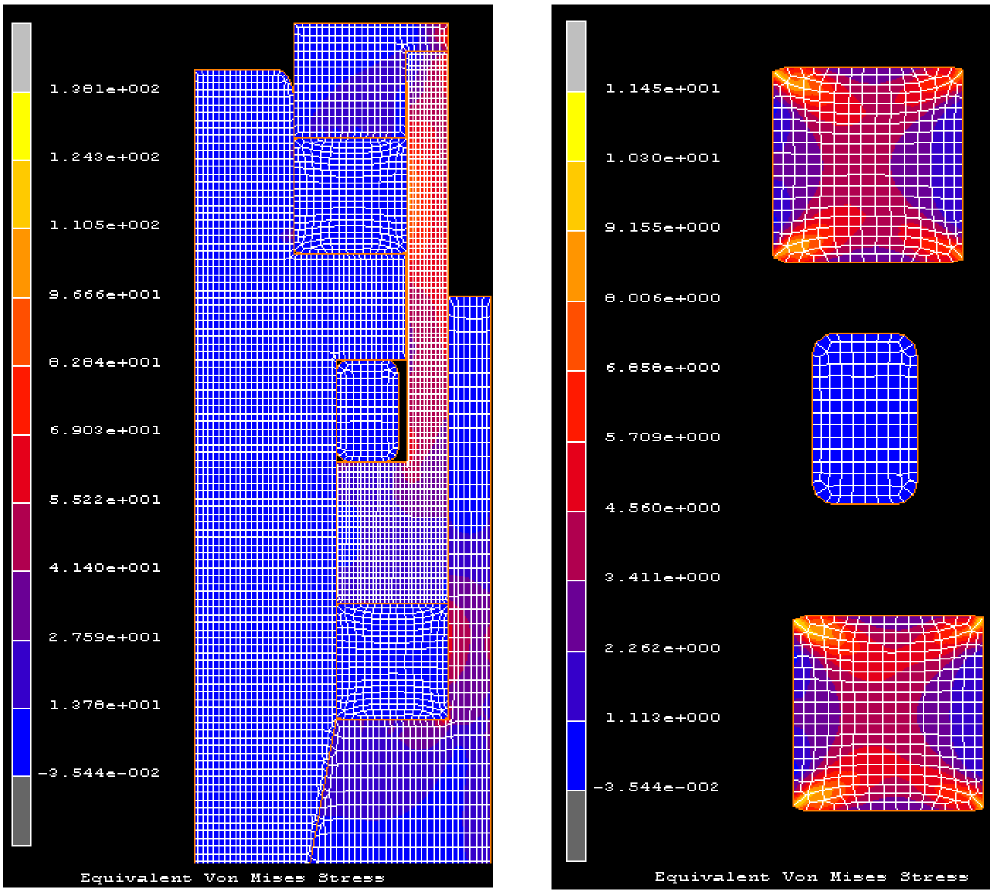


Figure 4.15 The comparison between the force-displacement relationships for Model 4 and for a natural tooth (target reduced curve)

In addition to optimization with respect to the axial mobility behavior, the strength of the whole system under maximum biting forces should also be considered before determination of the final model. For this purpose, Model 4 was further analyzed by applying a maximum vertical biting force of 200 N (reported for an incisor [40]). The equivalent von Mises stress distributions in the whole dental implant system and in the elastic components are shown in Figure 4.16. Accordingly, the highest stresses occurred in the abutment housing at the regions contacting the upper elastic part. The bulging of this elastic part towards the wall of the abutment housing under the action of compressive forces created the contact force between these parts. As the functional compressive load increases, the elastic parts completely fill the gaps and get stiffened. After that the contact forces increase more rapidly, creating large stresses around the contact region. However, the maximum stress both in the relatively rigid parts (138.1 MPa) and the elastic parts (11.4 MPa) were safely below the strength limits. **As a result, according to maximum von Mises stress criterion, the safety factors were about 6.2 and 4.3 for the relatively rigid parts and elastic parts, respectively**.



(a) **(b)**

Figure 4.16 The equivalent von Mises stress (in MPa) distribution under a maximum load of 200 N for Model 4 (80 Shore A polyurethane) (a) in the whole dental implant system, (b) in elastic parts

In most of the previous elastic implant designs, high stress concentration on the internal screw was an important problem that remained unsolved. In the design proposed in this study, under axial forces the internal screw was not subjected to high stresses. This was primarily due to the novel coupled load sharing system incorporated into the design. After the optimization process the two elastic (upper and lower) components shared the total load almost uniformly as can clearly be observed from Figure 4.16b. As a result only half of the total load was transmitted to the abutment screw through the upper elastic part, whereas the other half load is directly transmitted to the implant through the abutment housing and the lower elastic part. This saved the abutment screw from excessive loading.

For Model 4 with the material of elastic components being 80 Shore A polyurethane, the force-displacement relation, under the maximum bite force of 200 N, is presented in Figure 4.17. The maximum displacement at 200 N load was 0.094 mm with a general behavior being nonlinear, biphasic as in the natural case. This nonlinear behavior was achieved by both using nonlinear material modeling and applying certain geometric constraints in the design as described.

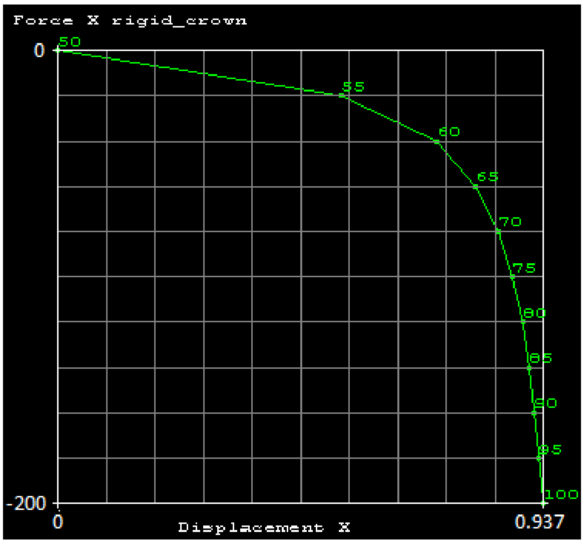
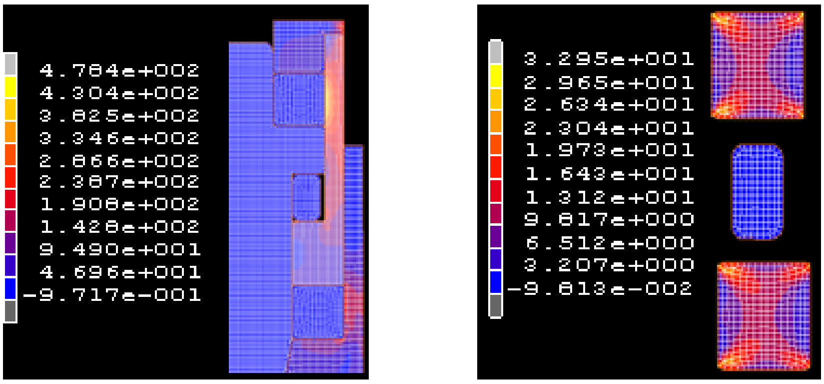


Figure 4.17 The force (N) – displacement (mmx0.1) relationship under maximum bite force of 200 N for Model 4 (with 80 Shore A polyurethane)

To conclude, Model 4 with 80 Shore A polyurethane for the elastic parts achieved the second and third objectives defined precisely in Section 1.7. More explicitly stating, the aim of optimizing the mechanical design parameters of the proposed dental implant system to yield a similar axial mobility as that of an incisor and to withstand functional loading safely without failure of any component was satisfactorily accomplished at the end of axisymmetric analysis stage.

