UNIVERSITÀ POLITECNICA DELLE MARCHE

FACULTY OF ENGINEERING

MASTER'S DEGREE IN BIOMEDICAL ENGINEERING

Finite Element Analysis of a dental prosthesis to simulate the behaviour under different masticatory loads

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ACADEMIC YEAR 2023/2024

To myself, my insecurities,
my anxiety,
and my determination,
to my dreams.

To my grandparents,
who have been my life teachers.

To the people who will see
their lives transformed by the
innovations, shaped today with
biomedical engineering.

I. Acknowledgements

I would like to express my deepest gratitude to all those who have contributed to the realization of this thesis and my academic career.

First and foremost, I extend my heartfelt thanks to my supervisor, Prof. Marco Mandolini, for his invaluable guidance, patience, and great advice. Thank you for supporting me at every stage of this project, allowing me to grow academically.

I would also like to thank my co-supervisor, Eng. Mattia Rondina, for his constant support and valuable suggestions. His expertise and availability have been a fundamental reference point for the successful completion of this work.

A special thanks goes to the company Yndetech and its owner Emidio Cennerilli, which hosted me during my internship, offering me the opportunity to apply the theoretical knowledge I had. Thanks, to Chiara and the entire team for their availability, professionalism, and warm welcome.

A special thanks goes to my family, who have always supported and encouraged me throughout my academic journey. Thank you to my parents, and my brother for their patience, and sacrifices. Thanks also to my grandparents, uncles, and cousins, whose interest in my studies given me the determination to overcome every challenge.

I would like to express my sincere gratitude to my company, Blu Pubblica Assistenza, and my colleagues of Potes 118 of Fermignano.

I would like to thank my friend Luca C., his father 'Gigi', and Giuditta B. for their invaluable clinical advice, which greatly contributed to the success of this project.

I would like to thank Luca D.C., Luca R., Romeo A., Andrea M., and my study and laughter companion Francesco C., for their friendship.

Finally, but not least, I would like to thank Elisa, who has always been by my side and never left me to face these last months of university alone.

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Abstract

Dentistry is a branch of medicine that deals with the diagnosis, prevention and treatment of diseases and conditions of the oral cavity, including dental, gums, and maxillofacial structures.

The digital dental market has greatly innovated implantology and dental prosthetics, which are the sectors that deal with the replacement of missing teeth by inserting dental implants through a surgical procedure, in order to support dental prostheses, such as: crowns, abutments, and even entire dental arches (e.g. the bridge in Toronto); instead, the components to fix such prostheses to the implant (i.e. to the endosseous screw) includes: fixing screws, prosthetic connections and T-Base.

3D imaging techniques, such as computed tomography (CT) or optical scanner, have become common practice and have replaced obsolete methods such as dental records with alginate. The optical scanners attached to a dedicated software can be handheld or over the counter and are widely used both in the company and in the dental offices.

The introduction of Computer-Aided Design (CAD) and Computer-Aided Manufacturing (CAM) technologies in the dental industry has revolutionized the way how dental devices and prostheses are designed and manufactured. These tools improve precision and accuracy, they are more efficient techniques from an economic and temporal point of view and can be controlled the quality of the product; therefore, having a more durable product.

The work done in this thesis was carried out in the R&D department of the company Yndetech, thoroughly analyzing their production method for dental prostheses. The current digital workflow includes optical scanning, file optimization via CAD software, such as Hypsocad and/or Rhinoceros 3D and then switch to CAM software (i.e. Oqton or Hyperdent) for milling or Selective Laser Melting (SLM), by obtaining the final medical device.

The aim is to bring innovation by incorporating CAE software such as ANSYS Workbench R1 or a dental CAE into ‘digital processing'. The latter allows to virtually test and optimize designs before physical production, analyzing how the prostheses will react to masticatory loads. By bringing possible changes to the structure and preventing failures, to reduce development time and costs. The state of the art is also updated with research on possible specific low-end software for the dental sector, one of these is DentalFEM.

The experimental part begins with the analysis of a partial prosthesis of the upper right arch of a four-element crown on T-Base, the teeth in question are 13,14,15, and 16. The latter was worn by a patient for 438 days and then break, a rupture analysis was made in which some hypotheses were advanced. The Zirconium Dioxide (ZrO_2) disc is by Aidite Technology Co. and has been milled. The technical specifications to be considered are the fracture toughness (K_{Ic}), a value of ≥ 158,11 MPa$\sqrt{mm}$, and the fractural strength of 1000 MPa. Defects in zirconia manufacturing range in an interval of [0,100 - 0,400] mm. K_{Ic} is used to calculate the critical length of the crack (a), the prosthesis does not reach immediate rupture up to cracks of about 0,319 mm by considering the K_{Ic}. Consequently, the Finite Element Analysis (FEA) on ANSYS was performed to obtain von Mises stress and total deformation values in the rupture areas, since any defects in the prosthesis were not known a priori.

The pressures, in which the forces were totally perpendicular to the occlusal surface and inclined forces of 45°, were considered. The simulations were conducted in two configurations: with normal chewing and with the Maximum Voluntary Bite Force (MVBF); scientific research provided the data. The project was carried out, first, simulating the case of study through the finite element method (FEM), then making changes to the failed prosthesis, increasing the thickness on the T-Base channels, or levelling them and lastly installing intermediate abutments between the prosthesis and the implant. In addition, a validation prosthesis has been created, modified 'ad hoc', under the minimum standards for production implementation, to validate the results obtained from the experiments.

The results showed, "border line" values for the failed prosthesis, that is a maximum of 157,35 MPa for the normal chewing which corresponds to an a of 0,321 mm. More meaning is given to the values in the normal chewing scenario (i.e. about 394.200 chewing cycles), since for the MVBF (i.e. a case of bruxism) the prosthesis would have had defects greater than 0,015 mm, and the cycles would not have had the same frequency.

The stress results of the other modified prostheses gradually decrease in all areas, starting from the one in which the wall thicknesses of the channels are increased to that equipped with intermediate abutments. So, a gradually increases to 5,32 mm for that of abutments, it means that before they break, they should exceed that dimensional threshold. The deformations extrapolated from the simulations have higher values for the broken prosthesis than the others, even null in some areas of rupture both for the cut prosthesis and for that with intermediate abutments. The data of the analysis and of validation method confirm the validity and accuracy of the FEM used for the simulations.

L'odontoiatria è una branca della medicina che si occupa della diagnosi, prevenzione e trattamento delle malattie e condizioni del cavo orale, comprese le strutture dentali, gengivali e maxillo-facciali.

Il mercato digitale dentale ha fortemente innovato l'implantologia e la protesistica dentale, che sono i settori che si occupano della sostituzione di denti mancanti mediante l'inserimento di impianti dentali tramite una procedura chirurgica, in modo da supportare protesi dentarie, come: corone, abutment e anche interi archi dentali (per esempio il ponte a Toronto); invece, la componentistica per il fissaggio di tali protesi all'impianto (cioè alla vite endossea) comprende: viti di fissaggio, connessioni protesiche e T-Base.

Le tecniche di imaging 3D, come la tomografia computerizzata (CT) o lo scanner ottico, sono diventate pratica comune e hanno sostituito metodi obsoleti come le impronte dentali con alginato. Gli scanner ottici annessi ad un software dedicato possono essere a palmare o da banco e trovano un grande utilizzo sia in azienda che negli studi dentistici.

L'introduzione delle tecnologie Computer-Aided Design (CAD) e Computer-Aided Manufacturing (CAM) nel settore dentale ha rivoluzionato il modo in cui vengono progettati e prodotti i dispositivi e le protesi dentali. Questi strumenti migliorano la precisione e l'accuratezza, sono tecniche più efficienti da un punto di vista economico e temporale e può essere controllata la qualità del manufatto; pertanto, avere un prodotto più durevole.

Il lavoro svolto in questa tesi è stato eseguito presso il reparto R&D dell'azienda Yndetech, analizzando minuziosamente il loro metodo di produzione per le protesi dentarie. Il workflow digitale attuale comprende la scansione ottica, l'ottimizzazione del file tramite software CAD, come Hypsocad e/o Rhinoceros 3D per poi passare ai software CAM (cioè Oqton o Hyperdent) per il fresaggio o il Selective Laser Melting (SLM), ottenendo il dispositivo medico finale.

Si vuole portare innovazione inglobando nella 'lavorazione digitale' un software CAE come ANSYS Workbench R1 o un CAE dentale. Quest'ultimo consente di testare e ottimizzare virtualmente i progetti prima della produzione fisica, analizzando come le protesi reagiranno ai carichi masticatori. Portando eventuali modifiche alla struttura e prevenendo i fallimenti, cosicché da ridurre i tempi e i costi di sviluppo. Lo stato dell'arte viene aggiornato anche con una ricerca su eventuali specifici low-end software per il settore dentale, uno di questi è DentalFEM.

La parte sperimentale inizia con l'analisi di una protesi parziale dell'arcata destra superiore di una corona a quattro elementi su T-Base, i denti in questione sono 13,14,15 e 16.

Quest'ultima è stata indossata da un paziente per 438 giorni per poi rompersi, è stata fatta un'analisi di rottura in cui si sono avanzate alcune ipotesi. Il disco di Ossido di Zirconia (ZrO_2) è della Aidite Technology Co., ed è stato fresato. Le specifiche tecniche da tenere in considerazione sono la tenacità (K_{Ic}), ovvero un valore di ≥ 158,11 MPa$\sqrt{mm}$, e la resistenza a flessione di 1000 Mpa. I difetti per la lavorazione della zirconia variano in un intervallo di [0,100 - 0,400] mm. K_{Ic} serve per calcolare la lunghezza critica della cricca (a), la protesi non arriva a rottura immediata fino a cricche di circa 0,319 mm, considerando il K_{Ic}. Dopodiché l'Analisi agli Elementi Finiti (FEA) su ANSYS è stata eseguita per ottenere valori di stress di von Mises e delle deformazioni totali nelle aree di rottura, siccome non si conoscevano a priori eventuali difetti della protesi.

Sono state considerate pressioni in cui le forze erano totalmente perpendicolari alla superfice occlusale e forze inclinate di 45°. Le simulazioni sono state condotte in due configurazioni: con la masticazione normale e con la Massima Forza di Morso Volontario (MVBF); la ricerca scientifica ha fornito i dati. Il progetto si è svolto, dapprima, simulando attraverso il Metodo agli Elementi Finiti (FEM) il caso di studio, per poi apporre modifiche alla protesi rotta, aumentando lo spessore nei canali dei T-Base o livellandoli ed infine installando degli abutment intermedi tra la protesi e l'impianto. Inoltre, è stata creata una protesi di validazione, modificata 'ad hoc', sotto gli standard minimi per la messa in produzione, cosicché da avvalorare i risultati ottenuti dalle sperimentazioni.

I risultati hanno mostrato valori "border line" per la protesi fallita, ovvero un massimo di 157,35 MPa per la masticazione normale che corrisponde ad una a di 0,321 mm. Si vuole dare più significato ai valori nello scenario di masticazione normale (cioè, circa 394.200 cicli di masticazione), dal momento che per il MVBF (cioè, un caso di bruxismo) la protesi avrebbe avuto difetti superiori a 0,015 mm, e i cicli non avrebbero avuto la stessa frequenza.

I risultati degli stress delle altre protesi modificate diminuiscono gradualmente in tutti i settori, partendo da quella in cui gli spessori delle pareti dei canali vengono aumentati fino a quella equipaggiata con abutment intermedi. Quindi, le a aumentano gradualmente fino ad arrivare per quella ad abutment a 5,32 mm, il che significa che prima di arrivare a rottura dovrebbero superare quella soglia dimensionale. Le deformazioni estrapolate dalle simulazioni hanno valori più alti per la protesi rotta rispetto alle altre, addirittura nulle in alcune aree di rottura sia per la protesi tagliata sia per quella con abutment intermedi. I dati delle analisi e del metodo di validazione confermano la validità e l'accuratezza del FEM utilizzato per le simulazioni.

1 Introduction

Prosthetics play a crucial role in improving the quality of life of many people, the improvement of physical functions through prostheses reduces dependence on others and increases the possibility of full participation in social, working, and recreational life. Then, help to restore the natural appearance of the body, reducing the negative psychological impact, and allow to replace damaged or non-functioning parts, enhancing the general health of the patient. Prosthetics represent an extraordinary combination of medical science and engineering; the combination of these disciplines guarantees a high-quality product.

The oral cavity is a complex biomechanical system, due to this complexity and limited access, most biomechanical research of the oral environment such as prosthodontics and implantology has been performed in vitro; the material science must be a discipline embedded in dental clinical practice.

So, dental prostheses are artificial medical devices used to replace missing parts into the mouth, such as crowns, stumps, or entire dental arches, to restore functionality and improve the quality of life.

In the '80s and '90s, the dental industry has evolved, by abandoning obsolete techniques and entering the digital world using the latest generation of Computer-Aided Design (CAD), Computer-Aided Manufacturing (CAM), and Computer-Aided Engineering (CAE) software.

Despite advances in technology and materials, prostheses can fail for several reasons. Understanding these failures is essential to improve the design, materials, and clinical outcomes. Prosthetic failure analysis is a critical area of research in biomedical engineering and healthcare useful in a company R&D department.

The design and development of dental prostheses represents a complex engineering challenge that requires an accurate understanding of mechanical stress and oral biomechanics.

Deformation and stresses are generated when masticatory loads are applied to a structure. This is usual and is how a structure performs its structural function. But if stresses become excessive and exceed the elastic limit, structural failure may result.

These stresses cannot be directly measured in vivo, and it is not easy to understand why and when a failure process is initiated in complex structures.

Furthermore, dental prostheses must withstand high forces during chewing, maintain functionality over time and offer comfort and safety to patients.

To address these challenges, it takes advantage with Finite Element Analysis (FEA), which has been widely used through numerical analysis that has been successfully applied in many engineering and bioengineering areas since the 1960s. The Finite Element Method (FEM) is a numerical procedure for analyzing structures. The problem addressed is too complicated to be satisfactorily solved by classic analytical methods. The problem may concern stress analysis, deformations, and eventually stress-life analysis.

FEA is based obtaining a solution to a complex physical problem by dividing the problem domain into a collection of much smaller and simpler domains, termed 'elements' connect through 'nodes'.

This thesis aims to explore the application of FEA simulations in the field of dental prosthetics, with the goal of improving the mechanical performance and durability of prostheses, through a series of tests and structural changes.

Identifying the stress distributions and the total deformation in critical areas onto the prosthesis during chewing is essential to avoid sudden breakage; these values must be compared with the fracture toughness and flexural strength of the material used in order to understand any criticality.

Indeed, a focus on mechanical properties such as fracture toughness and flexural strength will ensure that prostheses are not only functional but also durable in the long term.

The use of FEA simulations in denture design offers many advantages, including the ability to virtually test different configurations and materials, thus reducing the need for costly and lengthy experimental tests. This approach allows potential structural problems to be identified and solved at the design stage, leading to safer, more effective, and more comfortable prostheses.

Other advantages of this method compared with other research methodologies are the low operating costs, reduced time to carry out the investigation and information provision that cannot be obtained by experimental studies.

Through a systematic and empirical approach, the future aim is to develop safer, longer lasting, and more comfortable prostheses.

The project of this thesis sees the application of such methods in a pre-existing digital workflow of the company, Yndetech, which worked based on CAD, and CAM design.

For this, the results obtained will help to establish guidelines for a possible future design and implementation of this new method in the digital workflow, thus optimizing dental medical devices.

The use of FEA simulation has proven to be an effective tool for understanding the mechanisms of denture failure. Therefore, the FEA simulation has a positive impact not only in prosthetics but also in the study of bone conditions surrounding the implant, guided surgery, and many other dental practices.

Recently with the advances of digital imaging systems (i.e. optical scanner and CT), it has become possible to extrapolate the individual specific data of teeth and gums geometries and properties to an FEA model; this helps biomedical engineers and dental laboratories to extrapolate the geometry of oral structures quickly and accurately. Thus, to be optimized and analyzed by FEA. Through the internship, it was understood how certain Yndetech issues, such as the premature breakage of prostheses during patient mastication or due to the manufacturing process, can be resolved with FEA simulation during post-production. Additionally, dental technician study projects that do not meet dimensional standards can be improved during pre-production phase by implementing a conservative design. So, this thesis aims to give an answer to the possibility of integrating FEM, multi-approved and realistic, in the company production process for dental supplies, increasing the reliability of prostheses and cost-effectiveness.

2 State of the art

An increase in the world population and its life expectancy, as well as the ongoing concern about our physical appearance, have elevated the relevance of dental implantology in recent decades. Engineering strategies to improve the survival rate of dental implants have been widely investigated, focusing on implant material composition, geometry (usually guided to reduce stiffness), interface surrounding tissues and Computer-Aided Design (CAD) and Computer-Aided Manufacturing (CAM) process. [1]

CAD and CAM technology has revolutionized dentistry, offering numerous benefits in various dental procedures. These technology assists in the planning, designing, and manufacturing of dental implants.

Further implementation of the Computer-Aided Engineering (CAE) software, for Finite Element Analysis (FEA), will improve even more the cost and productivity of dental prosthesis.

2.1 Dental anatomy

The artificial teeth are placed in oral cavity (Figure 2.1) [2] attached to the upper or inferior jaw or both. In the mouths of humans and other mammals, food is mechanically ground by the teeth, through the movement of physiological chewing. In humans, mainly the masticatory function is carried out by molars and pre-molars which, by rubbing against each other, reduce food to mush. Instead, incisors and canines allow to tear off food by grinding it roughly. Each tooth is housed in a cavity in the mandible or maxilla (i.e. upper and lower jawbone) called a dental alveolus. The dental anatomy is a field of anatomy dedicated to the study of human tooth structures. The development, appearance, and classification of teeth fall within its purview. Tooth formation begins before birth, and the teeth's eventual morphology is dictated during this time. Dental anatomy is also a taxonomical science: it is concerned with the naming of teeth and the structures of which they are made, this information serving for practical purpose in dental treatment to facilitate the medical doctor job.