As a final remark, all the work done above was based on the simulation of the behavior of an incisor. It was reported that the molars are less mobile than incisors, that is, greater forces are required for the same amount of displacement for a molar compared with an incisor. The same approach implemented in this study for an incisor can be easily adapted to optimization for a molar. By changing the shape factor of the elastic components, decreasing the clearance left for deformation in horizontal direction and employing a stiffer type of polyurethane, the mobility of a molar could also be reproduced with an elastic dental implant. Just a single finding is included to show that proposed mechanical design could also withstand the higher maximum bite loads that may act on molars. The design in Model 4 was used with stiffer polyurethane (95 Shore A) and a maximum vertical bite load of 720 N (between the reported values for molar [40]) was applied on the system. The maximum stress was 478 MPa in the relatively rigid parts and 33 MPa in the elastic parts (Figure 4.18), both remaining below the compressive strength limits. Thus the proposed design was verified to be applicable also for molars.



(a) (b)

Figure 4.18 The equivalent von Mises stress distribution (in MPa) under a maximum load of 720 N for Model 4 (with 95 Shore A polyurethane) (a) in the whole dental implant system, (b) in the elastic parts

4.3 Results of the Three-Dimensional Finite Element Analyses

The material presented in this section includes the results of the three-dimensional finite element analysis of

- a single dental implant system under axial loading,
- a single dental implant system under lateral loading and
- a dental bridge supported by a rigid and elastic dental implant systems.

4.3.1 Results for Axial Loading

The finite element analysis of the proposed elastic dental implant system under an axial load of 80 N was carried out by defining polyurethane with 80 Shore A Durometer hardness as the material of the elastic components. The results of three-dimensional analyses are compared with the results of axisymmetric analyses obtained in the previous section. The geometric model implemented in the analyses was the final optimal design (Model 4) obtained at the end of the procedure applied in the previous section. The intention was to verify the results of the axisymmetric analysis.

In Figure 4.19, the force-displacement relation of the crown obtained by three-dimensional analysis together with the result of the axisymmetric analysis is presented. For 80 Shore A polyurethane, the resulting displacement curve successfully captured the behavior in the reduced curve for a natural tooth up to a load of 20 N. After that point, the curve remained slightly above the reduced curve. Compared with the displacement results of the axisymmetric analysis, although the magnitudes of displacements were larger in the three-dimensional analysis, the general nonlinear behavior was very similar. The results of both analyses, being in agreement with each other, showed that the elastic dental implant provided a satisfactory approximation to the axial mobility of a natural tooth. Besides, in order to determine the exact mobility behavior of the proposed dental implant system, a uniaxial compression test should be implemented to a prototype of the model. On the other hand, the mobility curve of natural teeth may change for different people. Having simulated the general mobility behavior of a natural tooth, it is possible to optimize the design according to a different mobility behavior by modifying the design parameters.

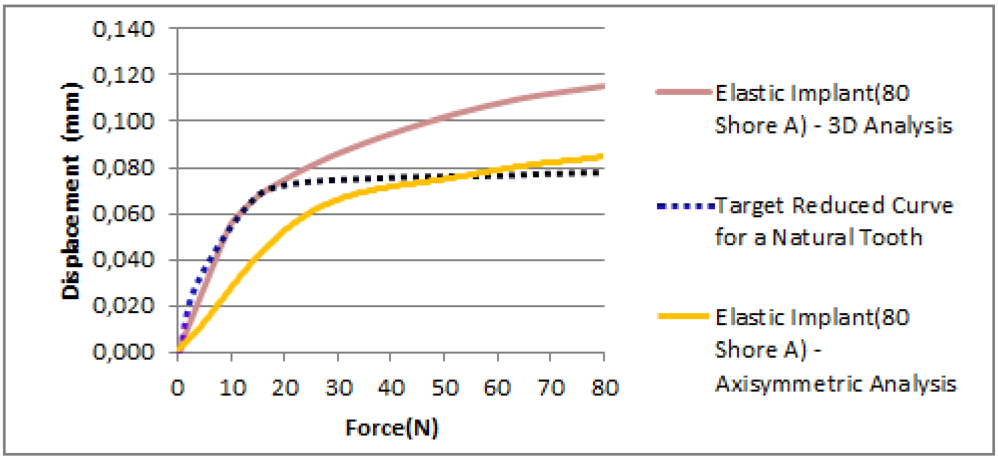


Figure 4.19 The comparison between the force-displacement relationship obtained in the three-dimensional analysis with the result of the axisymmetric analysis

The stress distribution in the three-dimensional model is shown Figure 4.20. Different from the axisymmetric analysis, in the three-dimensional analysis, the preload on the abutment screw was also simulated. By the action of preload, interference occurred at the conical mating region of the screw-implant connection, as intended during design in order to provide resistance against loosening. Due to the interference, the highest stresses occurred in this region. The maximum equivalent von Mises stress was 253.4 MPa around the conical region. The maximum stresses in the abutment assembly (31.9 MPa) were in the wall of the abutment housing, which also revealed in the axisymmetric analysis. The stresses in the elastic parts were comparatively lower (below 8 MPa) than those in the relatively rigid parts. In Table 4.7, maximum stresses in the elastic parts after preloading and after axial loading (80 N) are presented. For the upper and lower elastic parts, the maximum stress after preloading increased with application of the axial load. The stress values in these parts were close to each other, indicating that load sharing was uniform. On the other hand, the initial prestress of the middle elastic part was higher than others, but decreased by application of axial load. In fact, the function of this component was not resisting compressive loading but providing a continuous unloading after removal of the applied loads. As a result, all three elastic parts functioned as intended during the design of the dental implant system.

Table 4.7 The maximum stresses after preloading and after axial loading

Part Name	After Preloading (MPa)	After Axial Loading (MPa)
Upper Elastic Part	1.62	6.12
Middle Elastic Part	4.70	0.73
Lower Elastic Part	1.85	7.38

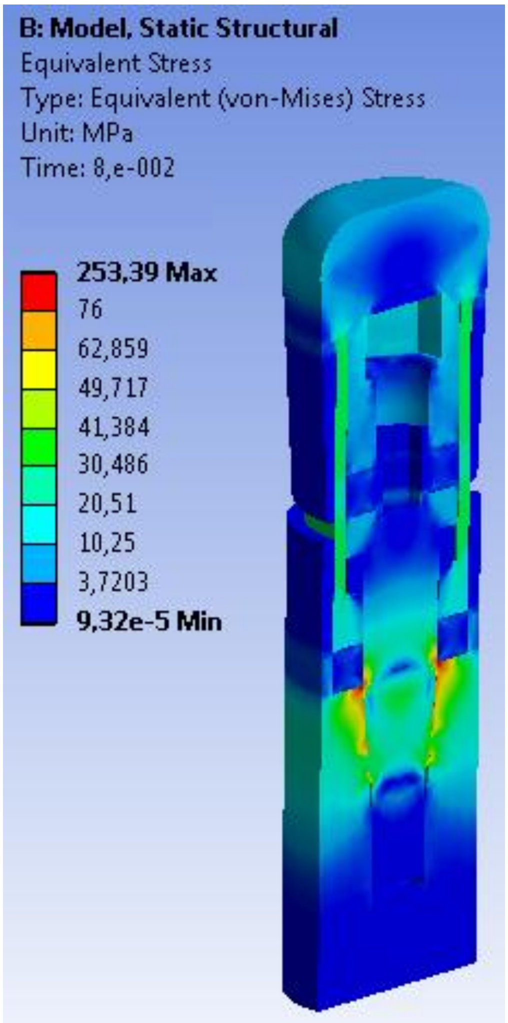


Figure 4.20 The equivalent von Mises stress distribution (in MPa) in 3-D model (with 80 Shore A polyurethane) under an axial load of 80 N

4.3.2 Results for Lateral Loading

The optimal model determined by the analyses under axial loading conditions was subjected to horizontal load of 100 N, simulating the maximum lateral load on a natural tooth. Figure 4.21 illustrates the stress distribution within the dental implant system resulting from lateral loading. The maximum equivalent von Mises stress was 282.2 MPa occurring around the contact region between implant neck and abutment housing. The stress concentration at the neck was caused by the moment which was created due to application of the horizontal force to the tip of the abutment. The elastic parts were not stressed, since they were designed only to carry axial loads. **In conclusion, the system was found to be safe under a maximum lateral load of 100 N with a safety factor of 3.0 according to maximum von Mises stress criterion.** On the other hand, maximum horizontal displacement was 0.0013 mm at the tip of the abutment. This value is by far below the maximum lateral displacement of a natural tooth (0.15 mm [39]) as expected, since the design was made to provide only axial mobility.