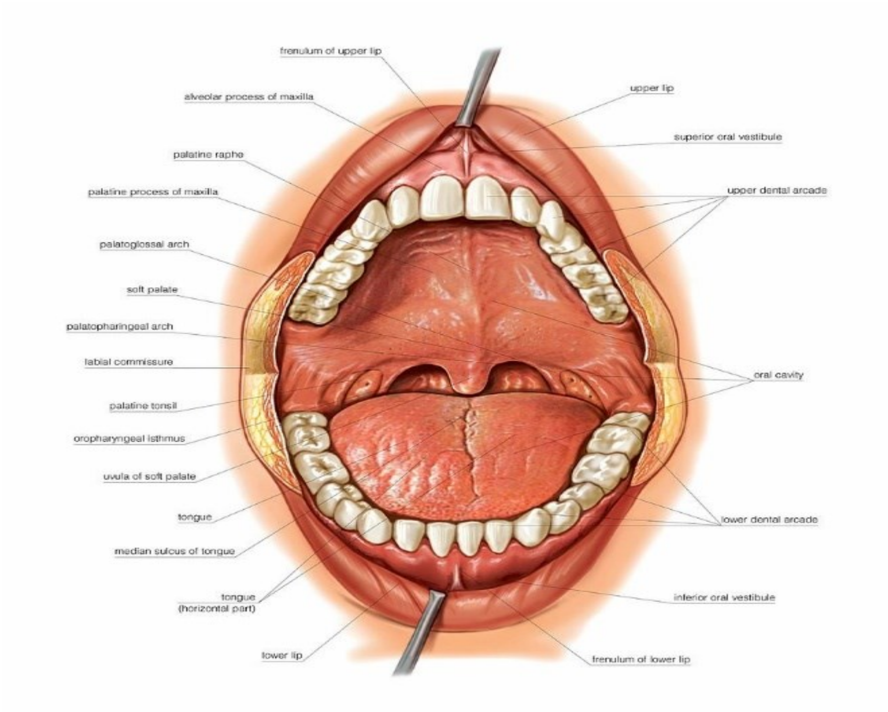


Figure 2.1 Oral cavity

Usually, there are 20 primary teeth (regarding baby) and 32 permanent teeth (regarding adult), the last four being third molars or "wisdom teeth", each of which may or may not grow in. Among primary teeth, 10 usually are found in the maxilla (i.e. upper jaw) and the other 10 in the mandible (i.e. lower jaw). Among permanent teeth, 16 are found in the maxilla and the other 16 in the mandible.

Understanding dental anatomy is fundamental to provide effective dental care, since it enables dental professionals to diagnose problems accurately and develop appropriate treatment plans to maintain oral health.

The teeth are divided in crown, neck, and root. The crown is the part of the tooth that protrudes from the socket. It is visible to the naked eye, it takes on a different shape depending on the type of tooth: in the incisors it is flattened and sharp, in the canines it is pointed and widened, in the molars and pre-molars it has several cusps. The neck is located between the root and the crown where it forms a transitional tissue (transmucosal passage) around which the gum surrounds it. At last, one or more roots insert into the alveolus, they are anchored to its walls by connective tissue ligaments (i.e. periodontal ligament). The root is single in the incisors, canines, and lower pre-molars, while it is double or triple in the upper pre-molars and molars. Moreover, inside teeth there are several layers (Figure 2.2):

- Enamel: the outermost layer, which is the hardest and most mineralized part of the tooth.
- Dentin: a layer beneath the enamel, softer than enamel but harder than bone.
- Pulp: the innermost part, consisting of nerves, blood vessels, and connective tissue.

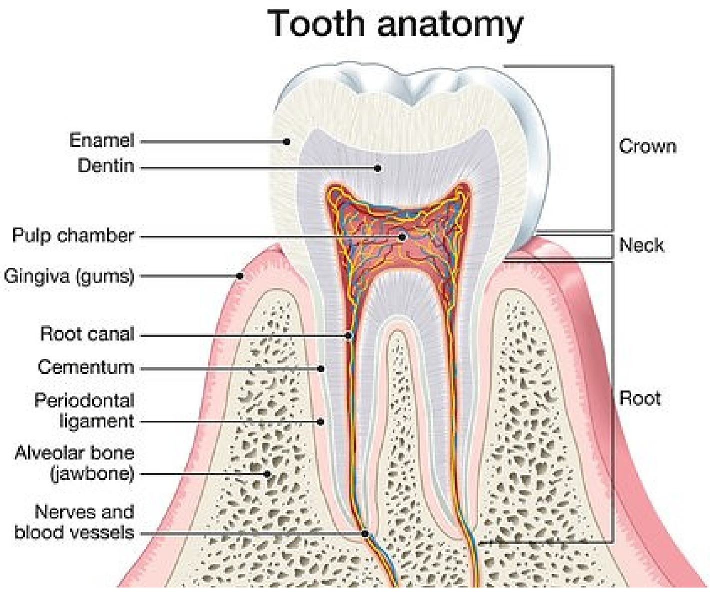


Figure 2.2 Tooth anatomy

2.1.1 *Nomenclature and numbering system*

Teeth are named by their sets, arch, class, type, and side. Teeth can belong to one of two sets of teeth: primary teeth or deciduous and permanent teeth or succedaneous.

Furthermore, the name depends upon which arch the tooth is found in. The term, maxillary, is given to teeth in the upper jaw and mandibular to those in the lower jaw. There are four classes of teeth: incisors, canines, pre-molars, and molars which belong to the permanent dentition. Pre-molars are found only in permanent teeth; there are no pre-molars in primary teeth. The incisors are divided further into central and lateral incisors. Among pre-molars and molars, there are first and second pre-molar. Going inside the mouth first, second and third molars. The side of the mouth in which a tooth is found may also be included in the name.

As regards numbering system, the teeth are numbered in a consistent manner to facilitate communication between dental professionals. There are several different dental notation systems for associating information to a specific tooth. The three most commons systems are the 'World Dental Federation (FDI)', 'Universal numbering system' and 'Palmer notation method'. The FDI notation system is used worldwide, and the Universal is used widely in the United States. In the Figure 2.3 the numbers and classification of the teeth by considering the FDI World Dental Federation notation for deciduous and permanent teeth. [3]

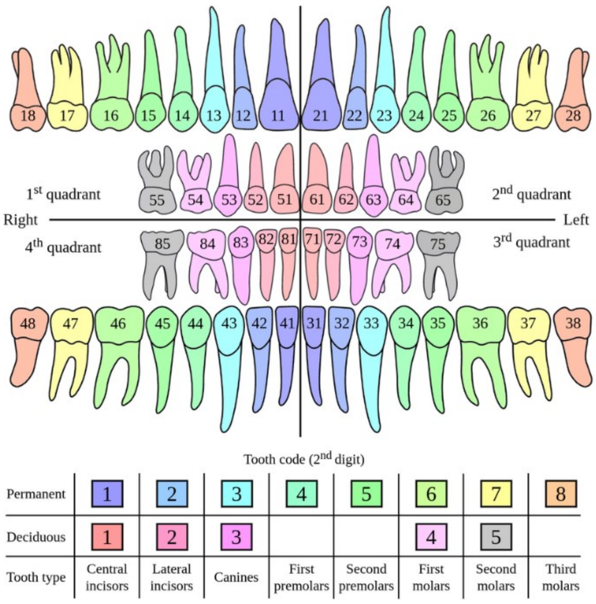


Figure 2.3 The World Dental Federation (FDI) notation

2.2 Digital dental market

Digital dental market refers to the sector within the dental industry that involves the use of digital technologies and computer science for various dental procedures, diagnostics, treatment planning, production, and marketing. This market encompasses a wide range of products and services aimed at improving efficiency, accuracy, and patient outcomes in dental care.

The digital dental market continues to evolve rapidly with advancements in technologies such as artificial intelligence, virtual reality, and augmented reality, which are being integrated into various aspects of dental practice to further enhance efficiency, accuracy, and patient experience.

Additionally, the market is driven by factors such as increasing adoption of digital workflows by dental professionals, growing patient demand for advanced treatments, and the need for improved clinical outcomes and productivity.

2.2.1 Evolution of the dental market

Dentistry embraces new materials and new technologies that date back decades. Soon after the discovery of anaesthetics the dental drill was invented, which meant that filling materials such as silicates and amalgams became widely used. In the early 20th century Dr. William H. Taggart introduced the loss-wax casting process to dentistry for the construction of crowns and bridges, which was adapted from the method then used in the jewellery business, in fact materials as polymer and gold was the most used. The developments in new polymers during the 1940s and 1950s resulted in the use of acrylic resins for dentures, acidic polymers for restorative cements and monomers for composite resin restorative materials. The lasting contributions of Michael Buonocore, Dennis Smith, Raphael Bowen, John McLean, Alan Wilson, and many others in this respect are well known. The discovery, by Dr. Per-Ingvar Brånemark, of the special properties of titanium metal did not take long to be translated into an explosion in dental implantology. Thus, dentistry has shown itself to lead the medical disciplines in embracing new materials and new technologies to utilize. So, it has also proved over time to make use of new technologies such as CAD and CAM software.

CAD and CAM began its dental life in 1970s with the first worker to explore its application in dentistry being Prof. François Duret and Dr. Jack D. Preston.

This was followed by the work of Prof. Werner H Mörmann in the 1980s. CAD and CAM have now become a well-accepted technology in most modern dental. The first digital revolution took place many years ago, now with the production of dental restorations such as veneers, inlays, crowns, and bridges using dental CAD and CAM systems and new improved systems appear on the market with great rapidity. The reducing cost of processing will ensure that these developments will continue with the succession of the years.

Regarding the manufacture of prostheses this is currently dominated by subtractive machining technology, but it is inevitable that additive processing, or layer manufacturing, has also taken hold, such as Fused Filament Fabrication (FDM), Stereolithography (SLA), Selective Laser Melting (SLM) and inkjet printing, will start to have an impact.

In principle there is no reason why the technology cannot be extended to all aspects of production of dental prostheses and include customized implants, full denture construction and orthodontic appliances. In fact, anything that can expect a dental laboratory to produce can be done digitally and potentially more consistently, quicker and at a reduced cost. [4]

With the improvements in the speed, reliability, and accuracy of the hardware, additive manufacturing will seriously compete with traditional manufacturing in creating end-use products. One advantage with additive manufacturing is that it eliminates much of the expensive and highly skilled laboratory associated with traditional manufacturing. Another good feature of additive manufacturing is that it can make any number of complex products simultaneously.

To conclude, the digital technology is advancing rapidly in dentistry. Computers are making what were previously manual tasks easier, faster, cheaper, and more predictable. Layered manufacturing processes can produce complex shapes at affordable prices with little or no waste. The challenge for the dental materials research community is to marry the technology with materials that are suitable for use in dentistry.

2.3 Implantology and dental prosthetics

A high number of patients have one or more missing tooth, and it is estimated that one in four American subjects over the age of 74 have lost all their natural teeth.

Indeed, the edentulism remains a major disease worldwide. Many options exist to replace missing teeth, but dental implants have become one of the most used biomaterials to replace one (or more) missing tooth over the last decades. Contemporary dental implants made with titanium have been proven safe and effective in large series of patients. [5]

Implantology and dental prosthetics are two fundamental disciplines of modern dentistry that often work in synergy to restore the function and aesthetics of patient's teeth.

These fields represent a great engineering challenge, resulting in precision, accuracy and control of mechanical quantities such as fracture mechanics.

Implant-supported prostheses offer greater comfort, stability and superior aesthetics, improving the patient's quality of life. Implantology and dental prosthetics offer advanced solutions for the replacement of missing teeth, greatly improving the quality of life of patients. The collaboration between these two disciplines allows to achieve excellent results both from a functional and aesthetic point of view.

Dental prosthetics is a branch of dentistry that deals with the design, construction and application of dental prostheses to replace missing or damaged teeth. The choice of the type of prosthesis depends on the specific needs of the patient, the condition of the mouth and the desired aesthetic result. The dental prostheses can be fixed, removable or supported by implants.

Hence, it can simply explain that implantology and prosthetics are two fundamental branches of dentistry for each other. Implantology is about what it cannot see from external, namely the interaction with the upper and lower jawbone and the methods and components of attachment with the prostheses. Instead, dental prosthetics is about what it can see from the outside: aesthetic components and finishing methods to make them as realistic as possible.

Dental prosthetics, combined with implantology, offers advanced solutions for a wide range of dental problems. Not least is the study of materials that must ensure biocompatibility, bioactivity for osseointegration, and bioinert especially for the parts in contact with the mucous membranes that would represent cases of hypersensitivity and could result in failure of the medical device.

Speaking about dental implantology, it is a specialized field of dentistry that focuses with the placement, restoration, and maintenance of dental implants, it deals permanent implantation or attachment of artificial teeth.

Dental implants are screws, so they are artificial tooth roots made of different materials, usually titanium, that are surgically placed into the maxillary or mandibular bones to support dental prostheses such as crowns, bridges, or entire dental arches (e.g. Toronto bridge).

Essentially, a dental implant prosthesis is composed by three elements (Figure 2.4):

- Endosseous screw (implant): it mimics the root features tooth that is lost. Often, it is composed with titanium to match with the bone;
- Abutment: it serves as a connection among the screw and the prosthesis that it will substitute (there are some different abutment);
- Dental prosthesis: it replaces the lost teeth; it can be made with different material that can be ceramic, resin, or metal.

Figure 2.4 Fixed dental prosthesis section

Basically, it considers several steps, as first, an examination of the patient's dental and medical history, along with diagnostic imaging such as X-rays or CT scans to evaluate the bone structure and condition of the jaw. This helps the dentist determine if the patient is a suitable candidate for dental implants and plan the treatment accordingly.

Once the treatment plan is finalized, the dental implant(s) are surgically placed into the jawbone. This procedure is typically performed under local anaesthesia, though sedation. The implants are strategically positioned to provide optimal support and stability for the prosthetic teeth.

After the implants are placed, a process called osseointegration occurs, during which the bone tissue fuses with the surface of the implant. This integration provides a strong and durable foundation for the artificial teeth.

Once osseointegration is complete, the implants are ready to be restored with artificial teeth. This may involve attaching support to the implants, which serve as anchors for the prosthetic teeth.

The prosthetic teeth are custom-made to match the colour, shape, and size of the patient's natural teeth for a seamless appearance.

Implantology in conjunction with dental prosthetics offers numerous benefits, including improved aesthetics, function, and stability compared to traditional dentures. Moreover, the surgical practice wants to preserve the residual bone.

With proper care, dental implants can last for many years. Implantology improve patient's comfort in chewing, confidence in smiling and speaking and, the overall psychological health. It can provide support for a partial denture.

At the end of the XXth century, a major advance in the treatment of tooth loss is represented by the discovery of dental implants. Although implantology has not the same importance that other surgical techniques concerned with life-threatening diseases, the correction of a dental deficit influences the physiological and psychological condition of the patients and improves their quality of life in all aspects. [4]

From the manufacturer side, there are many factories in the world that produce models of different implants, the most famous companies of this field are: 'Straumann', 'Nobel Biocare', 'Zimmer Biomet', and 'Sweden & Martina'.

2.3.1 *Dental implant prosthesis*

The fixing system can be different, so the odontologist can set up a screwed prosthesis, a cemented one or a conometric implant (this last without screws and cement but it considers the friction forces). This thesis aims with screw-retained dental prosthesis for the implantology field. The manufacturer can produce a single implant or a multiple implant. Multiple consider from two connected implants to a whole dental arch. In Figure 2.5 is designed for a patient the entire superior dental arch on Rhinoceros 3D called 'implant bar'.

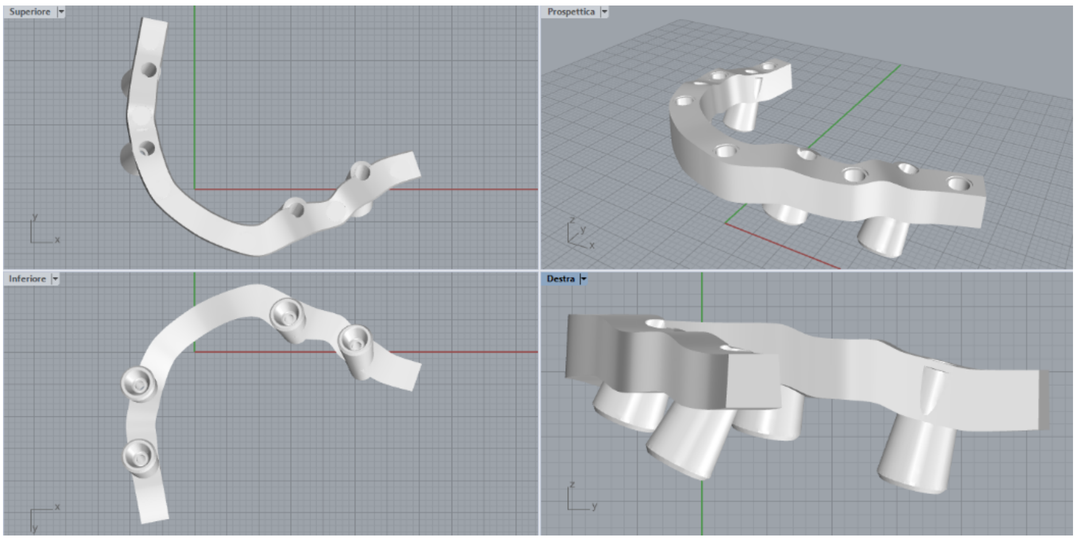


Figure 2.5 Implant of dental arch

2.4 Implant components

Implant components refer to the individual parts that make up a dental implant system. Dental implants are used to replace missing teeth and typically consist of several components that work together to mimic the structure and functions of a natural tooth. The implant components are different and depend on the devices used. Logically, on the basal bone the dentist will screw an endosseous screw. An abutment will mate with the endosseous screw with a shape coupling at the lower extremity (normally the screw has an external or internal hexagonal). The abutment and the crown with the prosthetic connection can be the same piece or not and directly they will couple with the bone implant screws. Hence, the implant components are:

- Crown (the outermost and visible part of the implant);
- Abutment (the inner layer of the crown);
- Rotational or irrotational prosthetic connections (it can be produced alone or as a single piece with the abutment or crown);
- T-Base (connection between endosseous screw and prosthesis);
- Endosseous screw (innermost part of the implant that is in direct contact with the jawbone).

The entire system can be composed either with the abutment or with T-Base, it depends on the design part of the prosthesis; not all items are used at the same time, it is at the discretion of the customer which requires the work and the system adopted.

2.4.1 Crown

An artificial crown, also known simply as a dental crown or a cap, is a prosthetic device used to restore the shape, size, strength, and appearance of a damaged or decayed tooth. Crowns are custom-made to fit over the entire visible portion of a tooth above the gum line, effectively covering and protecting it. Hence, it has different function, as for example, restoring the function of a damaged or weakened tooth. It protects a tooth from further damage or decay. Then, improving the appearance of a misshapen or discoloured tooth. Strengthening a tooth that has undergone root canal therapy or other extensive dental treatments.

Logically, it is impossible to recreate the perfect shape and size of an original dental crown. So, the concept of design an artificial crown is based on the reverse engineering that is made with the use of optical or laser scanner, that it is discussed further on.

Furthermore, the artificial crowns, as represented on Rhinoceros 3D in Figure 2.6, are a versatile and effective solution for restoring the health, function, and appearance, helping patients maintain their oral health and quality of life.

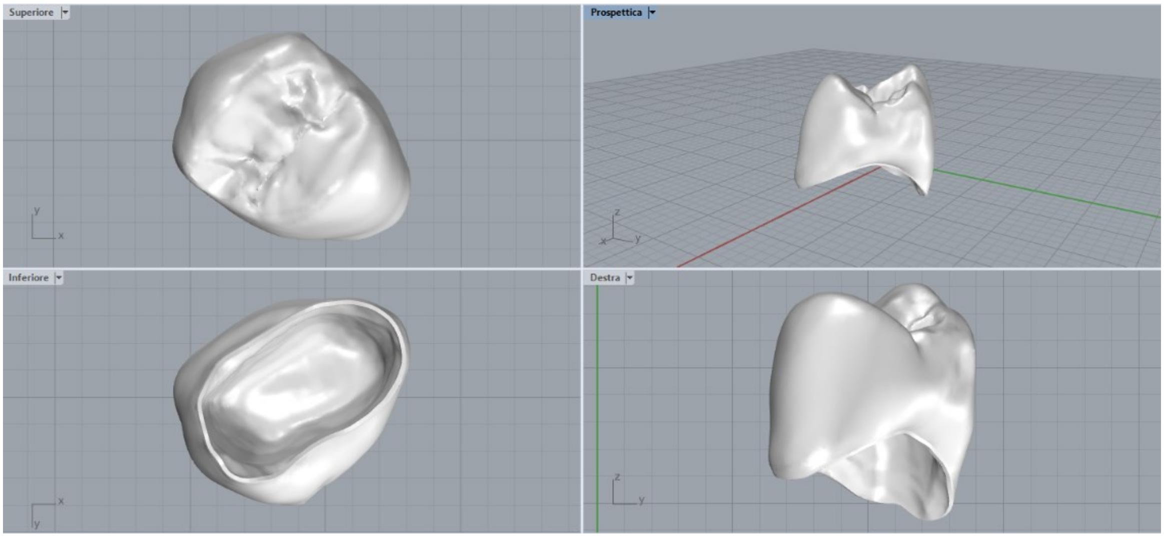


Figure 2.6 Artificial dental crown

The crown can be done with several materials, gold, porcelain, zirconium dioxide (ZrO_2), metal harvested with ceramic materials and composites. Then, it has cemented with zinc oxide or glass ionometrics, the last is preferred for low patient's sensitive.

2.4.2 *Abutment*

Abutment refers to the natural stump of tooth and not the crown. Hence, an artificial dental abutment is a component used in dental implants to support dental prostheses such as crowns. It serves as a foundation onto which the artificial crown is attached.

The abutment is typically made from materials such as titanium, zirconium dioxide, Cobalt-Chromium (CoCr) or a combination of materials. It is surgically placed in conjunction with the endosseous screws. Once the abutment has connected successfully, it provides a stable and durable anchor for the dental prosthesis.

Artificial dental abutments are made in various shapes and sizes to accommodate different clinical situations and aesthetic requirements. They can be customized to match the natural teeth in terms of colour, shape, and size, ensuring a natural-looking smile for the patient, if it want to get a finished product, and are called ‘direct screw crown abutment’ (i.e. a single piece). The most common are those that represent the element that fills a gap between the artificial crown and the endosseous screw, that they are simply cylinders or other forms.

Additionally, advancements in dental technology have led to the development of CAD and CAM technology, which allows for precise fabrication of custom abutments for optimal fit and function.

It is possible to manufacture it either above or a single piece with the prosthetic connection. Furthermore, depending on the geometry of the prosthetic connection the abutment can have a rotational or irrotational behaviour to meet customer requirements. In Figure 2.7 (a) the CAD project of a crown abutment with irrotational prosthetic connection. Instead, in Figure 2.7 (b) a simpler abutment which fills the gap between the implant and prosthesis.

The artificial abutment serves to cement the artificial crown because it is not aesthetic component, whereas the crown is very important for the functioning of chewing and aesthetic question.

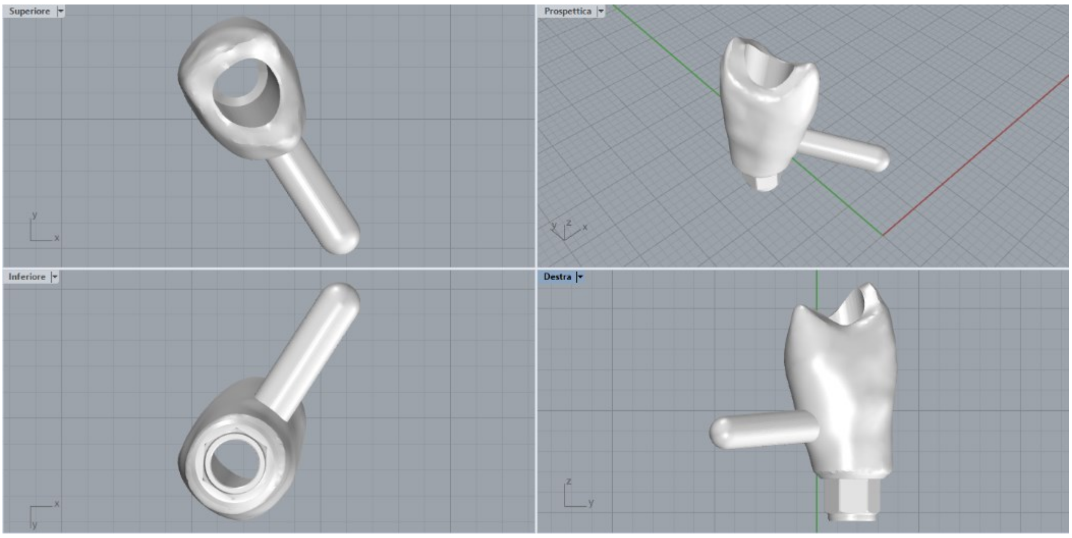


Figure 2.7 (a) Crown abutment with irrotational connection

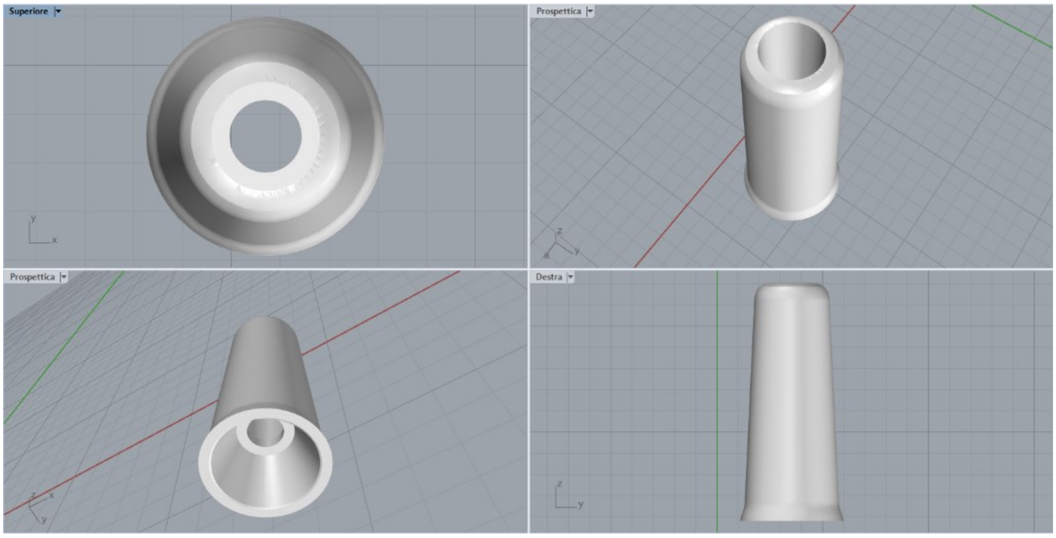


Figure 2.7 (b) Cylindrical abutment

2.4.3 *Irrotational prosthetic connection*

Irrotational prosthetic connection couple match with dental implant, and the piece above, whatever it is. It refers to a type of implant that is designed to remain stable and fixed in its position once it's been placed, without the need for rotation or adjustment after implantation.

Traditional dental implants typically require precise placement and may need adjustments during the healing process to ensure proper alignment and integration with the surrounding bone.

Hence, an irrotational connection has a particular shape that cannot permits to rotate, and it is fixed in its axes before it is applied on the patient. Often, it has an anti-rotational surface in contact with the screw body, and then the prosthesis is screwed with the internal thread of the endosseous implant.

In Figure 2.8, it is depicted in Rhinoceros 3D CAD form, an irrotational implant and the model is: 'Sweden & Martina Global 3.8'. The connection depends on the implant company, it is personalized for different type of implants and not mass produced. It is connected inside with the artificial crown or abutment, but it can be also manufactured as single piece with the external component. It depends to the requirements of implant.

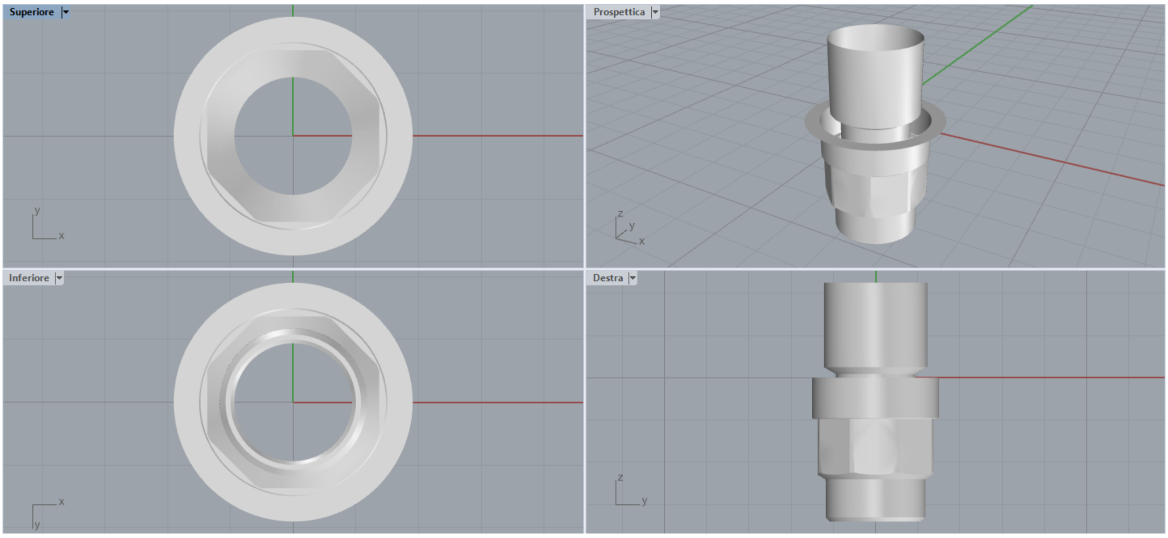


Figure 2.8 Sweden & Martina Global irrotational implant 3.8

2.4.4 Rotational prosthetic connection

The rotational prosthetic connection refers to a type of dental implant that can be rotated during the placement process to achieve optimal positioning within the jawbone. This rotational capability allows the dentist or oral surgeon to adjust the implant's orientation to ensure proper alignment with adjacent teeth and to optimize support for prosthetic components such as crowns and abutments.

Rotational dental implants are designed with features that enable controlled rotation during insertion, such as specific thread designs or external features on the implant body. This flexibility in placement can be particularly beneficial in cases where there are anatomical considerations or limited space within the upper and lower jaw.

It's important to note that while rotational dental implants offer increased versatility during placement, they still require precise surgical technique to ensure successful integration and long-term stability within the jawbone. Additionally, the rotational capability does not imply ongoing rotation after placement; once the implant is fully integrated and prosthetic components are attached, it should remain fixed in position like any other dental implant. In this case, in Figure 2.9 is reported, in Rhinoceros 3D CAD form, a rotational implant. In the same manner, the model is the same of irrotational device but rotational: 'Sweden & Martina Global 3.8', for consistency. It is possible to appreciate the form smoother of the body part in contact with the screw with respect the anti-irrotational surface of the previous.

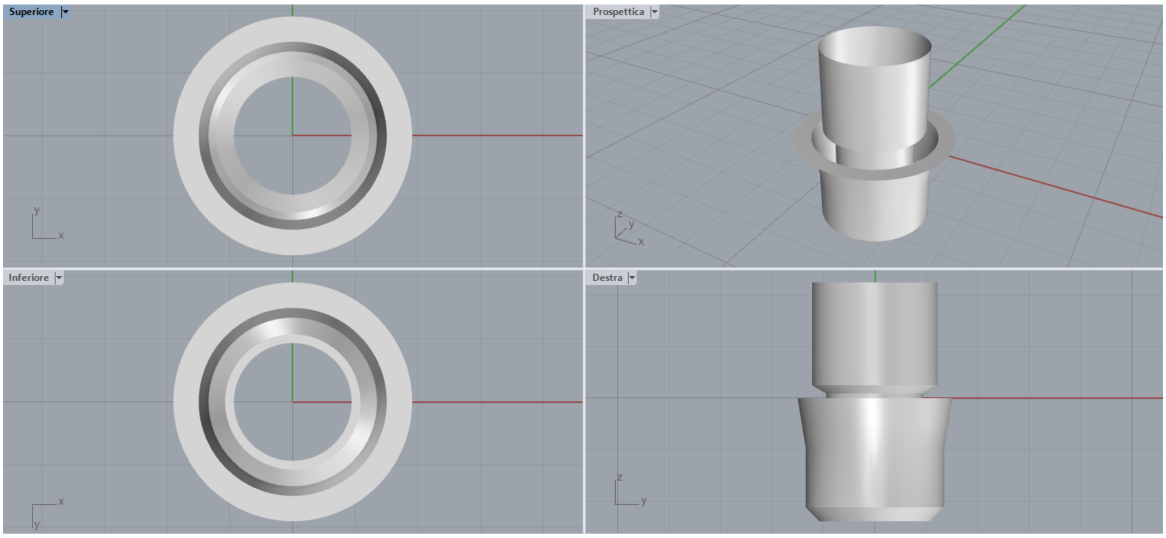


Figure 2.9 Sweden & Martina Global rotational implant 3.8

2.4.5 *Analogous*

It can imagine from the name the concept of this device; in fact, it represents the substitute of endosseous screw. This kind of device is very important in modelling and designing. Indeed, the dental technician uses the analogous as taking over of the endosseous screw into moulds. This because an endosseous screw in a model is very expensive and not necessary. The analogous serves to couple the scan body so it is possible to scan the oral cavity and it represents the perfect position of the abutment or crown screwed in the jawbone. In Figure 2.10 the analogous CAD project.

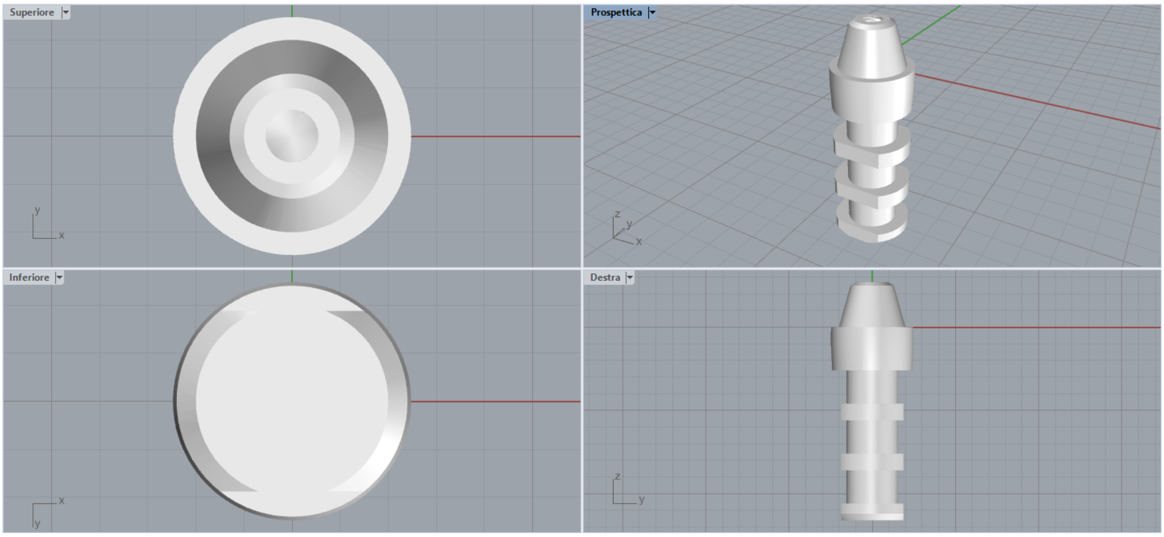


Figure 2.10 The analogous

In Figures 2.11 (a) – (b) the coupling CAD project of the analogous with the scan body, in a clipping version in order to have an inside vision of the items and its matching. Scan body and analogous are matched with the same (negative and positive) connection. In real implant there will be endosseous implant for the analogous, and abutment or crown with the prosthetic connection for the scan body.