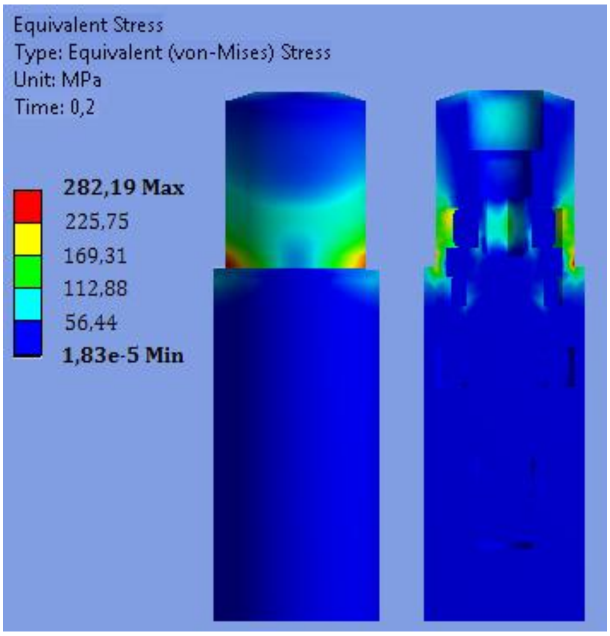


Figure 4.21 The equivalent von Mises stress distribution (MPa) in the optimal model under a lateral load of 100 N (full view and sectional view)

4.3.3 Comparison between Rigid and Elastic Dental Implant in a Bridge

In this section, the results of the finite element analysis, carried out to examine and compare an elastic and a rigid dental implant system supporting a dental bridge prosthesis in combination with a natural tooth, are presented.

In both elastic and rigid implant cases, a vertical load of 40 N was applied on the top surface of the crown bridge. For the bridge supported by a rigid implant (Figure 4.22), high stress concentrations occurred at crown bridge (157.9 MPa), abutment (266.5 MPa) and implant neck (341.3 MPa) as shown in Figure 4.23. These large localized stresses were created by the action of moments formed due to dissimilar mobility between the implant and tooth supports, leading to uneven load sharing between the supports. The vertical displacement of the crown on the natural tooth support was incongruent with that of the crown supported by the implant (Figure 4.24). The crown on top of the natural tooth experienced more than 3 times higher displacements during compressive loading compared with the crown at the implant side (Figure 4.24a). Moreover, the rotational motion of the crown bridge in addition to translation due to dissimilar mobility of the supports can be observed.

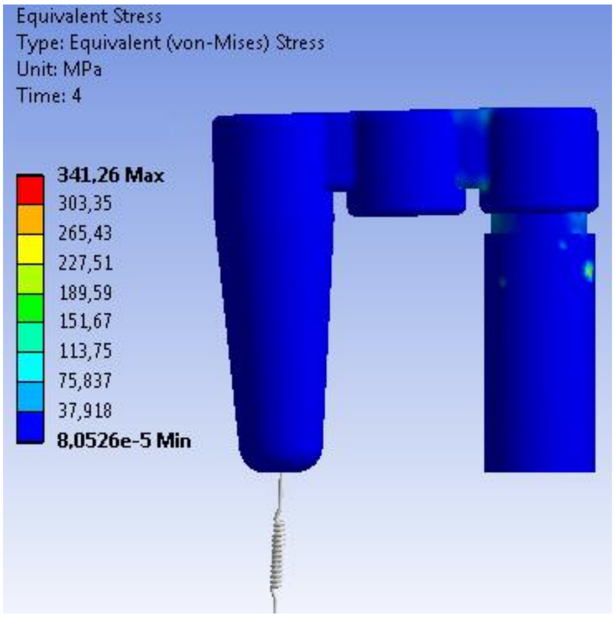
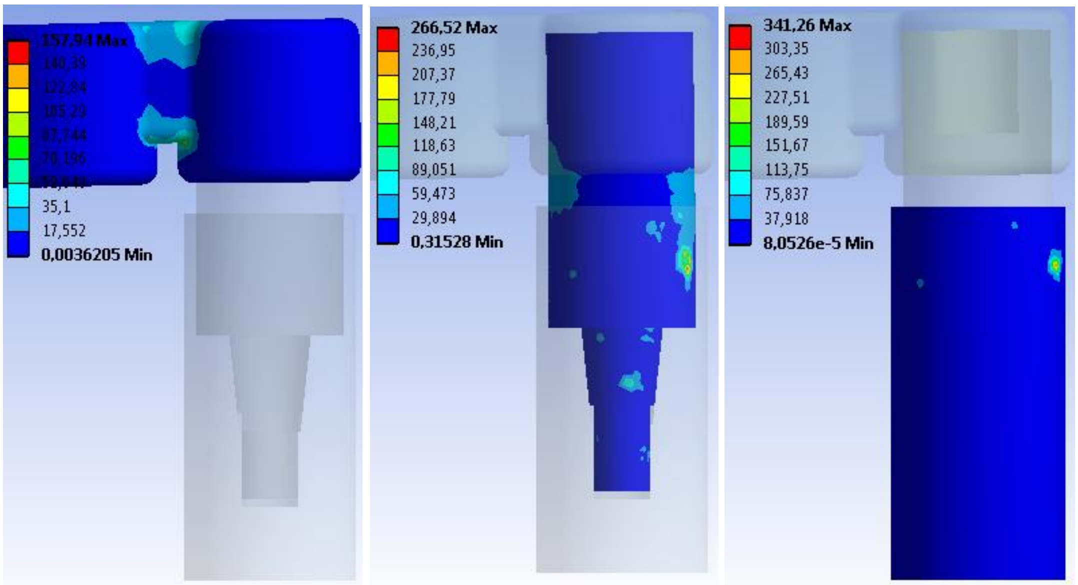
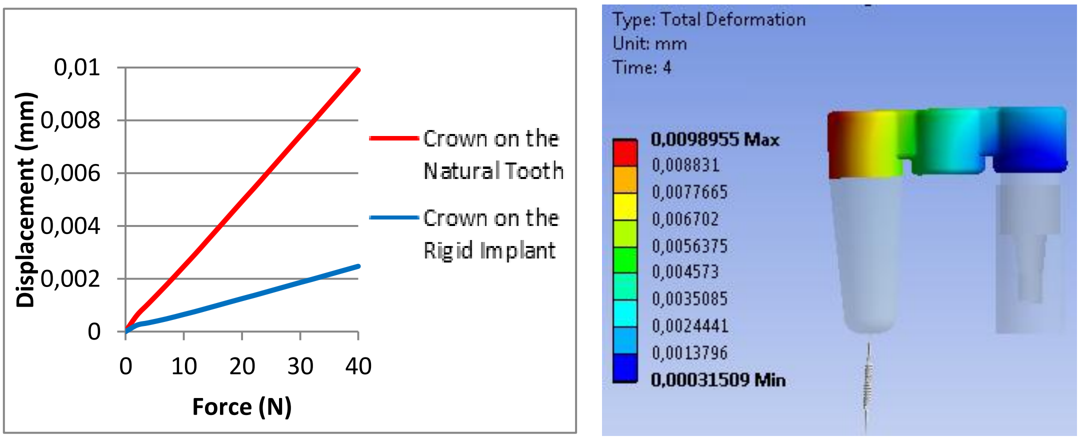