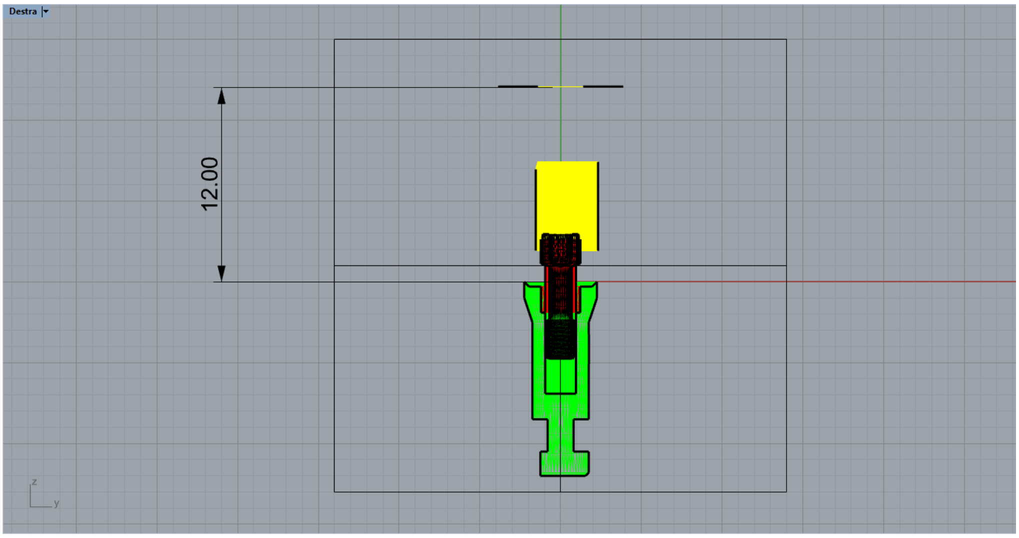


Figure 2.11 (a) Clipping view of the analogous-scan body coupling

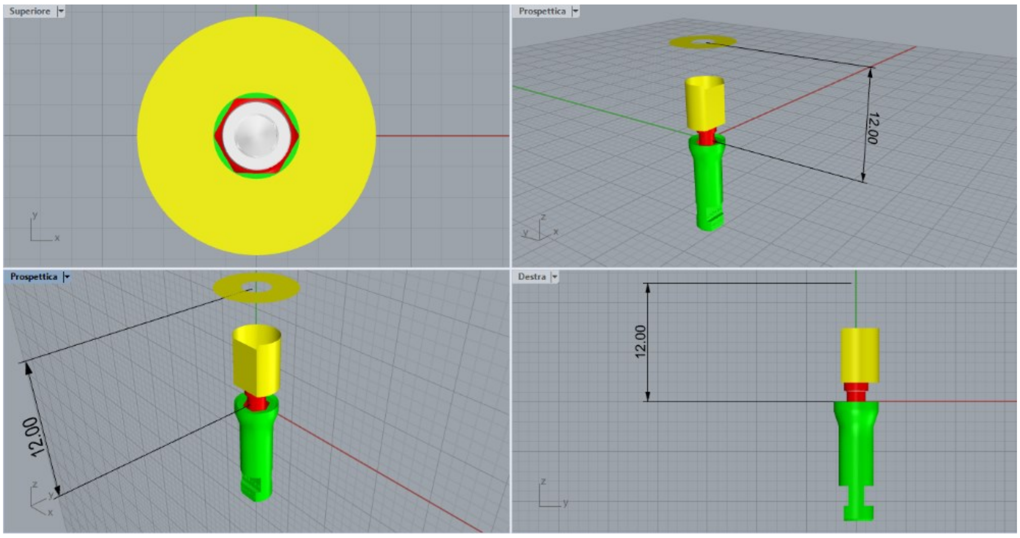


Figure 2.11 (b) All view of the analogous-scan body coupling

In Figure 2.11 (a) the clipping view where the yellow part refers to the scan body, the red part to the prosthetic connection, the green part to the analogous and at last the grey (visible only on Figure 2.11 (b)) is the screw that serves to fix the scan body with the analogous. The scan body quote has depicted because it is a very important parameter because if there are discrepancy between scan body during the intraoral scan phase the implant will be wrong manufactured. On Figure 2.11 (b) all the views of the coupling.

2.4.6 T-Base

This device is mass produced whereas the prosthetic connection, the abutments and the crowns are customed to the patient's implant. Essentially, both the abutment and the T-Base carry out the function of tooth stump. Furthermore, it is possible to choose an irrotational or rotational T-Base like for the connections.

In Figures 2.12 (a) - (b), it is possible to see the difference between the two CAD project of T-Bases. The irrotational T-Base presents a no-smooth face at the base of the cylinder that a further constraint on rotation, whereas the rotational one has smoother surfaces.

The advantageous of T-Base is that is simple to produce but it cannot manufacture as single piece, but always it must paste with the artificial crown. T-Base connect to the endosseous implant and then it is cemented with the artificial crown, and inside there is the fixation screw.

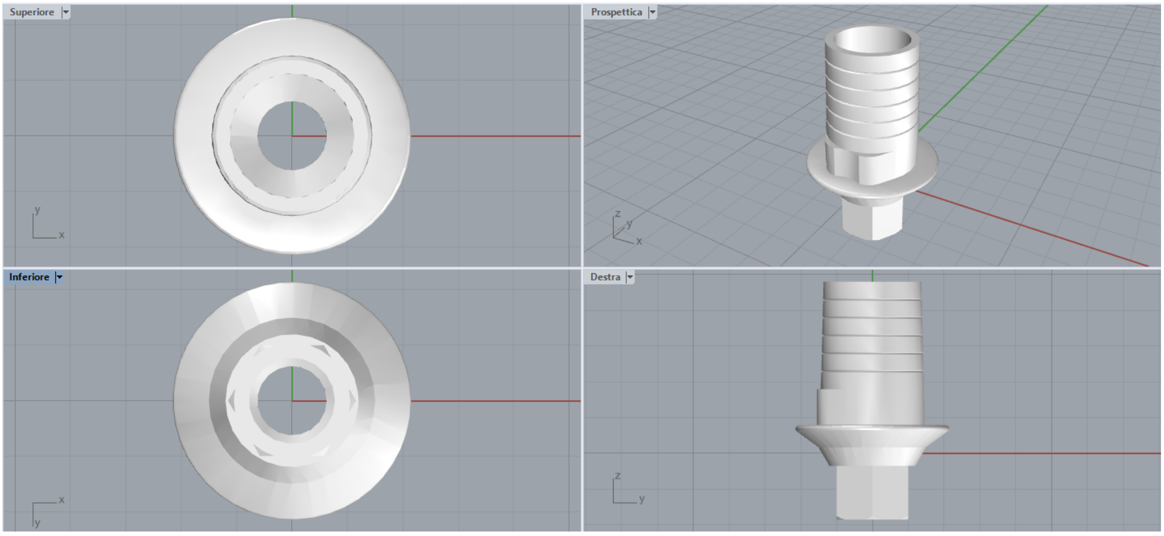


Figure 2.12 (a) Irrotational T-Base

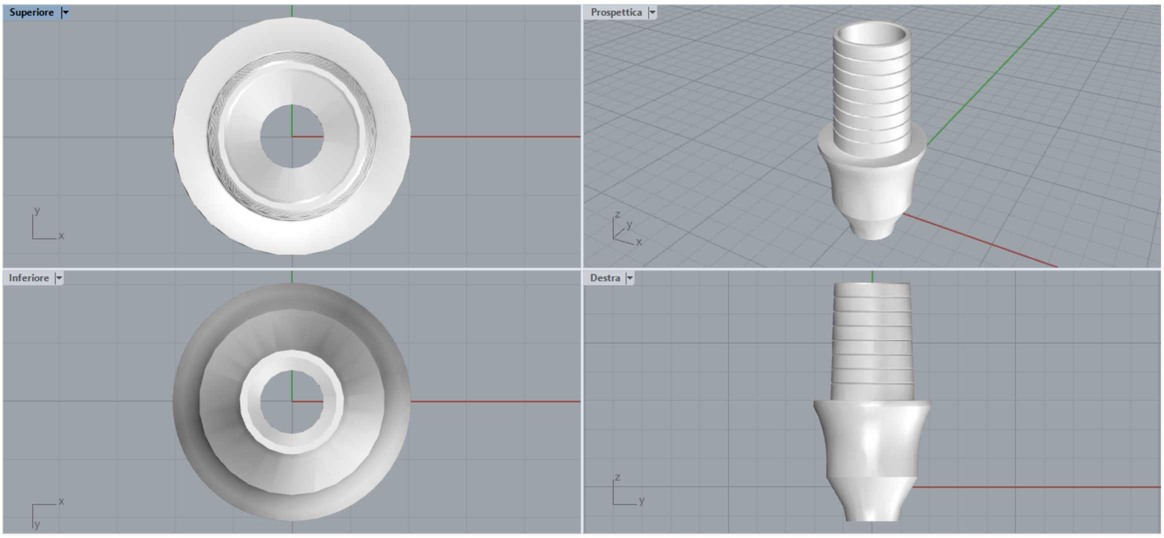


Figure 2.12 (b) Rotational T-Base

2.4.7 Endosseous screw

An endosseous screw is a type of dental implant used to support dental prostheses. It is a small screw-shaped device typically made of titanium or a titanium alloy (Ti6Al4V is the most used) to match with the characteristics of the bone. Endosseous screws are surgically implanted into the upper or inferior jawbone to serve as artificial tooth roots.

The term endosseous refers to within the bone, indicating that these screws are placed directly into the upper or lower jawbone. Over time, the bone fuses with the surface of the implant through the osseointegration process, providing a stable foundation for the dental prosthesis.

Endosseous screws are commonly used in modern dental implant procedures due to their high success rates and durability. In Figure 2.13 (a) a CAD project of endosseous implant. It provides two threaded bodies, one in direct contact with the jawbone and its function is to bind with the bone promoting the osseointegration process. The rounded off screw pitches assure no tearing of the graft.

Inside the screw, there is another thread which serves to ensure fixation with the prosthetic. The fixation screw will lock with the endosseous screw to guarantee the total fixation and support to the prosthetic implant.

Thus, in Figure 2.13 (a), it can see a Sterolitography (.Stl) file on Rhinoceros 3D of a sectioned endosseous screw where it is possible to see the two threads and with the internal hexagonal part. It means the coupling, between the connection prosthetic and the endosseous screw, gets done inside the screw cavity.

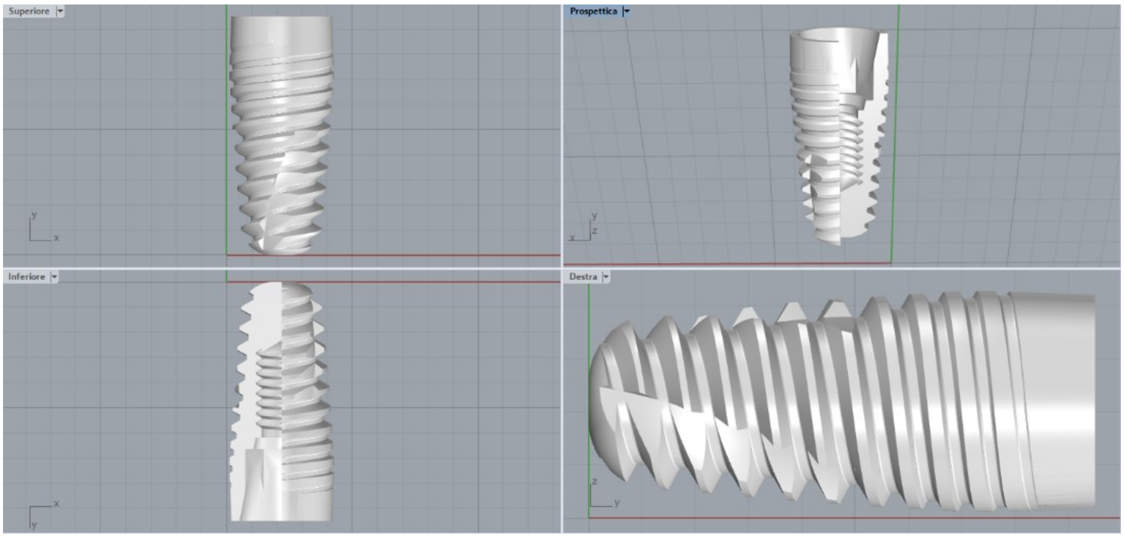


Figure 2.13 (a) Endosseous screw with internal hexagonal form

Instead, in Figure 2.13 (b) there is a CAD version of an endosseous screw with external hexagonal form. In this case, the connection prosthetic and endosseous screw match outwards.

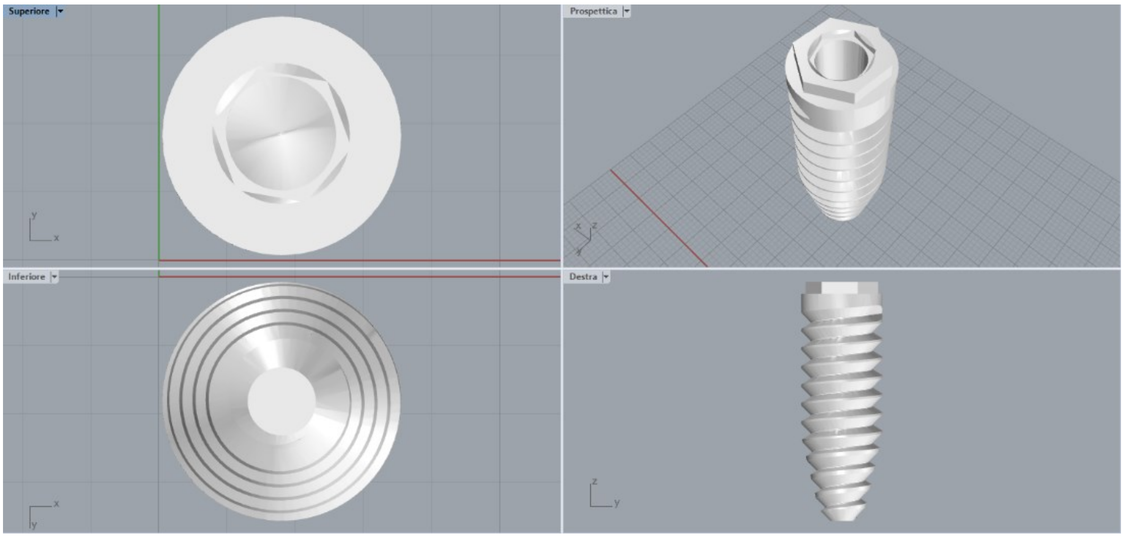


Figure 2.13 (b) Endosseous screw with external hexagonal form

Since the introduction of the osteointegration concept and the titanium screw in 1952 with Dr. Per-Ingvar Brånemark, the titanium screw has become the most popular in world today. The shape of this components can be with a thread, cylindrical or even cone shaped. The titanium can be pure or with a rough surface which increase in surface area, and more retention in the bone.

So, the rough area can be done with plasma-sprayed titanium coated, hydroxapatite coated, or titanium spherical grains coated.

2.4.8 Fixation screw

A fixation screw is a type of medical device used in implant surgery to secure the artificial abutment or the body tooth (i.e. the artificial crown) onto the endosseous screw. They provide stability and support, allowing for proper healing and function.

Fixation screws are typically made of biocompatible materials such as titanium or stainless steel. These materials are well-tolerated by the body and are designed to withstand the forces and stresses encountered during healing and normal movement. In Figure 2.14 the CAD project on Rhinoceros 3D of the fixation screw.

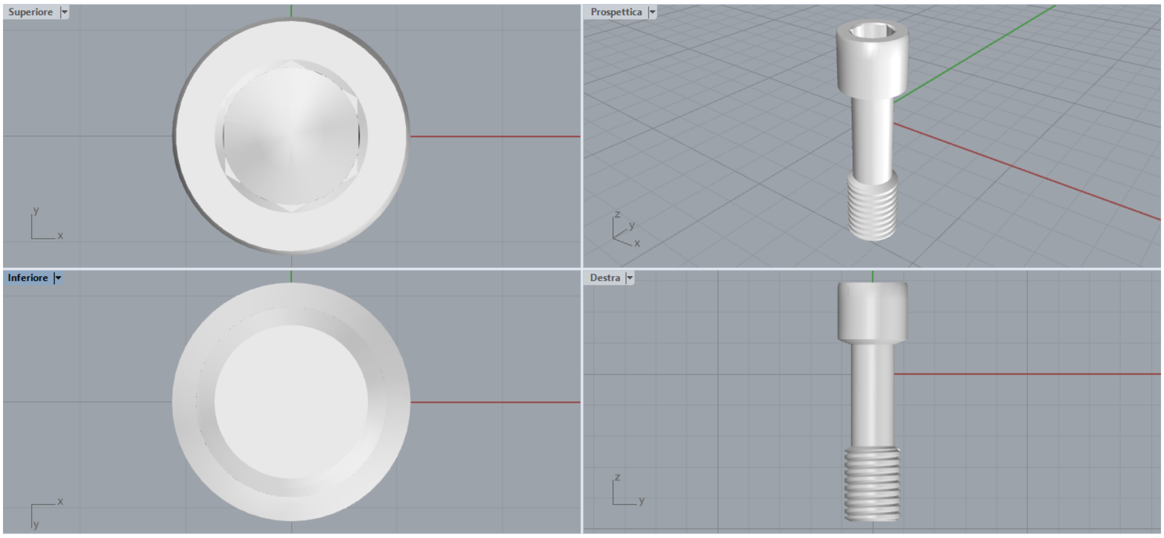


Figure 2.14 Fixation screw

In Figure 2.15 the section CAD project of the assembly, the endosseous screw (green), the crown with connection prosthetic (yellow), and the fixation screw (grey) among the endosseous screws and prosthesis which fixed the prosthesis.

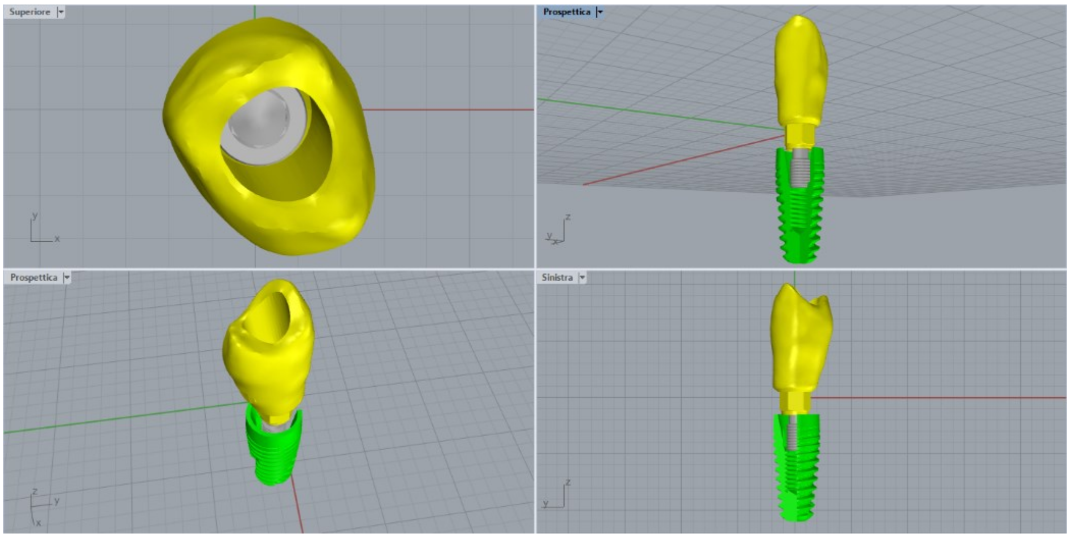


Figure 2.15 Endosseous screw-prosthesis coupling

2.5 Juxta-osseous implant

Juxta-osseous implant is not a standard fastener but custom-made device, designing for a single patient based on its cranial basal bone. The need to make this kind of implant is for some pathologies as for example the severe partial or total atrophy of maxille and mandible.