Figure 4.22 The equivalent von Mises stress distribution (in MPa) in the dental bridge supported by a rigid implant



(a) (b) (c)

Figure 4.23 The equivalent von Mises stress distribution (in MPa) in (a) crown bridge, (b) abutment and (c) implant



(a) (b)

Figure 4.24 Comparison of the vertical displacements of the crowns on the natural tooth and the rigid implant support: (a) Force-displacement curve, (b) Displacement distribution at maximum load (40 N)

For the bridge supported by an elastic dental implant, on the other hand, the results were favorably unlike the rigid implant case. Under same compressive loading condition, no localized stress concentrations occurred in the abutment and at the neck of the implant

(Figure 4.25). The maximum stress was 235.5 MPa occurring at the region of conical interference of the implant and internal screw, which was not caused by the vertical load but the screw preload in accordance with the design preference for screw-implant connection.

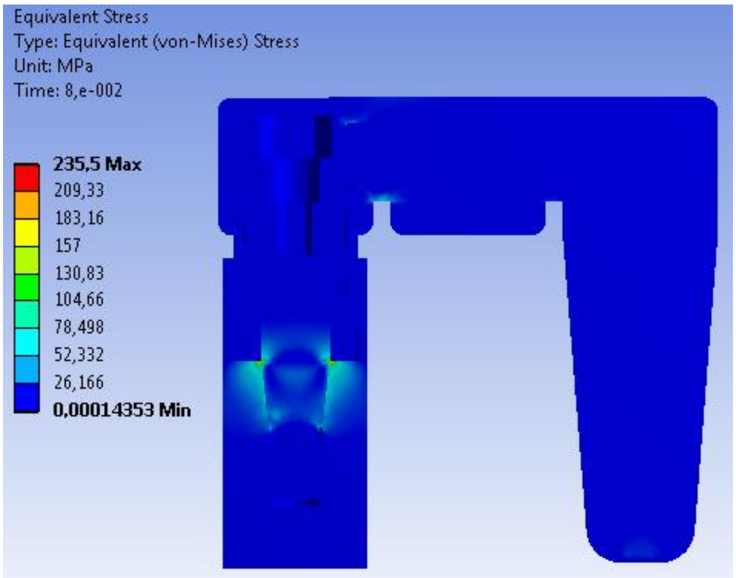


Figure 4.25 The equivalent von Mises stress distribution (MPa) in the dental bridge supported by an elastic dental implant system

In the structure involving an elastic implant, the maximum stresses occurring in crown bridge, abutment and implant reduced prominently compared to prosthesis supported by a rigid implant. By comparing von Mises stress results in Figure 4.23 and Figure 4.26, the substantial decrease of the maximum stress in the crown bridge (from 157.9 to 28.4 MPa) and abutment (from 266.5 to 12.7 MPa) can be observed. Besides, the maximum stress at the implant neck was further diminished to 3.3 MPa from 341.3 MPa. These consequences implied even load sharing among the supports of the crown bridge. The reason for the excessive reduction of the stresses was the achievement of a similar mobility behavior for the elastic dental implant as that of a natural tooth. Considering the force-displacement relationships of the crowns supported by the natural tooth and by the elastic dental implant (in Figure 4.27), a very close conformity draws attention.

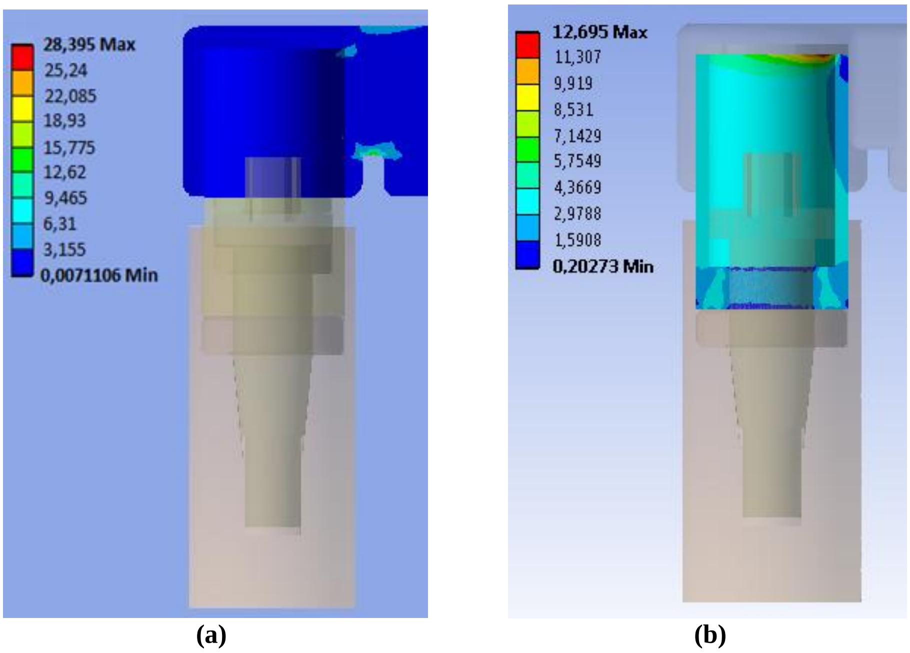


Figure 4.26 The equivalent von Mises stress distribution (MPa) in (a) the crown bridge, (b) the abutment housing

After a compressive load of 30 N, the elastic implant provided slightly higher axial mobility, which was in compliance with the results obtained in the previous sections. In fact, this slight difference was responsible for the local stresses in the crown bridge that can be seen in Figure 4.26a.

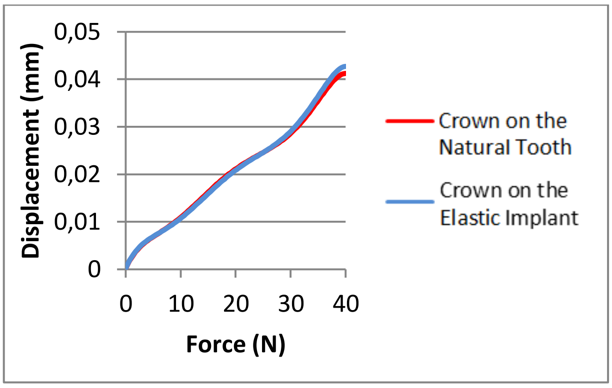


Figure 4.27 Comparison of the vertical displacements of the crowns on the natural tooth and the elastic implant support

To conclude, supporting a dental bridge with an elastic dental implant instead of a rigid one provides significantly improved distribution of load on the whole structure, decreasing the magnitude of the peak stresses. As another important inference drawn from the results, in the dental bridge supported by a rigid implant, in addition to the dissimilar mobility behavior between the crowns on each support, the total displacement of the crown supported by the natural tooth was lower than the same crown in the bridge structure involving an elastic implant. Figure 4.28 helps to better understand the remark. At the maximum load, the displacement of the crown on the natural tooth support was about 0.04 mm for the structure involving elastic implant system, whereas the same crown experienced a displacement of only 0.01 mm when the elastic implant was replaced with a rigid one. This result can be explained by the fact that the limited mobility of crown on the rigid implant constraints the motion of whole bridge.