The upper partial juxta-osseous implant with or without pterygoid extension is an alternative solution to the zygomatic, pterygoid, tuberositary implants. Otherwise, to the sinus lift if the bone availability is less than four millimetres and if there are local contraindications or risks in the procedure. The partial juxta-osseous technique allows for re-operation if other procedures have been failed.

The lower partial juxta-osseous implant is indicated for the lower basal bone, that is the jaw, and/or as a replacement for short-implants (4-6 mm) where regenerative techniques are not predictable and burdened by significant failures in the short-medium term.

Partial juxta-osseous implants are indicated only for the replacement of a resorbed edentulous area of at least three teeth. [6]

If, previously, the technique was adopted only by very expert operators, digital technology has contributed to simplifying the procedure and making it more usable. Based on an analogue or digital model of the basal bone, the expert programs a design respecting the anatomical sites not subject to further resorption, so the device can be created in fusion or 3D printing (i.e. with 3D printer or selective laser melting).

Chromium-cobalt-molybdenum (Vitallium), titanium alloys (e.g. titanium-niobium or titanium-aluminium-vanadium) are the materials most tested for their biocompatibility and for resistance to peak and cyclic compressive and tensile loads.

The juxta-osseous implant can be posed to the bone in these ways: direct bone contact and held to it by sharpey fibers (i.e. matrices of connective tissue), or fibrous contact, it can be integrated into the bone and enveloped in the deep periosteum and retained by sharpey fibers. Clinically, no modality has proven superior to another; what matters is that the implant maintains its stability without signs of suffering and inflammation.

A CAD representation of juxta-osseous implant is in Figure 2.16.

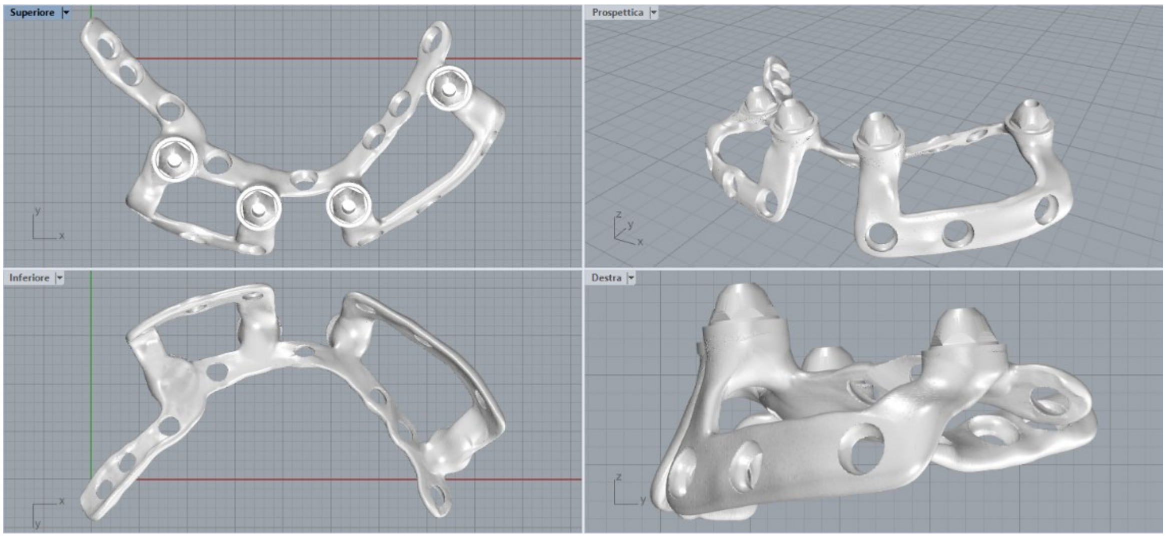


Figure 2.16 Juxta-osseous implant

2.6 Standard surgical procedure

Despite advances in preventive dentistry, edentulism is still a major public health problem worldwide. Edentulism remains a major disease worldwide, especially among older adults.

The artificial implant has made with different materials due to the functional and mechanical properties requirements. The dentistry implantology field is very similar to the orthopaedics one. The concept of such procedure is the surgical placement of the implant into the jawbone to substitute a/more missing tooth/teeth and its root(s); very similar, for example, to hip and shoulder arthroplasty surgical interventions.

Historically, in 1952, Professor Per-Ingvar Brånemark, a Swedish surgeon, while conducting research into the healing patterns of bone tissue, accidentally discovered that when pure titanium comes into direct contact with the living bone tissue, the two grow together to form a permanent biological adhesion, in medical jargon the "osseointegration" process.

The dental implant aims to increase the support, stability, comfort in chewing, aesthetics, overall psychological health and providing support for a partial denture. The implant can be of various type: juxta-osseous, endosseous and trans-osseous (this latter is not dealt in this thesis, and it is the more invasive because penetrate the entire jaw).

It is important to have standardized surgical procedures because the risks the patient faces are the same as in other medical fields, particularly the sterilization of instruments, and the procedure phases are:

- Preparation: the patient undergoes pre-operative evaluation, which may include medical history review, physical examination and possibly imaging tests like X-rays or CT scans to assess the area where the implant will be placed;
- Anesthesia: dentist administers local anaesthesia so numbing the specific area or general anaesthesia only in hospital (rendering the patient unconscious);
- Incision: the oral surgeon makes an incision at the site where the implant will be inserted. The size and location of the incision depend on the type of implant and the surgical technique being used;
- Implant placement: dental implant (in this case the endosseous screw) are placed into the jawbone with a very precise surgical procedure. The implant remains covered by gum tissue while fusing to the jawbone.
- Implants uncover: after approximately six months of healing, the implant is exposed, and a healing post is placed over top of it so that the gum tissue heals around the post;
- Prosthetic phase (teeth): once the gums have healed, an implant crown is fabricated and screwed down to the endosseous screw;
- Follow-up: patients typically have follow-up appointments with the doctor to monitor healing, check for any complications and ensure the implant is functioning properly. Moreover, the patients likely must follow an antibiotic therapy.

It is important to note that specific details of the surgical procedure can vary widely depending on factors such as the type of implant being placed, the patient's overall health, and the surgeon's preferences and experience. In Figure 2.17 depicted the procedure steps.

The implant survival obtained with the use of guided implant surgery shows high percentages, which means increasing the technology in the clinical process. Many recorded failures occurred early, due to a lack of osseointegration, or errors in the procedure, and the variables that come into play in the survival of the implants are many. [7]

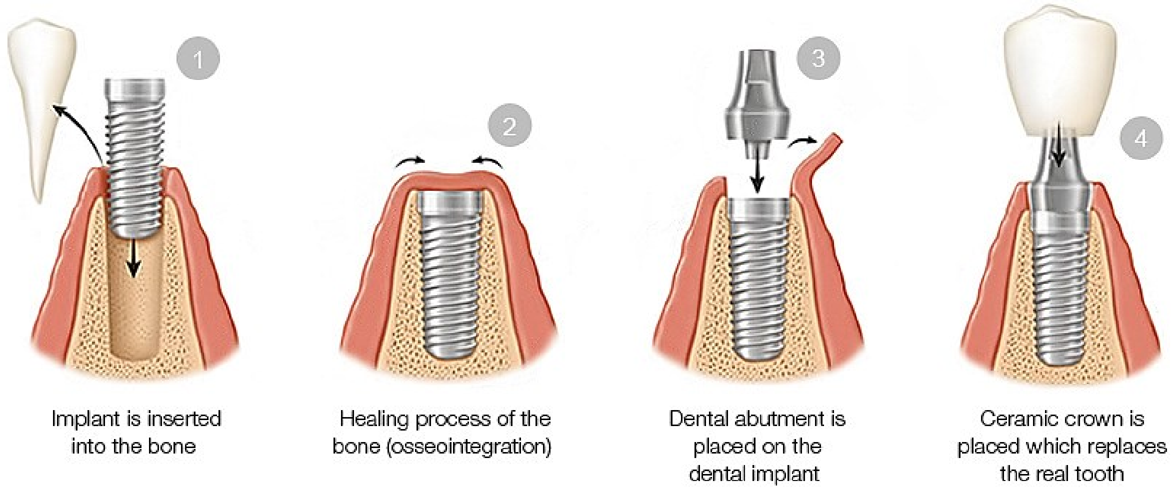


Figure 2.17 Surgical procedure steps for dental implant prosthesis

2.7 Oral scan

An intraoral scan refers to a digital scan of the inside of a patient's mouth, particularly of their teeth and surrounding structures. This technology is commonly used in dentistry and orthodontics for various purposes such as creating digital impressions for crowns, bridges, and orthodontic treatments like clear aligners.

Traditionally, dental impressions were made using a putty-like material that patients would bite into (e.g. alginate), creating a mould of their teeth. However, intraoral and extraoral scanning technology has revolutionized this process by allowing dentists and orthodontists to capture highly detailed, three-dimensional geometry of the patient's teeth and gums using a handheld intraoral scanner.

These scanners typically use either optical or laser technology to capture the images, which are then converted into digital models that can be manipulated, analyzed, and used for treatment planning. Intraoral scans offer several advantages over traditional impressions, including increased accuracy, faster times and greater comfort for patients as they eliminate the need for impression materials. Intraoral scanning saves considerable time in post processing when compared with conventional dental impressions.

Overall, intraoral and extraoral scanning have become an essential tool in modern dentistry, improving the efficiency and precision of various dental procedures while enhancing the overall patient experience. Moreover, the dental's factories job, which produce complex implants, becomes easier and more accurate due to the passage to digital form. So intraoral optical scanner has become an essential tool in implant manufacturing. There are scanner of different types and size, they can be handheld or over the counter (i.e. fixed). The handheld one can be managed by the odontologist inside the patient's mouth, whereas fixed scanner or scan-desk is greater in dimension and the operator applies the dental cast in a support and the machine digitalizes, semi-automatically, the images taken from different angles.

In conclusion, the resolution of the intraoral scanner is primarily defined by the system hardware and optimized for default scans. Significant differences of the various models in terms of scan time and number of images captured per scan. A software high-resolution mode that obtains more data over a longer time may not necessarily benefit the scan accuracy, while the tooth preparation and surface parameters do affect the accuracy. [8]

2.7.1 Scan body

The scan body or also called scan abutment is a tool that allow the operator to obtain a precise and detailed overview of the position of the implant, waiting the prosthesis to be created.

This device takes over of the artificial crown and abutment when the dental technician has to design the prosthesis from the model. Then, every scan body must be designed for each connection of the abutment and so the analogous. It must have a simple shape to be easily digitized by the scanner. [9]

All manufacturers develop different shapes. The Yndetech, as host company for the internship of this master's degree thesis, has its model. In Figure 2.18 the CAD project developed on Rhinoceros 3D of the scan body of an 'Alphabio TCT 2.5'.

As previously seen above, in Figure 2.11 (b), the assembly of the scan body with the analogous. The circular plate on top represents the height of the scan body and this value must be very precise and coherent with the implant. At scan moment, the dental technician use as many scan bodies as there are implants in the model.

For this purpose, it is very important that they are all the same heights, because if one or more have discrepancy in heigh the prosthesis shows a certain instability if loaded on one side, called in jargon ‘finger test’. Logically, when the prosthesis match with implant has not to move. In the images there is an empty space between the upper part and lower part of the scan body, because the geometry changes among different models.

Another geometrical important aspect regards the flat faces of the scan body. In fact, they must be aligned with the anti-rotational surfaces part of the connection prosthetic, if it is irrotational.

The relationship of the dental technician and the prosthesis manufacturing company is fundamental to have a good product. In effect, the dental technician creates with impression material the patient dental cast. In according with the oral surgeon, it decides where to place the implant(s), in geometry terms (e.g. inclination, height etc.). Once the cast with the fictitious implant is ready, the dental technician can perform the semi-automated scans with the scan bodies.

So, it creates a .Stl file domain which will be sent to production department that create the prosthesis in the declared material.

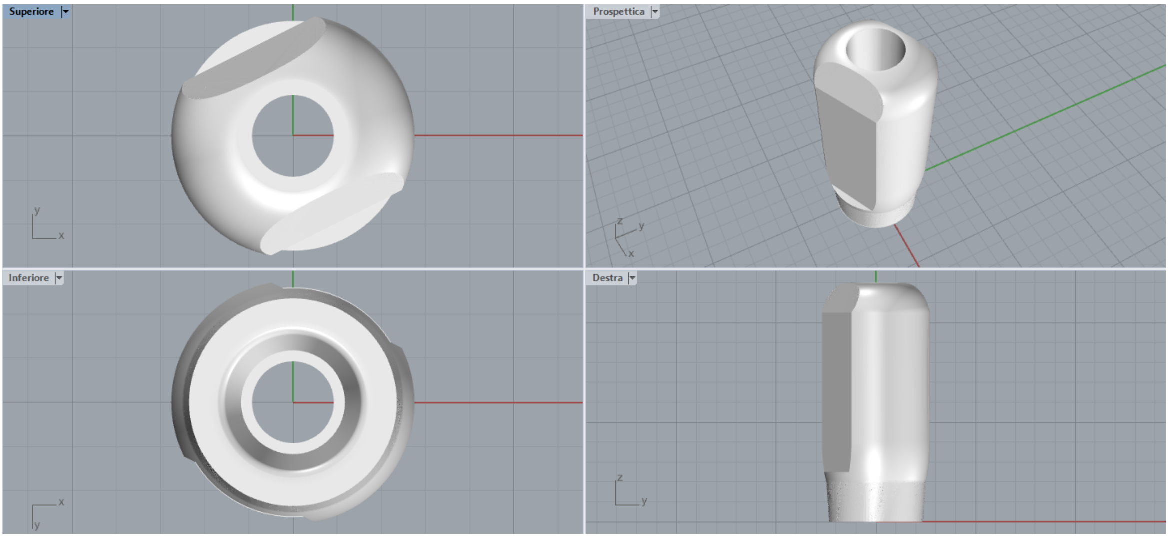


Figure 2.18 Scan body of Alphabio TCT 2.5

2.7.2 *Medit i700*

It can be wireless or wired intraoral scanner. It is portable and handheld having a very comfortable usage for both the patient and the doctor. The wired version can be directly linked to the PC by using only a USB cable. In Figure 2.19 the image of the device and in Table 2.1 the technical specifications. The distal molar area is simpler to detect with respect the oldest techniques.

It digitalizes the scanned object, and this device has also an own software to load and work the files that is called 'Medit App'.

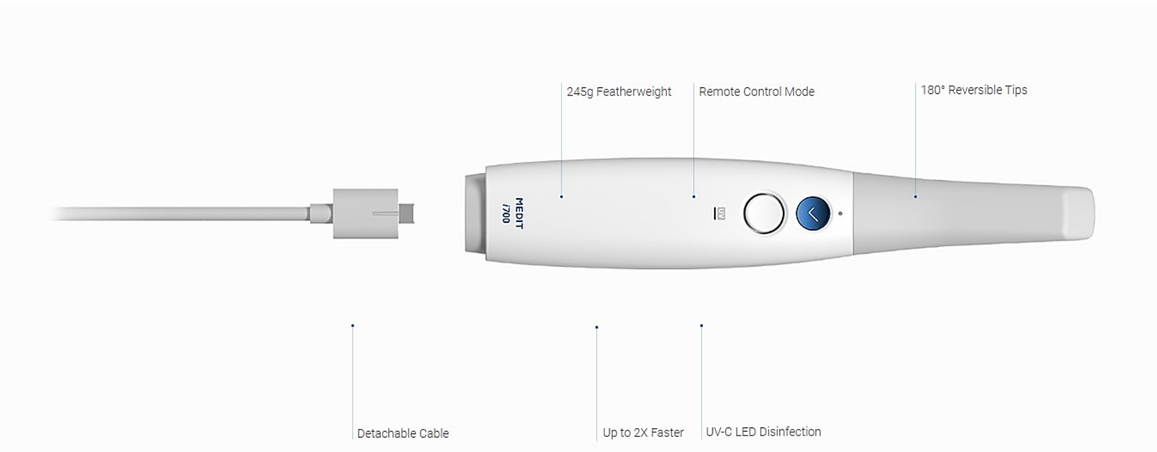


Figure 2.19 Medit i700

Table 2.1 Medit i700 technical specifications

Technical specifications	Parameters
Scanning frame	Up to 70 FPS
Imaging technology	3D full color capture
Light source	LED
Anti-fogging technology	Adaptive anti-fogging
Full arch	10.9 µm ± 0.98
Dimensions	248 x 44 x 47.4 mm
Weight	245 g
Tip size	22.5 x 17.1 mm
Scan area	15 x 13 mm
Autoclavable	150 times
Reversible tip	Yes
Remot control mode	Yes
Cable length	2.0 m
Connectivity	USB 3.1

2.7.3 *Medit T710*

It belongs to the 'T-Series Lab Scanners' of said company. It is a bench scanner with blue-light scanning technology. It has four high-resolution cameras with 5 Megapixel which guarantee very accurate performance and images sharp and detailed, eliminating any blind spots. The operator scans the dental mould of the upper or lower arch, or entire mouth. Physically, the mould is placed on a plate that is rotated and inclined by a robot arm.

The Medit T710 (Figure 2.20) desktop scanner is equipped with a fast scan engine and highly efficient software algorithm which work in tandem to produce a full-arch scan in just eight seconds. The advanced, high-speed positioning system of the new 'T-Series' is designed for optimal performance, speeding up the workflow and increasing productivity.

Auto-elevation feature permits to the scanner to decide the scanning height for the object. Moreover, it can scan more object contemporaneously. The four cameras in the T710 are positioned in a way to ensure that there are no blind spots in your scan data.

It creates a .Stl file extension to work it in all CAD and CAM software. It is very reliable in fact it adheres to strict international standards, including ISO 12836, ANSI/ADA standard No. 132 and VDI 2634.

The other scan system uses three or five steps to capture the object, but this latest generation scanner works just with two scan steps.

It is also provided of some articulators where the dental arch can be fixed to simplify the design phase on CAD software. At last, it can be combined with the Medit i700 to have the maximum possible results. On Table 2.2 the technical specifications of this device.

Figure 2.20 Medit T710

Table 2.2 Medit T710 technical specifications

Technical specifications	Parameters
Point spacing	0.040 mm
Scan area	100 x 73 x 60 mm
Scan principle	Phase-shifting optical triangulation
Size	505 x 271 x 340 mm
Weight	15 kg
Light source	LED; 150 ANSI-lumens; Blue LED
Connection	USB 3.0
Power	AC 100-240 V, 50-60 Hz
Accuracy	4 µm
Full arch scan speed	8 sec (7 cuts)
Full arch impression scan speed	45 sec

2.7.4 Ynde.SCAN^{DUE}

It is a 3D table scanner. The advantageous with respect the other is the competitive price and excellent performances. It is very fast and can perform single or multiple scans.

Moreover, Ynde.SCAN^{DUE} includes a wide range of accessories to optimize work both in terms of time and scanning quality as for example calibration plate, plate for multiple scans up to seven abutments or plate for multiple implants and many others. On Figure 2.21 the device and in Table 2.3 the technical specifications.

Figure 2.21 Ynde.SCAN^{DUE}

Table 2.3 YNDE.SCAN^{DUE} technical specifications

Technical specifications	Parameters
Cameras	2 x 2 MP
Accuracy	5 µm
Light source	Structured light
Size	395 x 400 x 275 mm
Weight	13 kg
Scan method	Swing method
Scan working	Interproximal scan of plate
Output format	STL, OBJ, OFF
Power supply	AC 100-240 V, 50-60 Hz

2.8 CAD and CAM software for dental applications

Remembering the acronyms, Computer-Aided Design (CAD) and Computer-Aided Manufacturing (CAM) are software which serves to improve the design and creation of dental restorations especially dental implant and prostheses, including all components of the previous chapters.

Furthermore, CAD is the use of computers to aid in the creation, modification, analysis, or optimization of a design. This software is used to increase the productivity of the designer, improve the quality of design, improve communications through documentation and to create a database for manufacturing. CAD output is often in the form of electronic files for print, machining, or other manufacturing operations. Usually, the file format is Stereolithography (Stl) because .Stl extension domain can also be used for interchanging data between CAD and CAM systems.

Hence, CAD software is used by engineers to create precise 2D and 3D drawings and models. There are several CAD software packages available, each with its own features, capabilities, and target industries.

CAD is one part of the whole digital product development activity within the product lifecycle management processes, and can be used together with other tools, which are either integrated modules or stand-alone products, such as Computer-Aided Engineering (CAE) and Finite Element Analysis (FEA), and CAM which includes instructions to computer numerical control machines.

So, CAM is the use of software to control machine tools in the manufacturing of work pieces. This is not the only definition for CAM, but it is the most common. It may also refer to the use of a computer to assist in all operations of a manufacturing plant, including planning, management, transportation and storage. Its primary purpose is to create a faster production process and components and tooling with more precise dimensions and material consistency, which in some cases, uses only the required amount of raw material, reducing energy consumption.

Thus, CAM is a subsequent computer-aided process after CAD and sometimes CAE, as the model generated in CAD and verified in CAE can be input into CAM software (a .Stl file format) which then controls the machine tool.

The first CAD and CAM system used in dentistry was produced in the 1970s by Prof. François Duret and colleagues. The process contains several steps. Firstly, an optical impression of the intraoral environment was obtained by scanning with an intraoral scanner. The digitized information was transferred to the monitor where a 3D graphic design was produced.

The restoration could then be designed on the computer, and the final restoration was processed with milling or additive manufacturing techniques. To date, technology and programs have improved and many different and increasingly specific software have been created.

The advent of CAD and CAM technology has brought much innovation to dentistry. In dental surgery, orthodontics and others field of application, this technology has been incorporated during diagnosis and treatment plans.

Moreover, CAD and CAM technology facilitates a fully digital workflow; various protocol studies have postulated clinical reliability, and it has been indicated that this technology produces very favourable feedback from patients. [10]

The CAD and CAM technology allows the delivery of a well-fitting, aesthetic and durable prosthesis for the patient. These technologies are used to increase the speed of design and creation, increasing the convenience or simplicity of the design. Also, to make possible restorations and prosthesis that otherwise would have been infeasible. Other goals include reducing unit cost and making affordable restorations and appliances that otherwise would have been prohibitively expensive.

The CAD and CAM technology does not specify the method of production except that whatever method is used takes input from the software CAD to CAM, and today additive and subtractive methods are both widely used.

The process consists of capturing non-digital data, then converted into a digital format by means the scanner, subsequently converted back into a physical form with the exact dimensions and materials specified during the digital design process, usually by either 3D printing, selective laser melting or milling. This set of stages is known as a "digital workflow".

The application in dentistry field provides means of fabricating dental prostheses that are used to restore or replace teeth. This is an alternative to the traditional process of prosthesis fabrication using physical techniques, in which the dentist makes an impression of the site that is to be restored. This is then transported to the dental technician laboratory where a model is made.

An imitation of the final design is made using wax which represents the size and shape of the finished dental prosthesis. The wax is then encased in a mould, burned out and replaced with the desired material as part of lost wax casting.

Hence, CAD and CAM process makes such procedures unnecessary. The impression is recorded digitally, and the manufacture of the object is accompanied by additive (i.e. selective laser melting or 3D printing) or subtractive (i.e. milling) technology. [11]

The process consists of a CAD and CAM stage and the key stages can broadly be summarised as the following:

- Optical scanning that captures the intraoral or extraoral condition of the patient;
- Use of software that can turn the captured images into a digital model so the dental prosthesis and/or components can be designed and prepared for fabrication;
- The conversion of the design into a product by way of 3D printing or laser melting or milling depending on the CAD and CAM system used.

2.8.1 Advantageous of CAD and CAM system

The advantages of CAD and CAM systems with respect the traditional method is the following:

- Allows for use of materials otherwise unavailable in the dental laboratory;
- Provides cheaper alternatives when compared with conventional materials;
- Decreases labour cost and time for dental technicians;
- Standardises the quality of restorations.

Actually, ceramic materials can require a lot of processing time. To make a ceramic dental prosthesis by hand, the technician has to meticulously build up porcelain powder and sinter it onto the surface of the item.

With CAD and CAM, labour times are significantly reduced, only five or six minutes of CAD. In this way, the cost of production is reduced because labour costs are lower.

Furthermore, CAD and CAM systems mill prosthesis from blocks of material, print or melt from powders, again reducing costs for the dental offices and laboratories when compared with traditional techniques.

This technological innovation in the dental field completely revolutionizes the operational vision of dental prostheses in the field of implantology and dental prosthetics.

3 Materials and methods

This chapter shows the method of work followed by Yndetech to manufacture the dental prostheses and its component and how the project of this thesis was performed. Indeed, Yndetech has several departments, from the marketing office to the production department.

Particularly, this thesis is based on the work of the Research and Development (R&D) department and interfacing with other technical offices, as for example the service and design department.

In the next paragraphs the entire process from the digital workflow to a proximal explanation of the manufacturing methods.

3.1 Digital workflow

A digital workflow refers to the process of managing and executing tasks or operations using digital tools and technologies. It involves the use of CAD and CAM software and other type of software (e.g. CAE), automation, and digital communication to streamline and optimize various aspects of a workflow.

A well-designed digital workflow can help the organizations to streamline operations, improve productivity, enhance collaboration, reduce costs, and adapt to changing business needs more effectively.

Next paragraphs show the processing algorithm followed by Yndetech to manufacture its product. Particularly, in this thesis is represented the digital workflow from the intraoral or extraoral scans to the final product, that is the manufacture process but only from a digital point of view and not the processing itself.

Clearly, the digital workflow of the cases of study are too complex to show in this thesis (because the prosthesis was produced in the past), so the explanation of the digital workflow is disconnected with the cases of study even if the steps are, exactly, the same.

The models are created starting from the mouth or teeth of the patients, so this approach regards a reverse engineering process. It is impossible to mass-produce dental prostheses because the products must recreate or replicate the anatomy of the single anatomical morphology of the patient, or to create compatible systems or components to reproduce the exact geometries and functions of lost teeth.

Therefore, the first step in reverse engineering is acquiring data about the object being studied. This involves the 3D scanning, then it needs to be processed and converted into a usable digital format (normally .Stl file domain). Furthermore, the engineers can analyze and modify the design of the digital object in CAD software. At last, the file is loaded on CAM software to be produced by machinery. [12]

3.1.1 Clinic scan

The oral scan paragraphs introduced the concept and the models which can perform this task. This thesis illustrates an intraoral scan with Medit i700 wired. The scan works with two apps ‘Medit link’ that is for the management of the files, and ‘Medit Scan for Clinics’, which serves for the scanning process.

The operator follows five steps to carry out the process by approaching the dental arches with the tip of the device.

First occurs the scan of the upper dental arch, after the scan body(ies), and then the scan of the lower dental arch. The last a scan of the entire occlusal plane (i.e. the bite). In Figure 3.1 the digital platform of the app ‘Medit Scan for Clinics’.

The software can optimize the scansion and it takes all intraoral scans and it makes a mesh, then convert to either .Stl or .Ply file (the last to visualize the colours).

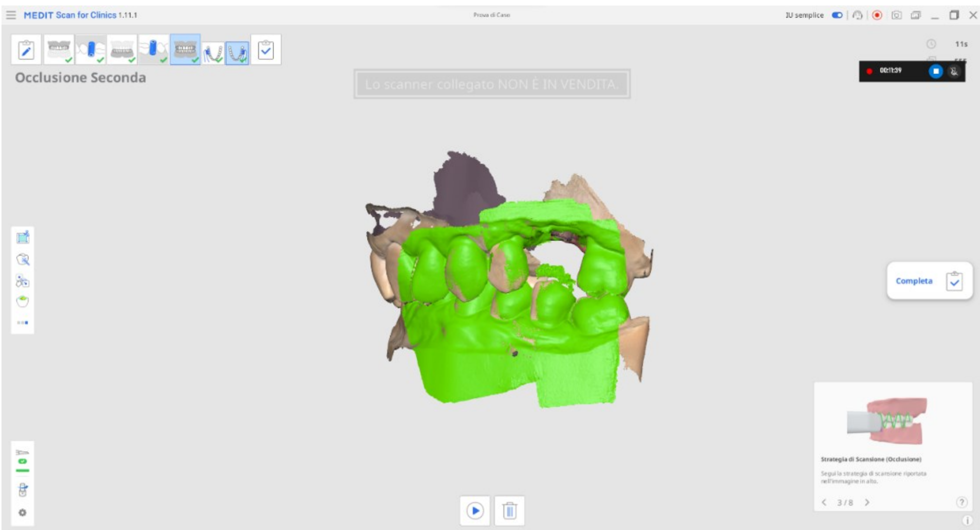


Figure 3.1 Clinic scan of the entire dental arch with Medit i700

3.1.2 *Hypsocad*

It is a dental CAD software which is invented and developed by Yndetech to improve the dental technician's job. It is possible to carry out different tasks, for example to design custom abutment, full denture, implant bar, anatomic crowns and bridges, or implement commands as reduced crown and bridges and so on.

In the digital workflow it represents the second step, from 'Medit link', where the .Stl file is imported directly on Hypsocad.

First, some data are inserted into the software, as for example the number of tooth/teeth which want to be processed and the material to be manufactured as well as the patient's data.

The CAD work is divided in some steps as 'preparation', 'margin line', 'external', 'structure', 'finishing', and 'export'.

Since in this case it is an implant work, the 'preparation' step is necessary to align the scan bodies to coincide a digital library which will have to use. It is already loaded into software, of the proper either the irrotational or rotational implant that is required by the dentist.

Then, the further step is the 'margin line', it is about the transmucosal passage, so the line in contact of the crown with the gum. Hence, the design of the crown or some dental components to avoid problems with the gum so the operator can modify the geometry of aforesaid line to find the right position.

After, the principal phase of the Hypsocad process, called 'external', it permits to model the crowns, the outermost and visible part of the prostheses.

Often, the crowns are manufactured with metals or ceramic (i.e. cobalt-chromium, titanium or zirconium dioxide) but forgiveness in aesthetics. The dental technician must cover the entire prosthesis with a ceramic material and so it is necessary to erode the natural stump, this task is carried out by Hypsocad. Subsequently, the prosthesis can be cemented or screwed.

In 'structure' step the screwed prosthesis involves the making of the channel where the screw passes. The final step is the 'export' of the file, the project is sent to the production centre to pass in a CAM software and be produced in the machinery.

Some problem can arise if the client wants a prosthesis where it is needed a decryption because it is an implant screwed and uncemented; in this case the file before passes on R&D Yndetech department to decrypt and so, to lay the proper geometry of the prosthetic connection, but it is said in the next section.

In the bottom of the Figure 3.2 there are the various step, which some are mandatory and others optional.

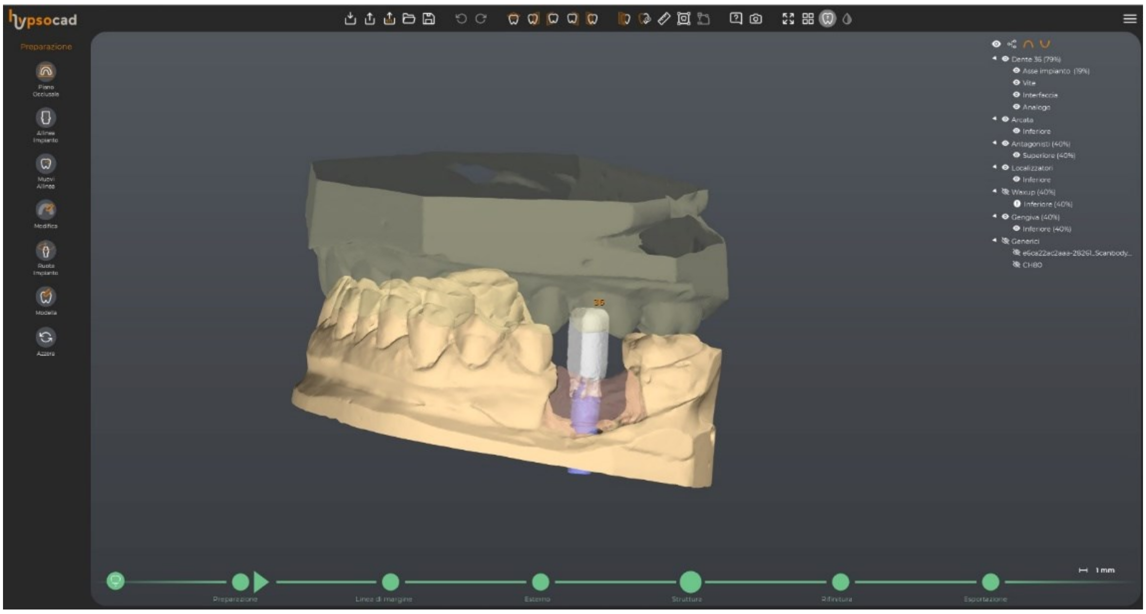


Figure 3.2 Hypsocad interface

3.1.3 Rhinoceros 3D for geometric decryption

Rhinoceros 3D is a well-known CAD software, a freeform surface modeler that utilize the 'Nurbs' mathematical model.

As for regards the digital workflow, Yndetech utilizes this software to decrypt the geometry of the connection prosthetic which go inside the endosseous screw. It is done because each factory that produces implants must "protect" its items in such a way that customers use their products.

So, Yndetech uses Rhinoceros 3D in pre-manufacture phase, to design and match the proper geometry of the implant. Engineers must decrypt the actual geometry and substitute with the proper geometry to fit to the patient's implant.

The interface of Rhinoceros 3D (Figure 3.3) to decrypt the geometry of the prosthetic connection with the endosseous screw is the last step before the production. Indeed, this process restores the right geometry to fit it on endosseous screw, it is possible to see in the right of the image the entire digital library of all geometries of each different implant factory. Selecting the proper connection prosthetic, it appears on the origin of the software reference frame (the green component on the Figure 3.3), which will substitute the crypto geometry.

Once it has done, the file can pass on CAM software to manage the production. In this case, the client wants to fit an irrotational prosthetic connection of brand 'Megagen Anyone AR1'. This process occurs only if the dental technician must work in a screwed implant and not cemented.

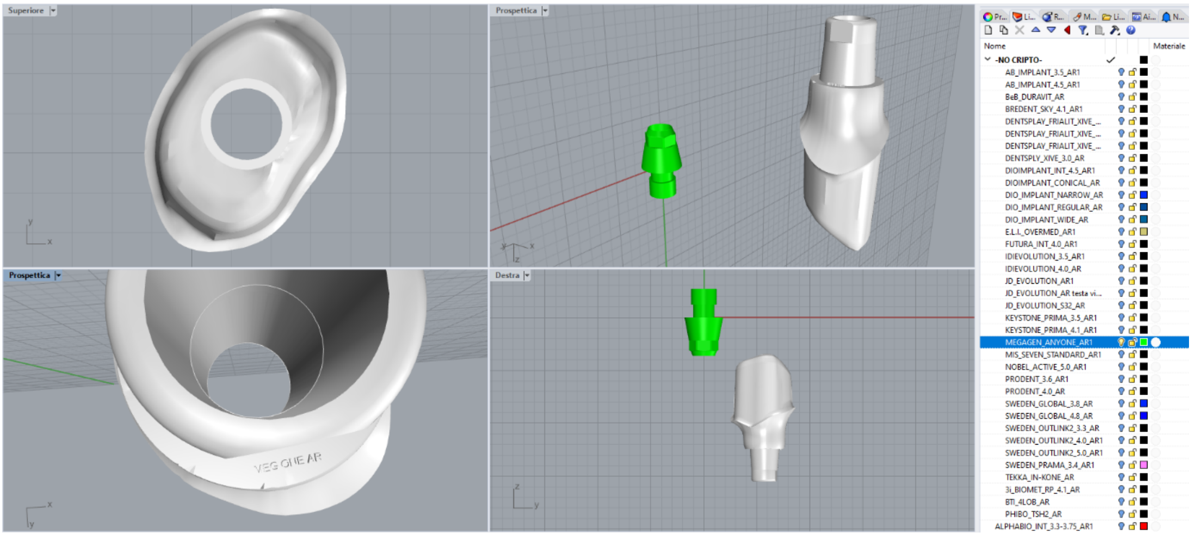


Figure 3.3 Interface of Rhinoceros 3D to decrypt the geometry

3.1.4 HyperDent and Oqton as CAM software

Speaking about CAM programs, HyperDent is a dental CAM software developed by 'Follow-Me Technology' (the interface on Figure 3.4). It is designed specifically for dental professionals, and it is used by Yndetech for the milling process.

HyperDent enables users to design and management dental restorations digitally, such as crowns, and implants. The software utilizes advanced algorithms and tools to streamline the design process, allowing for precise customization and efficient production of dental prosthetics.