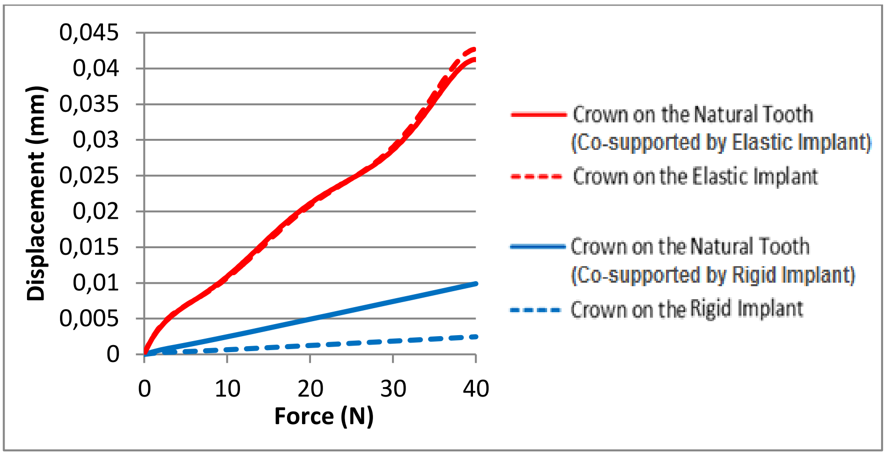


Figure 4.28 Comparison of the vertical displacements of the crowns on the natural tooth and implant supports for the rigid and elastic implant cases

From the clinical point of view, constraining the mobility of the crown on the natural tooth support adversely affects the natural structure of the connection between the tooth and the jawbone. In particular, the periodontal ligament at the interface is prevented to perform its normal mechanical functions as a resilient tissue. As a result, it loses its capabilities and starts to get thinner in time. At the end, the periodontal ligament completely disappears leading to an undesired phenomenon, called ankylosis, which is a term defining the state of rigid bonding of tooth roots to the jawbone.

CHAPTER 5

CONCLUSIONS

5.1 Summary and Conclusions Based on the Methodology and the Results

The problem of tooth loss and the attempts to find out an efficient way to overcome this problem has been a common concern of human beings for ages. However, it has been only a few decades that a standard and permanent clinical dental treatment technique was developed by the modern breakthroughs in the field of dental implantology. Today, dental implant treatment is the most functional and long-lasting way of missing tooth replacement. It is continually becoming more widespread due to its evident superiorities to other alternative conventional methods like dental bridges and removable dentures.

Dental implant is an artificial, usually metallic device surgically placed to the location of lost tooth on the jawbone in order to serve as an artificial dental root to support the new false tooth. Although the current commercial products are quite successful in terms of mechanical strength and long term survival rate, extensive research unceasingly continues to further improve the products and resolve the problems still existing. The investigations are carried out mainly on enhancing the biocompatibility of the implant surface and the mechanical design of the dental implant system.

From the mechanical point of view, all the presently available commercial dental implant systems involve totally rigid mechanisms which enable almost no mobility for the crown anchored by the implant. However, it is not the case for the natural teeth. At the interface between the jawbone and the roots of a living tooth, an elastic, shock-absorbent thin layer, called the periodontal ligament, forms a cushion which provides certain mobility to the natural tooth. The lack of such an elastic interface in the dental implant system or at the implant-jawbone connection leads to some critical complications. Shock loads create large localized stresses in the implant system and the surrounding bone, causing bone resorption and early implant failure. Moreover, the differential mobility between the implant-supported tooth and the neighboring natural teeth results in uneven load sharing during biting or chewing, which is a more prominent problem in the applications of dental bridge prostheses supported by dental implants in combination with natural teeth.

Incorporating flexibility to the design of dental implants to compensate the absence of the periodontal ligament and to simulate its mechanical behavior is a concept attempted by several researchers. Although a number of distinct design ideas, providing significant contribution to the basis of elastic dental implant concept, have been proposed up to the present, no practical and clinically applicable solution has been developed yet.

In this thesis study, it was aimed to achieve a simple, solid and complete mechanical design, including certain material specification, of a dental implant system involving elasticity by incorporating resilient components into the upper structure of the system in order to mimic natural tooth with a periodontal ligament with respect to the mechanical behavior when subjected to functional loading.

The design procedure implemented in the study was based on a systematic and iterative optimization method, developed by introducing a novel approach to the problem. Firstly, the implant (root part of the system) was designed in detail according to the geometric design criteria (like shape, size, thread profile and neck region) in the view of the information obtained from literature. The second step was the design of the abutment (upper structure), where the major contribution was made. The design of the abutment involving resilient components began with a conceptual design stage where a number of different concepts were generated. The proposed concepts were roughly assessed with respect to the design objectives and the best concept was determined prior to proceeding with further optimization via applying an analytical approach and finite element method.

The analytical approach was used to optimize the geometrical design of the connection between the implant and abutment. By implementing taper with screw connection, instead of only screw connection, at implant-abutment interface, the loosening of the abutment screw, which is one of the foregoing complications in implant-supported restorations, was resolved. This type of connection takes the advantage of friction locking action provided by the interference of the contacting conical surfaces of the implant and abutment screw. In the systems involving only screw connection, being subjected to recurrent compressive biting forces causes decrease in the screw preload, thereby, screw loosening in the long run. For a taper with screw connection, on the other hand, the biting forces acting in the direction of abutment insertion increase the degree of engagement in the conical region. As a result, repetitive loads contribute to secure the connection in contrast to causing loosening. It was shown that almost all of the resistive torque against loosening could occur at the conical region instead of threaded section by making an optimal design. Therefore, the problem of screw loosening was eliminated.

By following an iterative static, axisymmetric finite element analysis procedure, the previously determined best design concept for the abutment was optimized. The optimization was based on the design objectives. The optimal model was verified by further three-dimensional FE analyses. The elastic components included in the abutment were modeled to have a nonlinear material behavior involving large strains and large deformations. At the end of the optimization procedure carried out by several FE analyses and accordingly making model updates, polyurethane with 80 Shore A Durometer hardness was determined to be the optimal material for the elastic components in the proposed dental implant system.

The final optimized model involved three resilient components within the upper structure of the proposed dental implant system. In line with the thesis objectives, the optimized model yielded a very similar axial mobility behavior as that of a natural tooth with a periodontal ligament. Moreover, it was verified that the proposed design could withstand functional loading at component level without mechanical failure, especially, considering the internal screw and elastic parts, by providing uniform load sharing on two elastic components and transmitting only about half of the total applied load to the internal screw, while directly transmitting the other half of the load to the implant. The third intermediate elastic component was included to provide continuous unloading after removal of force and to bear possible tensile shock forces that may act on implants supporting fixed overdentures.

It was also verified that the proposed elastic dental implant system, compared to a rigid counterpart, when supporting a dental bridge in combination with a natural tooth, provides significantly improved distribution of load and decreases the magnitude of the peak stresses on the whole structure. This was achieved by replicating the nonlinear, biphasic axial mobility behavior of the natural teeth. The problem of overloading of implants, due to dissimilar mobility with the neighboring teeth, was successfully overcome.

The proposed elastic dental implant design also avoided the problems of soft tissue irritation and moisture contamination of elastic parts by confining the elastic components within a closed system. The top part of the abutment, that is the abutment cap, in the design enabled modifications (like cutting to a specific geometry) before fastening the crown, which is a common practice employed by prosthodontists.

In short, in accordance with the aims of the study, the complete detailed mechanical design of an elastic dental implant system with a reliable mechanical integrity, practical for mass production and appropriate for direct clinical application was satisfactorily achieved. In particular, the proposed elastic dental implant system

1. Overcomes the problem of loosening of the internal screw,
2. Yields a very similar axial mobility behavior as that of a natural tooth with periodontal ligament,
3. Withstands axial and lateral biomechanical loads without failure, and
4. Avoids soft tissue irritation problem.