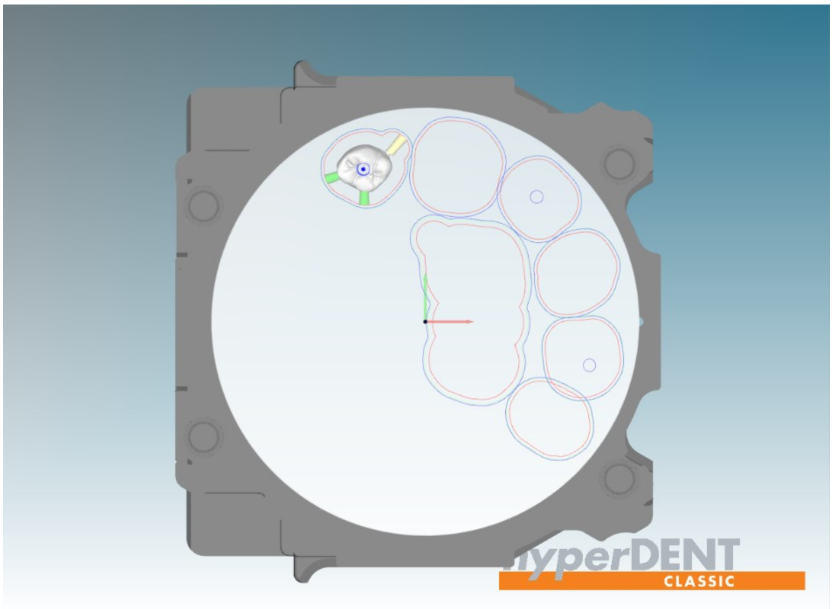


Figure 3.4 HyperDent interface for Milling

As it can see on Figure 3.4, the crown is manufactured by loading the project done, in the previous phases, on the program. This is the last step to obtain the finished item. The dental technician sets the object in a disk called "wafer", also defines and controls the path of the milling machine.

Logically, the engineers, with the production development, decide and check the manufactured process. If the prostheses want to mill the dental technician use HyperDent, but if the objects are made with SLM the software is Oqton.

Thus, Oqton permits to develop and manufacture the files from the designing department to the production centre for SLM. On Figure 3.5 the interface of the software.

On Image it is possible to see several kinds of prostheses in the disk which will be produced. Each prostheses have a tag (in red) to identify the products. In this phase the role of engineer is fundamental in terms to place and supervise the canals (in green) that support the items during the process. This step is fundamental in terms of product accuracy.

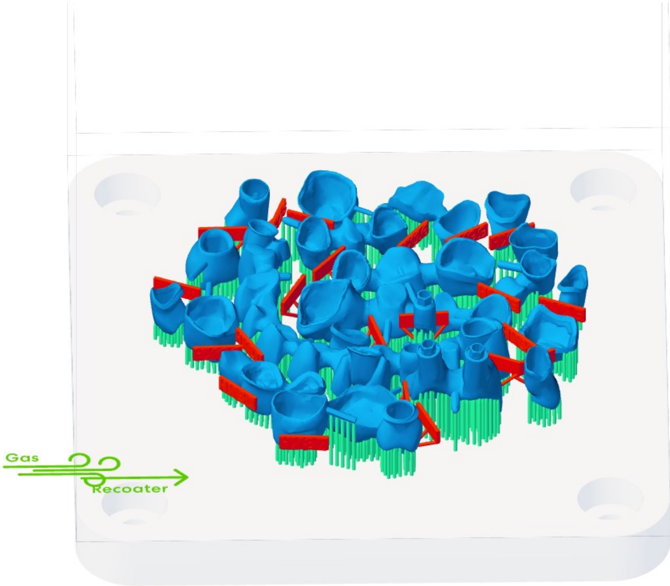


Figure 3.5 Oqton interface for SLM

3.2 Production techniques of dental prosthesis

The production department is the last step in the product processing chain, and it predicts the utilize of several machines and processing techniques. Yndetech has several industrial machines which are the following:

- Cobalt-Chromium (CoCr) SLM;
- Titanium SLM;
- Composite resins 3D printer;
- Zirconium dioxide milling machine;
- Polymethyl methacrylate (PMMA) milling machine;
- PEEK milling machine;
- Lithium disilicate milling machine;
- Glass and carbon fibers milling machine;
- Metal milling machine;
- Autoclaves.

Overall, both milling, selective laser melting, and 3D printing offer unique advantages and considerations in terms of mechanical properties. The choice between these processes depends on factors such as part complexity, material requirements, production volume, costs, and desired mechanical properties.

3.2.1 *Selective Laser Melting (SLM) for dental prosthesis*

The SLM is a process characterized by 3D printing technology. It represents the best solution for creating geometries complex than any other manufacturing method.

The laser melts the metal powder of either titanium or CoCr layer by layer until it is created the three-dimensional body, by allowing the creation of prosthetic products with the highest precision, and at the same time having a completely digital flow.

SLM permits to reduce the price of the product significantly, by avoiding the use of traditional instrumentation and its limits of accuracy.

The factory aims to have the highest level of quality that allows to apply it technology in all branches of dentistry. On Figure 3.6 the final product machining with SLM that contains different prostheses from the single crown to an entire dental arc.

The 3D printing of patient-specific surgical prostheses could be effective in dental and jawbone reconstruction, in addition surgical procedures are simplified. The reconstructions are achieved with high accuracy, and long-term results size are warranted. [13]

With this technique can be made almost all dental products on the market.

Figure 3.6 CoCr finished SLM prostheses

The mechanical properties of laser-melted parts can vary depending on factors such as powder composition, processing parameters, and post-processing treatments. A potential problem with laser-melted parts is the presence of porosity, which can affect mechanical properties such as strength and fatigue resistance. It can exhibit anisotropic mechanical properties, particularly along the build direction.

Hence, SLM 3D printing technology was able to reproduce the customized implant designs and produce high density and strength and adequate dimensional accuracy. Great stress distribution and low maximum micromotions were observed. [14]

3.2.2 Milling for dental prosthesis

Yndetech milling centre carries out processing with high performance multi-axis machines to ensure maximum precision on a wide range of materials and types of production.

Therefore, this technique is applicable to different materials, such as PMMA, lithium disilicate, PEEK, Glass and Carbon fibres, ZrO_2, CoCr, titanium and aluminium.

Instead, the products can be cemented crowns, abutments, screw-retained crowns, and implant bridge or implant bar.

The accuracy of the dental element details produced, determine the processes simpler and faster with respect the older technique and a successful outcome, but even less innovative than the SLM. The produced materials had a homogeneous and dense nanostructure by considering zirconia milling.

The clinical significance regarding milling technique is that the properties of dental restorations produced from zirconium dioxide of hybrid composition by milling technology anticipate satisfactory performance in oral cavity. [15]

The purpose of this machine is bilateral, it can be used for the metal milling (titanium or CoCr) and to refine some part with a high level of accuracy (Yndetech machines have probes to measure the accuracy of the parts), as for example the part of the connection prosthetic which will be in contact with either the endosseous screw or its analogous.

Milled parts typically exhibit high strength and can produce excellent surface finishes, but they are not free from micro-structural defects, as it will see below. This process can achieve high dimensional accuracy and tight tolerances also an anisotropic behaviour.

On Figure 3.7 (a) on left, there are PMMA milled items, whereas on Figure 3.7 (b) on right, the CoCr milled parts.

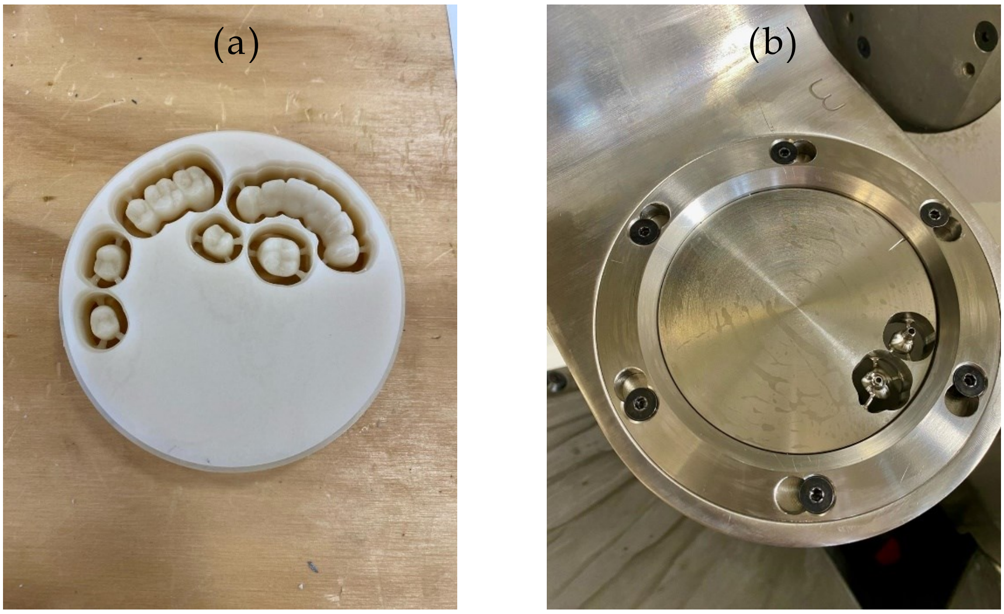


Figure 3.7 (a) PMMA crowns milled; (b) CoCr crowns milled

3.2.3 Resin 3D printer for dental prosthesis

A resin 3D printer, also known as a stereolithography (SLA) or digital light processing printer (DLP), is a type of 3D printer that uses liquid composite resin as its printing material. They may be castable or biocompatible resins.

This resin is photosensitive, meaning it solidifies when exposed to a specific wavelength of light, typically ultraviolet light (UV).

The printer begins the printing process by shining UV light onto the surface of the resin tank, where the first layer of the object is formed. The light selectively solidifies the resin, tracing out the shape of the first layer.

After the first layer is solidified, the build platform moves down by a small distance, typically around 25 to 100 μm, depending on the printer's resolution.

This allows the resin to flow over the solidified layer, covering it with a new layer of liquid resin. The UV light is then projected again, solidifying the resin according to the next layer of the object. This process repeats layer by layer until the entire object is formed.

Resin 3D printers are known for their ability to produce high-resolution prints with smooth surfaces and fine details.

This kind of product are primarily for aesthetics use, as for example bite or full denture. The materials most common are acrylonitrile-butadiene styrene and polylactic acid.

3.3 Chewing loads

Occlusal loading during clenching and biting is achieved by the action of the masticatory muscles and forms the basis for the evaluation of the functional and mechanical performances of prosthodontic and maxillofacial components. [16]

The maximum voluntary bite force (MVBF) can be used to assess the performance of either dental implants or prostheses in an extreme scenario. However, given the complex kinematic nature of mandibular movement, bite forces occur not only in the vertical direction but also in the transverse directions, particularly in chewing or bruxism.

When considering the total force exerted by the mandible, three-dimensional force measurements may ultimately enrich the diagnoses of masticatory pathologies and the use of mouth guards and the design and assessment of dental prostheses. Moreover, the magnitude of the forces is useful to analyze the resistance of dental prostheses.

Whereas total bite force is an indicator for overall masticatory functional performance, forces experienced by individual teeth should be considered when designing partial or full dental implants.

Occlusal force varies over different regions of the dental arch. Furthermore, the per-tooth or local occlusal force magnitude has been identified as one of the main parameters for dental implant fractures. Therefore, dental implants should be rigorously tested at the design stage to assure robustness and longevity while maintaining masticatory performance.

The use of virtual environments may be beneficial to achieve the answers to the structural problem, since implant designs may be rapidly analyzed and tested under a variety of occlusal conditions. As such, the use of simulations to aid in the virtual design of dental implants has become widespread.

Simulations have the advantage of providing detailed information, over an infinite variety of anatomy and load configurations, where obtaining such detail experimentally in vivo may be difficult, expensive, or impossible to obtain. [16]

The use of point loads to represent occlusal conditions remains widespread. Several studies have investigated the influence of load on tooth implants by applying single point loads to the occlusal surface. Point loads are also used to analyze the stress and strain distributions in the mandible using simplified models or those based on anatomical image scans.

Table 3.1 shows the force values in Newton (N) for all teeth in a healthy condition, and with the MVBF in an age range between 18-65 divided in first molars (16,26,36,46), second molars (17,27,37,47), first and second pre-molars (14,15,24,25,34,35,44,45), canines (13,23,33,43) and central and lateral incisors (11,12,21,22,31,32,41,42). [16]

Table 3.1 Chewing load

Location	Type of load	Force (N)
16,17,26,27,36,37,46,47	Healthy chewing	150
14,15,24,25,34,35,44,45	Healthy chewing	85
13,23,33,43	Healthy chewing	60
11,12,21,22,31,32,41,42	Healthy chewing	40
17,27,37,47	MVBF (18-65 age)	474
16,26,36,46	MVBF (18-65 age)	552
14,15,24,25,34,35,44,45	MVBF (18-65 age)	450
13,23,33,43	MVBF (18-65 age)	226
11,12,21,22,31,32,41,42	MVBF (18-65 age)	155

The force values average is 100 N for a healthy chewing during the normal activities such as eating, swallowing etc., but the Table 3.1 differentiates various values of normal chewing depending on the type of tooth. [17]

These data will permit to study the local stress in the critical areas of the case study prostheses. As for regards the normal chewing scenario, first, the behaviours are analyzed by putting a vertical force of 0 degree (i.e. perpendicular to the occlusal plane), then an inclined force of 45 degree with respect the vertical axis. Just a tilted force of 45 degree is considered because it would be too laborious to analyze each angle of inclination, therefore it is a mediety.

Instead, the values of MVBF by considering the location of each tooth will be studied to validate the ultimate strength of the prosthesis and so to guarantee the total reliability of the product, and to give a view of mechanical properties under extreme conditions.

The maximum stress is achieved for the first molars, for pre-molars and second molars the maximum compressive strength is about the same. Moreover, the canines and incisive represent low values than the previous descripted. However, the incisive have the function of traction rather than a compressive force. Another aspect to consider with the chewing load is the number of chews for each meal. Some studies report the normal number of chews in a range of 15 - 40 times before swallowing. [18]

Hence, by considering a hypothetical value of 30 chews before swallowing each part of food, it means 900 number of bites per day and so 328.500 per year. These values will be involved in case of study research to test the fatigue of the prosthesis during a year or more in a further stress-life simulation.

3.4 CAE Software

As already mentioned, Computer-Aided Engineering (CAE) is the use of computer software to analyze, simulate, and optimize engineering designs and processes. It encompasses a wide range of tools and techniques used by engineers and designers to evaluate the performance, reliability, structural properties, fluid dynamic analysis, electrical and magnetic analysis, and safety of products and systems before they are manufactured or implemented.

CAE software enables engineers to perform complex simulations and analyzes that would be difficult or impractical to carry out using traditional methods in vivo. This kind of software enables engineers to reduce time-to-market.

Interesting CAE areas in dentistry field are the following:

- Stress and strain analysis;
- Multibody dynamics and kinematics;
- Analysis tool for the manufacture process;
- Optimization of the product or process.

Usually, the CAE phase is after the CAD design to check the items and give the approval for the production, so to manage with CAM software. [19]

3.4.1 *Finite Element Analysis (FEA)*

Engineering has not only developed in the field of medicine but has also become quite established in the field of dentistry, especially orthodontics and implantology. FEA is a computational procedure to calculate some phenomena as for example the stress or deformations in an element, which performs model solutions. This structural analysis allows the determination of stress resulting from external force, pressure, thermal change, and other factors. This method is extremely useful for indicating mechanical aspects of biomaterials and human tissues that can hardly be measured in vivo.

The technique divides a complex structure or system into smaller, manageable parts called "finite elements", linked through "nodes". Each element is described by equations that approximate the physical behaviour of the system, by defining the materials used with their physical characteristics (e.g. the Young's modulus etc.), and boundary conditions, it is possible to obtain results such as displacements, stresses and other phenomena. [20]

FEA is a common technique used in CAE software to simulate the behaviour of structures and components under various loading conditions.

Generally, an .Stl file domain is loaded on CAE software. It prefers to study simpler elements and it uses numerical methods to solve for the stresses, strains, and deformations within each element.

It is very important gives to the program an exact geometry of the items, include dissimilar material properties (by loading the materials of prostheses), have an easy representation of the total solution and capture the local effect, so that the engineers can formulate an analysis.

Hence, FEA is a powerful tool in the hands of engineers which allows them to study mechanical properties and behaviours of the elements, in this case dental prostheses, in order to understand medical and mechanical complications.

The aims of this thesis regarding the study with FEA of four-element crown bridge, to value the stresses and deformation on critical areas of rupture, by studying the critical points of the prosthesis and testing some possible improvements that could be made during manufacturing.

FEA modelling is an effective computational tool to investigate the long-term stability of implants. The aim is to transfer such technology into clinical practice to help dentists in the diagnostic and therapeutic phases. To do this, future research should also deeply investigate the loading influence on the bone-implant complex at a microscale level.

This is a key factor still not adequately studied. Thus, a multiscale model could be useful, allowing to account for this information through multiple length scales. It could help to obtain information about the relationship among implant design, distribution of bone stress, and bone growth. Finally, the adoption of a standardized approach will be necessary, to make FEA modelling highly predictive of the implant's long-term stability.

The main purposes of the Finite Element Method (FEM) in the dental sector are:

- Stress and strain analysis: simulate how dental implants behave under different load conditions to ensure that they can withstand chewing forces without breaking or deforming;
- Design optimization: optimize the shape and structure of implants or prosthesis to improve their durability and functionality;
- Mechanical analysis of materials: study the mechanical properties of new materials and how they react under mechanical, thermal and environmental loads to ensure safety;
- Adaptation and comfort: assess how the prostheses fit the patient’s mouth and how they distribute the masticatory forces to avoid pressure points and improve comfort;
- Durability: analyze the duration of prostheses under daily wear and stress conditions;
- Pre-operative simulation: planning complex surgeries simulating different options and scenarios to improve results and reduce risks;
- Surgical guidance: use the results of FEM simulations to create surgical guides, with augmented or virtual reality;
- Movement of teeth: simulate the movement of teeth under the application of orthodontic forces to predict the results and optimize treatment;
- Dental occlusion: study how teeth meet and interact during chewing to identify and solve occlusion problems;
- Temporomandibular joint (TMJ): analyze forces and movements in TMJ;
- Research and development: accelerate the development of new dental devices through virtual simulations, reducing the need for physical prototypes and initial clinical trials.

So, the biggest advantages of FEM are:

- Precision: provide detailed and accurate analysis of dental structures and devices;
- Cost reduction: minimize the need for expensive physical prototypes and preliminary clinical tests;

- Customization: allow the customization of dental devices in according to the specific needs of patients;
- Safety: help identify and mitigate potential problems before clinical implementation.