5.2 Recommendations for Future Research

In spite of the important contributions of the study and the prominent advantages of the proposed design, some limitations were involved in this study. The issues requiring further future investigations are listed below.

1. Since the shock absorbing nature of elastic dental implants has been verified in several previous studies, dynamic analysis for impact loading conditions was not included this study. However, to examine how the proposed design is particularly successful in absorbing impact loads, a dynamic analysis can be carried out. In the dynamic analysis, damping due to viscoelasticity and strain rate dependency may also be considered for the material behavior of the elastic parts.
2. The effects of hysteresis and softening under cyclic loading can be investigated.
3. The proposed model provides mobility only in axial direction. The model can be improved by modifying the design so as to involve also lateral mobility.
4. The resistance of the proposed design to fatigue needs to be revealed.

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APPENDIX A

CALCULATIONS FOR TAPERED SCREW CONNECTION (IN MATHCAD)

A.1 Definitions of the Material and Surface Related Properties

$E := 113.8\ \text{MPa}$	Elastic modulus of implant and abutment
$\mu_s := 0.3 \qquad \mu_k := 0.3$	Static and kinetic friction coefficient

A.2 Definitions of the Geometric Parameters

The assigned values to the geometrical parameters presented below are the optimal values obtained at the end of the iterative calculation procedure.

a) Independent Parameters in the Tapered Region

$\theta := 6\ ^{o}$	Taper angle
$r_{it} := 1\ \text{mm}$	Top radius of taper hole of implant
$r_{ab} := 0.8\ \text{mm}$	Bottom radius of tapered of part of the abutment
$\Delta z := 0.02\ \text{mm}$	Axial displacement of the abutment during tightening
$b_2 := 2.1\ \text{mm}$	Outer radius of the implant

b) Dependent Parameters in the Tapered Region

$L_c := \dfrac{(r_{it} - r_{ab})}{\sin(\theta)} + \dfrac{\Delta z}{\cos(\theta)}$	Contact length of in the tapered section
$\delta := \Delta z \cdot \tan(\theta) = 0.0021$	Interference in the radial direction
$b_1(z) := r_{ab} + (L_c \cdot \cos(\theta) - z) \cdot \tan(\theta)$	Radius of tapered section of the abutment varying along z-axis

c) Parameters in the Threaded Region

$d_{major} := 1.4\ \text{mm}$	Major screw diameter
$p := 0.3\ \text{mm}$	Thread pitch
$n := 1$	(Since single thread is used)
$\alpha := 30°$	Thread angle
$d_m := d_{major} - 0.649519p$	Mean screw diameter
$l := np$	Lead of the screw
$a := 1$	Constant
$\lambda := a \cdot \operatorname{atan}\left(\frac{l}{\pi \cdot d_m}\right)$	Helix angle/Lead angle of the screw

A.3 Definitions of the Equations Related with Force and Torque

a) General Equations for Both Tightening and Loosening of the Screw

$$P_c(z) := \frac{E \cdot \delta \cdot \left(b_2{}^2 - b_1(z)^2\right) \cdot \cos(\theta)}{2 \cdot b_1(z)\, b_2{}^2}$$

Contact pressure due to interference of the tapered section

$$F_N := 2 \cdot \pi \cdot \int_0^{L_c \cdot \cos(\theta)} b_1(z) \cdot P_c(z)\, dz$$

Normal force due to interference fit in the tapered section

By integrating the equation of normal force given above,

$$F_N := \frac{\pi\ E \cdot \Delta z \cdot L_c \cdot \sin(2\theta)}{6 \cdot b_2{}^2} \cdot \left[3 \cdot \left(b_2{}^2 - r_{ab}{}^2\right) - L_c \cdot \sin(\theta) \cdot \left(3r_{ab} + L_c \cdot \sin(\theta)\right)\right]$$

b) Equations for Tightening of the Screw

$$F_T := F_N \cdot \sin(\theta) + \mu_k \cdot F_N \cdot \sin(\lambda) \cdot \cos(\theta)$$

Screw preload = Vertical resisting force in the tapered section during tightening

$$T_{Tscrew} := \frac{F_T \cdot d_m}{2} \cdot \left(\frac{l + \pi \cdot \mu_k \cdot d_m \cdot \sec(\alpha)}{\pi \cdot d_m - \mu_k \cdot l \cdot \sec(\alpha)} \right)$$

Resistive torque in the screw threads against tightening
(from the power screw formula for raising a load)

Since the horizontal component of the friction force ($\mu N\cos\lambda$) in the tapered section resists the tightening torque, the required torque to overcome this friction:

$$T_{Ttaper} := \left(2\pi \, \mu_k \cdot \int_0^{L_c \cdot \cos(\theta)} b_1(z)^2 \cdot P_c(z) \, dz \right) \cos(\lambda)$$

Resistive torque in the tapered part against tightening

$$T_T := T_{Tscrew} + T_{Ttaper}$$

Total resistive torque during tightening

c) Equations for Loosening of the Screw

$$F_L := F_N \cdot \sin(\theta) - \mu_s \cdot F_N \cdot \sin(\lambda) \cdot \cos(\theta)$$

Screw preload that resists loosening

$$T_{Lscrew} := \frac{F_L \cdot d_m}{2} \cdot \left[\frac{(\pi \cdot \mu_s \cdot d_m \cdot \sec(\alpha) - l)}{(\pi \cdot d_m + \mu_s \cdot l \cdot \sec(\alpha))} \right] \cdot \Phi(F_L)$$

Resistive torque in the screw threads
(from the power screw formula for lowering a load)

where the Heaviside step function: $\Phi(F_L) = 1$ if $F_L > 0$

$\Phi(F_L) = 0$ if $F_L < 0$

$$T_L := T_{Lscrew} + T_{Ltaper}$$

Total resistive torque against loosening

d) The efficiency of the tapered screw connection:

$$\eta := \frac{T_L}{T_T}$$

A.4 Parametric Definitions of the Equations

a) General Equations for Both Tightening and Loosening of the Screw

$$\delta(\Delta z, \theta) := \Delta z \cdot \tan(\theta)$$

$$L_c(\Delta z, r_{it}, r_{ab}, \theta) := \frac{(r_{it} - r_{ab})}{\sin(\theta)} + \frac{\Delta z}{\cos(\theta)}$$

$$b_1(z, L_c, r_{ab}, \theta) := r_{ab} + (L_c \cdot \cos(\theta) - z) \cdot \tan(\theta)$$

$$F_N(L_c, E, \Delta z, b_2, \theta, r_{ab}) := \frac{\pi\ E \cdot \Delta z \cdot L_c \cdot \sin(2\theta)}{6 \cdot b_2{}^2} \cdot \left[3 \cdot \left(b_2{}^2 - r_{ab}{}^2\right) - L_c \cdot \sin(\theta) \cdot \left(3 r_{ab} + L_c \cdot \sin(\theta)\right)\right]$$

b) Equations for Tightening of the Screw

$$F_T(F_N, \theta, \mu_k, \lambda) := F_N \cdot \sin(\theta) + \mu_k \cdot F_N \cdot \sin(\lambda) \cos(\theta)$$

$$T_{Tscrew}(F_T, d_m, \mu_k, \alpha) := \frac{F_T \cdot d_m}{2} \cdot \left(\frac{l + \pi \cdot \mu_k \cdot d_m \cdot \sec(\alpha)}{\pi \cdot d_m - \mu_k \cdot l \cdot \sec(\alpha)}\right)$$

$$T_{Ttaper}\left(\Delta z, \mu_k, L_c, E, \theta, \lambda, b_2, r_{ab}\right) := \frac{\pi\,\mu_k\cdot E\cdot \Delta z\cdot L_c\cdot \sin(2\cdot\theta)\cos(\lambda)}{8\cdot {b_2}^2}\left[L_c\cdot\sin(\theta)\cdot\left[2\cdot\left({b_2}^2 - 3{r_{ab}}^2\right) - L_c\cdot\sin(\theta)\cdot\left(4r_{ab} + L_c\cdot\sin(\theta)\right)\right] + 4r_{ab}\cdot\left({b_2}^2 - {r_{ab}}^2\right)\right]$$

$$T_T\left(T_{Tscrew}, T_{Ttaper}\right) := T_{Tscrew} + T_{Ttaper}$$

c) Equations for Loosening of the Screw

$$F_L\left(F_N, \theta, \mu_s, \lambda\right) := F_N\sin(\theta) - \mu_s\cdot F_N\sin(\lambda)\cdot\cos(\theta)$$

$$T_{Lscrew}\left(F_L, d_m, \mu_s, \alpha\right) := \frac{F_L\cdot d_m}{2}\cdot\left[\frac{\left(\pi\cdot\mu_s\cdot d_m\cdot\sec(\alpha) - l\right)}{\left(\pi\cdot d_m + \mu_s\cdot l\cdot\sec(\alpha)\right)}\right]\cdot\Phi\left(F_L\right)$$

$$T_{Ltaper}\left(\Delta z, \mu_s, L_c, E, \theta, \lambda, b_2, r_{ab}\right) := \frac{\pi\,\mu_s\cdot E\cdot \Delta z\cdot L_c\cdot \sin(2\cdot\theta)\cos(\lambda)}{8\cdot {b_2}^2}\left[L_c\cdot\sin(\theta)\cdot\left[2\cdot\left({b_2}^2 - 3{r_{ab}}^2\right) - L_c\cdot\sin(\theta)\cdot\left(4r_{ab} + L_c\cdot\sin(\theta)\right)\right] + 4r_{ab}\cdot\left({b_2}^2 - {r_{ab}}^2\right)\right]$$

$$T_L\left(T_{Lscrew}, T_{Ltaper}\right) := T_{Lscrew} + T_{Ltaper}$$

d) The efficiency of the tapered screw connection:

$$\eta\left(T_T, T_L\right) := \frac{T_L}{T_T}$$

APPENDIX B

GLOSSARY OF DENTAL TERMS

The definitions of the terms included in this section were adapted from the Glossary of Prosthodontic Terms [20] and Cambridge Dictionaries Online [178].

abutment: portion of a dental implant that serves to support and/or retain a prosthesis

acrylic resin: 1. pertaining to polymers of acrylic acid, methacrylic acid, or acrylonitrile; for example, acrylic fibers or acrylic resins, **2.** any of a group of thermoplastic resins made by polymerizing esters of acrylic or methylmethacrylate acids

alveolar bone: the bony portion of the mandible or maxillae in which the roots of the teeth are held by fibers of the periodontal ligament

alveolus: one of the cavities or sockets within the alveolar process of the maxillae or mandible in which the attachment complex held the root of a tooth after the tooth's removal

ankylosis: the situation when a tooth root becomes fused to jawbone due to loss of periodontal ligament

antagonist tooth: the opposing tooth in the other jaw

artificial crown: a metal, plastic, or ceramic restoration that covers three or more axial surfaces and the occlusal surface or incisal edge of a tooth

biocompatibility: being capable of existing in harmony with the surrounding biologic environment

biting: the act of incising or crushing between the teeth

biting force: see OCCLUSAL FORCE

bone: the hard portion of the connective tissue which constitutes the majority of the skeleton; it consists of an inorganic or mineral component and an organic component (the matrix and cells)

bone regeneration: renewal or restoration of a bone after injury or as a normal process

bone resorption: the loss of bone by physiologic or pathologic processes

bridge: see FIXED DENTAL PROSTHESIS

bruxism: 1. the parafunctional grinding of teeth, **2.** an oral habit consisting of involuntary rhythmic or spasmodic nonfunctional gnashing, grinding, or clenching of teeth, in other than chewing ovements of the mandible, which may lead to occlusal trauma—called also tooth grinding, occlusal neurosis

cancellous bone: the reticular, spongy or lattice-like portion of the bone; the spongy bone tissue located in the medulla of the bone; this bone is composed of a variable trabecular network containing interstitial tissue that may be hematopoietic

cementum: the thin calcified tissue of ectomesenchymal origin that covers the root of a tooth

chewing: crushing food into smaller, softer pieces with the teeth so that it can be swallowed

chewing force: see MASTICATORY FORCE

chewing cycle: see MASTICATORY CYCLE

clinical: of or related to or conducted in or as if within a clinic

collagen: a protein found especially in the joints

complete denture: a removable dental prosthesis that replaces the entire dentition and associated structures of the maxillae or mandible; called a complete removable dental prosthesis

connective tissue: a tissue of mesodermal origin rich in interlacing processes that supports or binds together other tissues

cortical bone: the peripheral layer of compact osseous tissue

crest: a ridge or prominence on a part of a body; in dentistry, the most coronal portion of the alveolar process

crooked teeth: teeth that are not straight; but have sharp bends

crown: 1. the highest part, as the topmost part of tooth; the summit; that portion of a tooth occlusal to the dentinoenamel junction or an artificial substitute for this, **2.** an artificial replacement that restores missing tooth structure by surrounding part or all of the remaining structure with a material such as cast metal, porcelain, or a combination of materials such as metal and porcelain

dental anatomy: a field of biology and medical science which deals with the structure and functions of teeth

dental arch: the composite structure of the natural teeth and alveolar bone

dental biomechanics: the relationship between the biologic behavior of oral structures and the physical influence of a dental restoration

dental bridge: see FIXED DENTAL PROSTHESIS

dental caries: a dental disease causing the destruction of enamel, dentin and/or cementum

dental implant: 1. a prosthetic device made of alloplastic material(s) implanted into the oral tissues beneath the mucosal or/and periosteal layer, and on/or within the bone to provide retention and support for a fixed or removable dental prosthesis; a substance that is placed into or/and upon the jaw bone to support a fixed or removable dental prosthesis, **2.** the portion of an implant that provides support for the dental implant abutment(s) through adaptation upon (eposteal), within (endosteal), or through (transosteal) the bone

dental prosthesis: an artificial replacement (prosthesis) of one or more teeth (up to the entire dentition in either arch) and associated dental/alveolar structures. Dental prostheses usually are subcategorized as either fixed dental prostheses or removable dental prostheses

dentate: having teeth or pointed conical projections

dentin: a calcareous material similar to but harder and denser than bone that comprises the principle mass of the tooth

dentition: the teeth in the dental arch

denture: an artificial substitute for missing natural teeth and adjacent tissues

edentulism: the state of being edentulous; without natural teeth

edentulous: without teeth, lacking teeth

elastomer: a polymer whose glass transition temperature is below its service temperature (usually room temperature). These materials are characterized by low stiffness and extremely large elastic strains

enamel: in dentistry, the hard, thin, translucent layer of calcified substance that envelopes and protects the dentin of the coronal aspect of the tooth; it is the hardest substance in the body

endosseous implant: see ENDOSTEAL DENTAL IMPLANT

endosteal dental implant: a device placed into the alveolar and/or basal bone of the mandible or maxilla and transecting only one cortical plate

exogenous: coming from or produced outside an organism or cell

fibrous: composed of or containing fibers

fixed bridge: see FIXED DENTAL PROSTHESIS

fixed partial denture: see FIXED DENTAL PROSTHESIS

fixed dental prosthesis: any dental prosthesis that is luted, screwed or mechanically attached or otherwise securely retained to natural teeth, tooth roots, and/or dental implant abutments that furnish the primary support for the dental prosthesis. This may include replacement of one to sixteen teeth in each dental arch.

gingiva: the fibrous investing tissue, covered by epithelium, which immediately surrounds a tooth and is contiguous with its periodontal membrane and with the mucosal tissues of the mouth

gum: either of the two areas of firm pink flesh inside the mouth which cover the bones into which the teeth are fixed

host bone: recipient bone

hydroxyapatite: a composition of calcium and phosphate in physiologic ratios to provide a dense, non-resorbable, biocompatible ceramic used for dental implants and residual ridge augmentation

implant: any object or material, such as an alloplastic substance or other tissue, which is partially or completely inserted or grafted into the body for therapeutic, diagnostic, prosthetic, or experimental purposes

implant dentistry: the selection, planning, development, placement, and maintenance of estoration(s) using dental implants

implantation: grafting or inserting a material such as an alloplastic substance, an encapsulated drug, or tissue into the body of a recipient

implantology: a term historically conceived as the study or science of placing and restoring dental implant

implant surgery: the phase of implant dentistry concerning the selection, planning, and placement of the implant body and abutment

incisor: one of the sharp teeth at the front of the mouth which cut food during biting

intraoral: within the mouth

jaw: see JAWBONE

jawbone: the bony structure bearing the teeth—see also MANDIBLE, MAXILLA

mandible: the lower jawbone

mastication: the process of chewing food for swallowing and digestion

masticating cycles: the patterns of mandibular movements formed during the chewing of food

masticatory force: the force applied by the muscles of mastication during chewing

masticatory cycle: a three dimensional representation of mandibular movement produced during the chewing of food

masticatory movements: mandibular movements used for chewing food

maxilla: the irregularly shaped bone that, with its contralateral maxilla, forms the upper jaw. It assists in the formation of the orbit, the nasal cavity, and the hard palate; it contains the maxillary teeth

molar: one of the large teeth at the back of the mouth in humans and some other animals used for crushing and chewing food

mucosa: a mucous membrane comprised of epithelium, basement membrane, and lamina propria

mucosal membrane: the thin skin that covers the inside surface of parts of the body such as the nose and mouth and produces mucus to protect them

occlude: **1.** to bring together; to shut, **2.** to bring or close the mandibular teeth into contact with the maxillary teeth

occlusal: pertaining to the masticatory surfaces of the posterior teeth, prostheses, or occlusion rims

occlusal analysis: an examination of the occlusion in which the interocclusal relations of mounted casts are evaluated

occlusal balance: a condition in which there are simultaneous contacts of opposing teeth or tooth analogues (i.e., occlusion rims) on both sides of the opposing dental arches during eccentric movements within the functional range

occlusal force: the result of muscular force applied on opposing teeth; the force created by the dynamic action of the muscles during the physiologic act of mastication; the result of muscular activity applied to opposing teeth

occlusion: **1.** the act or process of closure or of being closed or shut off, **2.** the static relationship between the incising or masticating surfaces of the maxillary or mandibular teeth or tooth analogues

osseointegration: **1.** the apparent direct attachment or connection of osseous tissue to an inert, alloplastic material without intervening connective tissue, **2.** the process and resultant apparent direct connection of an exogenous materials' surface and the host bone tissues, without intervening fibrous connective tissue present

osseous: bony

osteoblast: the cells of bone that make new bone

osteoclast: the cells responsible from absorbing bone

overdenture: any removable dental prosthesis that covers and rests on one or more remaining natural teeth, the roots of natural teeth, and/or dental implants; a dental prosthesis that covers and is partially supported by natural teeth, natural tooth roots, and/or dental implants

parafunctional: disordered or perverted function

partial denture: a removable dental prosthesis or a fixed dental prosthesis that restores one or more but not all of the natural teeth and/or associated parts and may be supported in part or whole by natural teeth, dental implant supported crowns, dental implant abutment(s), or other fixed dental prostheses and/or the oral mucosa; usage: a partial denture can be described as a fixed dental prosthesis or removable dental prosthesis based on the patient's capability to remove or not remove the prosthesis

periodontal: pertaining to or occurring around a tooth

periodontal ligament: a collagenous connective tissue, completely covering tooth roots and separating tooth from the alveolar bone

physioelasticity: the unique biologic quality of being capable of changing and resuming size

physiologic: **1.** characteristic of or conforming to the innate function of a tissue or organ, **2.** pertaining to organic processes or to functions in an organism or in any of its parts, **3.** the opposite of pathologic

pontic: an artificial tooth on a fixed dental prosthesis that replaces a missing natural tooth, restores its function, and usually fills the space previously occupied by the clinical crown

porcelain: a ceramic material formed of infusible elements joined by lower fusing materials. Most dental porcelains are glasses and are used in the fabrication of teeth for dentures, pontics and facings, metal ceramic restorations including fixed dental prostheses, as well as all-ceramic restorations such as crowns, laminate veneers, inlays, onlays, and other restorations

premolar: one of the two teeth immediately in front of the molars on both sides of the upper and lower jaws of humans and some other animals, used for grinding and chewing (crushing with the teeth) food

progressive loading: the gradual increase in the application of force on a dental implant whether intentionally done with a dental prosthesis or unintentionally via forces placed by adjacent anatomic structures or parafunctional loading

prosthesis: an artificial replacement of an absent part of the human body

prosthetic: relating to a prosthesis or prosthetics

prosthetics: the art and science of supplying artificial replacements for missing parts of the human body

prosthetic dentistry: see PROSTHODONTICS

prosthetic restoration: an artificial replacement for an absent part of the human body

prosthodontics: prosthodontics is the dental specialty pertaining to the diagnosis, treatment planning, rehabilitation and maintenance of the oral function, comfort, appearance and health of patients with clinical conditions associated with missing or deficient teeth and/or maxillofacial tissues using biocompatible substitutes

prosthodontists: a specialist in prosthodontics

pulp: the richly vascularized connective tissue of mesodermal origin with much innervation contained in the central cavity of the tooth

removable bridge: see REMOVABLE DENTAL PROSTHESIS

removable complete denture prosthesis: a removable dental prosthesis that replaces the entire dentition and associated structures of the maxillae or mandible

removable partial denture prosthesis: any prosthesis that replaces some teeth in a partially dentate arch. It can be removed from the mouth and replaced . It is also called partial removable dental prosthesis

removable dental prosthesis: 1. any dental prosthesis that replaces some or all teeth in a partially dentate arch (partial removable dental prostheses) or edentate arch (complete removable dental prostheses). It can be removed from the mouth and replaced at will, **2.** any dental prosthesis that can be readily inserted and removed by the patient

resilient: characterized or noted by resilience, as a) capable of withstanding shock without permanent deformation or rupture or b) tending to recover from or easily adjust to change syn. ELASTIC

resilient attachment: an attachment designed to give a tooth borne/soft tissue borne removable dental prosthesis sufficient mechanical flexion to withstand the variations in seating of the prosthesis due to deformations of the mucosa and underlying tissues without placing excessive stress on the abutments

resin: **1.** any of various solid or semisolid amorphous natural organic substances that usually are transparent or translucent and brown to yellow; usually formed in plant secretions; are soluble in organic solvents but not water; are used chiefly in varnishes, inks, plastics, and medicine; and are found in many dental impression materials, **2.** a broad term used to describe natural or synthetic substances that form plastic materials after polymerization. They are named according to their chemical composition, physical structure, and means for activation of polymerization

spongy bone: see CANCELLOUS BONE

subperiosteal dental implant: an eposteal dental implant that is placed beneath the periosteum while overlying the bony cortex

swallow: to cause food, drink, pills, etc. to move from mouth into stomach by using the muscles of throat, or to use the muscles of throat as if doing this

trabecular bone: see CANCELLOUS BONE