3.4.2 ANSYS WORKBENCH

ANSYS WORKBENCH is an engineering simulation software. FEA is the principal tool used to analyze the strength, toughness, elasticity, temperature distribution, electromagnetism, fluid flow, deformations, and other features in the scenario studied; and ANSYS possesses all these tools.

A user may start by load an object and then adding weight, pressure, temperature, and other physical properties.

In this thesis it is used as the CAD and CAE software in order to perform the FEA analysis of the four-element crown bridge.

In the medical field, ANSYS with the use of FEM is particularly useful for studying complex biomechanical systems that are difficult to study in vivo. It can be used to predict the mechanical response of tissues under different stimulation conditions, both in healthy and pathological states, and to assess the structural changes.

Moreover, the FEA method has been also extensively used during medical device design, allowing for the investigation of function and possible complications of novel devices. [21]

3.4.3 FEA software for dentistry

The FEA is an upcoming and significant research tool for biomechanical analyzes in biological research. It is an ultimate method for modelling complex structures and analyzing their mechanical properties.

In implantology, FEA has been used to study the stress patterns in various implant components and in the peri-implant bone. It is also useful for studying the biomechanical properties of implants as well as for predicting the success of implants in clinical condition.

Ansys Workbench R1 2024 is a general-purpose high-end software, so it is more convenient for Yndetech to use a specific low-end software for the dental field. The FEM is basically a numerical method of analyzing stresses and deformations in the structures of any given geometry. The structure is discretized into the so called 'finite elements' connected through nodes. The type, arrangement and total number of elements impact the accuracy of the results.

The steps followed are generally constructing a finite element model, followed by specifying appropriate material properties, loading and boundary conditions so that the desired settings can be accurately simulated, as will be explained in the next chapters.

Various engineering software packages are available to model and simulate the structure of interest may be implants or jawbone. [22]

An alternative tool to Ansys is DentalFEM, it is produced by a Spanish company. DentalFEM employs Artificial Intelligence (AI) and the FEM to provide accurate biomechanical response of implant treatments.

It generates a comprehensive 3D digital replica of the patient's oral cavity, reconstructed from Computed Tomography (CT) imagery. FEM simulations provide the biomechanical response of the treatment (stress-deformation patterns on teeth, implants, and surrounding bone and tissue) and implant survival can be evaluated.

DentalFEM operates seamlessly as a user-friendly web service, accessible from any device, operating system, or web browser.

The importance of dedicated FEM software is by virtually testing surgical outcomes before proceeding with the actual procedure, clinicians gain confidence in their treatment plans, reducing the likelihood of unnecessary surgeries or implant failures. This not only saves time and costs but also enhances patient satisfaction and overall treatment outcomes.

Dental implants do much more than replace missing teeth, they help maintain and strengthen bone structure, provide the ability to chew food, and give patients the confidence to smile. Since dental implants are common tooth-replacement options, complications can occur. Indeed, traditional approaches often fall short in delivering optimal implant biomechanical outcomes potentially reducing implant's lifespan (mechanical failure is around 10%).

As said, FEM is a numerical technique to study mechanical response of a structure computationally, despite FEM has proved to be a useful tool in implantology, the scientific skills needed and the steeped learning curves of general-purpose software (expensive license fees such as Ansys), preclude the use of a dental-specific approach in daily clinical practise.

The goal of using this approach is to reduce failure rates at least of 40% by using computational methods, to visualize stress distributions, analyze deformation patterns, study implant-bone interactions and make well-informed decisions based on evidence.

In the United States alone, it is estimated that approximately 100.000 dental implants fail each year (assuming a 10% failure rate). These failures result in a substantial cost of around $ 400 million annually. By increasing the budget by just 6% and implementing the use of FEM software, the country has the potential to save approximately $ 160 million on failure treatments each year.

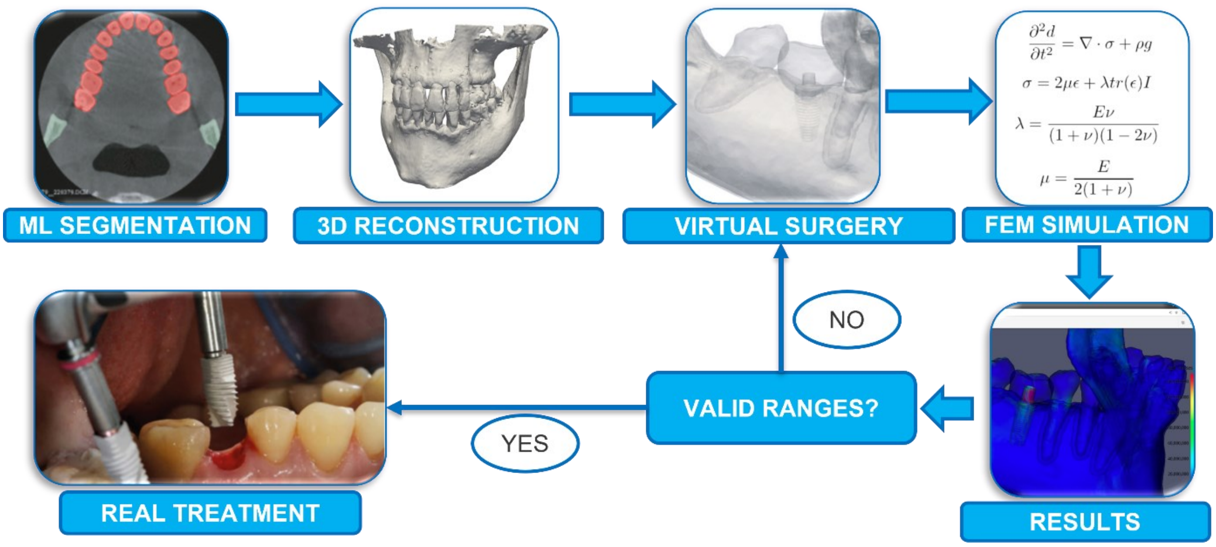


Figure 3.8 DentalFEM workflow

On Figure 3.8 it is possible to see the work chart of DentalFEM software, first it involves CT scan in order to obtain the regions of interest for the obtention of the patient's 3D digital replica. This process is called machine learning (ML) segmentation. Then, based on CT data an accurate 3D reconstruction of patient's oral cavity is obtained, containing all relevant structures for the finite element analysis. With this program is possible to reconstruct also periodontal ligament, and each region is assigned with variable material properties (i.e. density, Young's modulus and Poisson's ratio) to represent the biomechanics.

An interesting application of this software dedicated to dentistry is the 'Virtual surgery', the user can place the implant. In fact, it has a database with assemblies (i.e. screws, abutment, crowns etc.). The user in according with dentist can do several procedures how to position implant, define angles, implant materials, shape and dimension. This powerful tool allows to simulate also stages of the implant treatment (e.g. sinus augmentation or bone grafting) by modifying the virtual 3D geometry of the patient and placing the new implant, and so studying the outcomes for the patient from a mechanical perspective.

The next step is to obtain the structural response of the implanted protheses by solving solid mechanics equations by using FEA. The 3D model is meshed. This will provide biomechanical values at every point of the oral cavity geometry based on different loading conditions to understand the influence of the treatment, forecasting the outcomes before going to the chairside with the real patient.

Bone remodelling and osseointegration, together with non-linear material properties of tissue will be part of our final release, providing the most accurate oral biomechanical model developed so far.

Finally, the computer provides a wealth of data that includes stress and strain distributions among others. This will aid in patient specific surgery planning (e.g. material selection, diameter or length of implant etc.) and guarantee the values for this procedure are within the acceptable range to prevent implant failure. If results do not fall in valid ranges the algorithm returns to the virtual surgery and steps are repeated, otherwise it proceeds with the real treatment.

Therefore, this tool wants to obtain answer about the implant analyzed, considering some aspect as:

- The type of implant (i.e. material, shape, dimension etc.);
- The risk of failure for the treatment;
- The loading is immediately feasible to guarantee the reliability of the prosthesis based on bone quality;
- Stress valuation and if it is evenly distributed;
- Evaluation of the components;
- If stress could be an issue for osseointegration.

The main features of this software but in general for FEA tools are several as for example the process automation for all stages of FEM analysis. Logically, it is a non-invasive tool based on images and data obtained with CT.

For this software the outcomes can be analyzed interactively and immersively in real time with augmented reality.

Thus, DentalFEM is a useful tool in the field of dentistry, because it provides performance in conjunction with other devices (e.g. scanner or CT) that makes it complete for normal implantology and prosthetics activities.

The FEA tools provide a large amount of information which cannot be obtained with traditional methods in implantology and prosthetic such as the stress or deformation in the implant or periodontal ligaments.

Moreover, the user can reproduce the test as many implants as it wants under the same conditions, to decide the device with the best result. The result can be obtained very fast, and the user need only an internet connection and a PC. [23]

3.4.4 Development costs of dental CAE software

The development cost of a dental-specific CAE software depends on several factors. For example, detailed three-dimensional FEM simulation, integration with dental scanners, stress analysis on materials are features that increase costs, as well as artificial intelligence. In addition, the development team requires several professional figures who can elevate the cost. For the medical industry, rigorous testing and compliance certifications are critical and augment the expenditure. An approximation of the cost of production is based on different scenarios.

If the software must be completely new, if it only includes basic functionality for FEM as the static structural simulations (e.g. implementation of material, meshing, boundary conditions, stress and strain results), the cost could fluctuate between 70.000 € and 150.000 €.

Medium complexity software that includes more advanced FEM simulations such as 3D simulation, integration with diagnostic devices (e.g. CT, MRI, optical scanner), could cost between 100.000 € and 250.000 €.

Highly complex software with sophisticated FEM simulations, including all advanced features, all medical hardware integrations, stress-life analysis, the AI, and the augmented reality, the cost could exceed 300.000 €, arriving up to 500.000 €.

Instead, the integration of basic functionality on existing CAD, of relatively simple functionality, such as some basic structural analysis, the cost could be between 30.000 € and 70.000 €, significantly lower.

Furthermore, if Yndetech wants to integrate advanced features such as dynamic analysis and integration with specific hardware (i.e. scanners, 3D printers), the cost could vary between 70.000 € and 150.000 €, a more realistic perspective considering the company resources.

Instead, a highly complex integration or significant customization the cost could exceed 150.000 €, reaching up to 300.000 € or more. [24]

3.5 Current Vs New digital workflow

In this chapter are highlighted the principal differences among the current digital workflow used by the Yndetech company, and the new digital workflow proposed in this project. The new one intends to improve the production process and ensuring greater reliability of the dental prostheses implanted on the patient.

In Figure 3.9 the flowchart of the current digital workflow previously introduced. It would be, first, with the intra-extra oral scanning process, then CAD software to optimize and finish the design, the CAM software to manage the manufacturing process depending on the type of procedure used. If milled Hyperdent is used, instead Oqton for SLM.

As said, Rhinoceros 3D is transversal to Hypsocad for cases in which it is necessary to decrypt the geometries of the prosthetic connections.

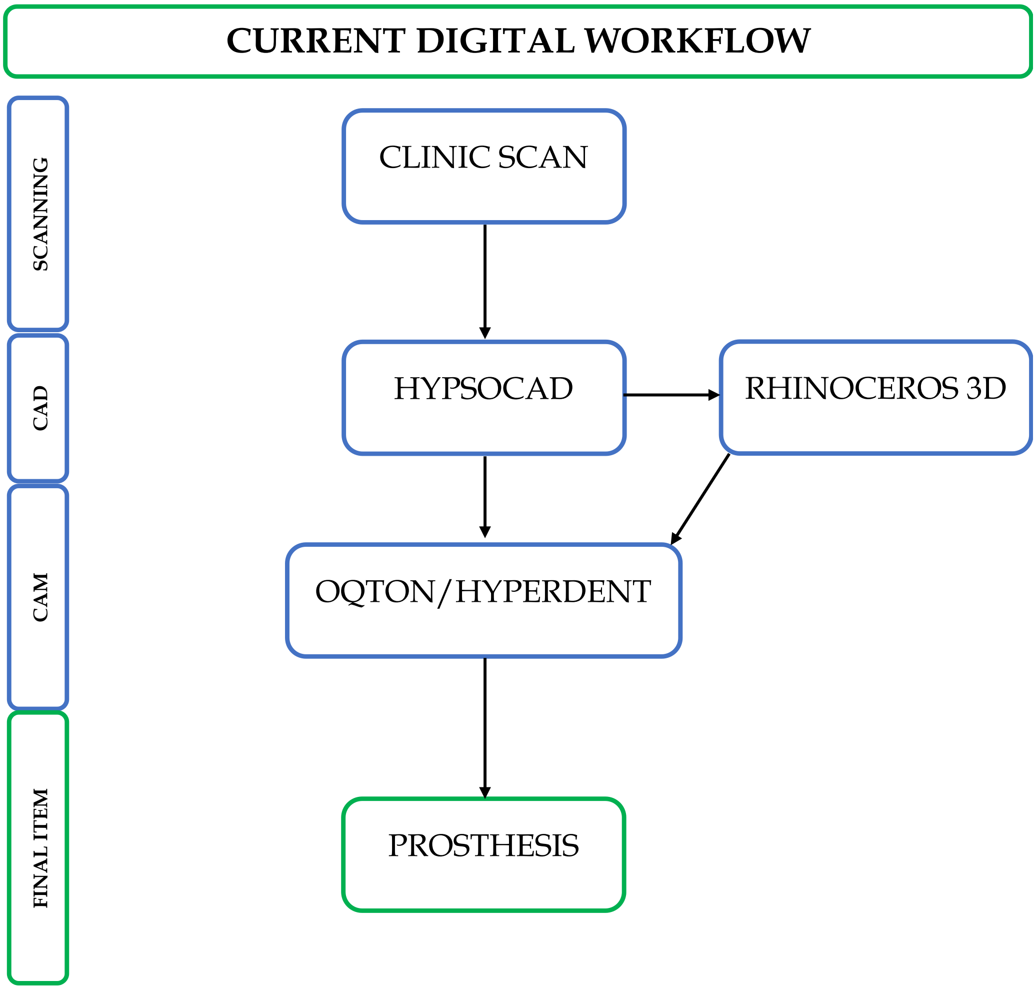


Figure 3.9 Current digital workflow

Instead, in Figure 3.10 it is possible to see the addition of the CAE block. The CAE software used is ANSYS WORKBENCH 2024 R1 in order to study the structural phenomena previously descripted.

Hence, with the definition of the material and boundary conditions, this tool can be used to simulate the behaviour of the prostheses produced. CAE software can be very useful to see any critical points.

Once, the simulation is done the engineers can correct either the geometry (on a CAD software) or the material or the setting process to make the dentures as reliable as possible in the patient's mouth and avoiding ruptures during production and the wearing.

If the prostheses withstand from a structural point of view the devices can be produced and so implemented on CAM system and finally obtaining the prosthesis.

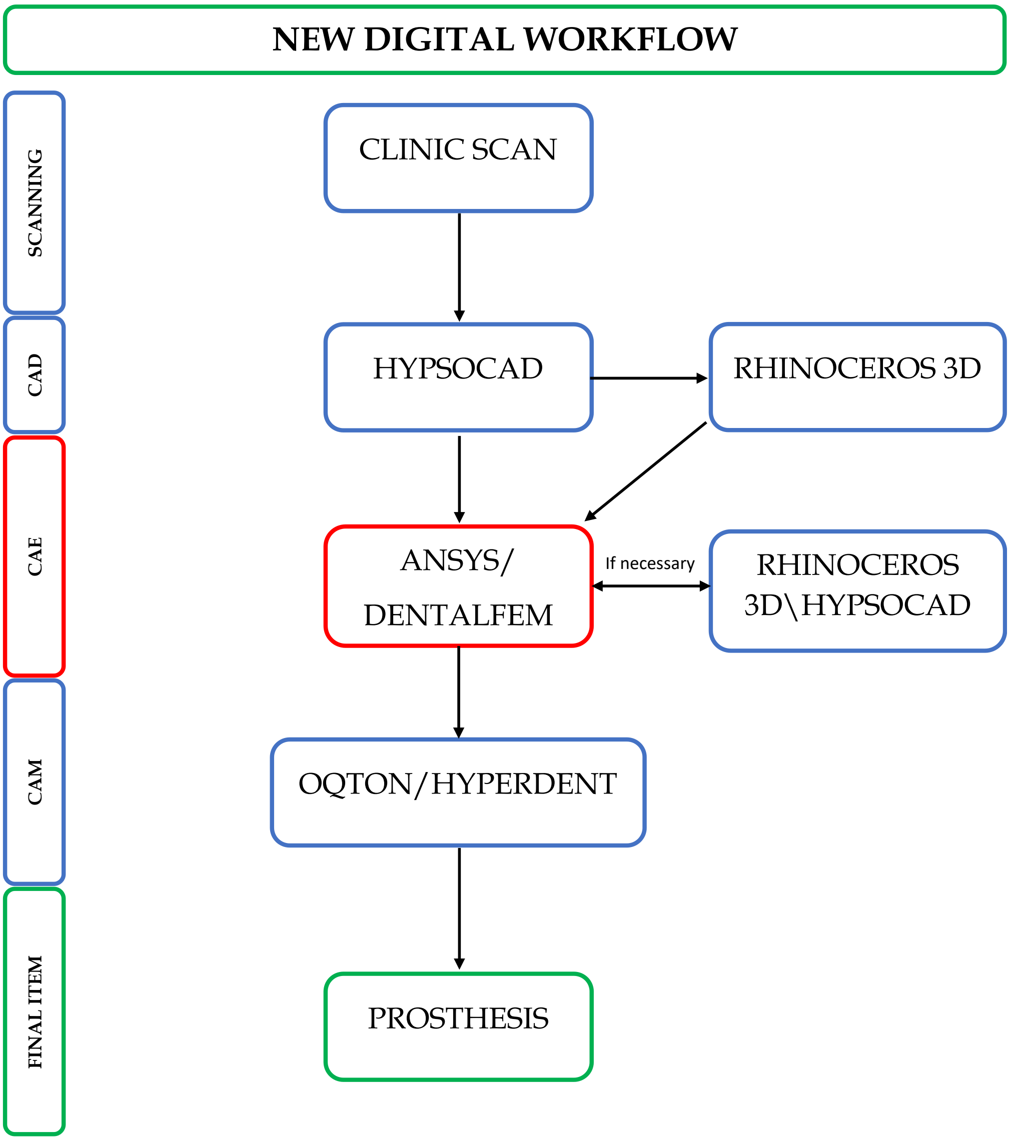


Figure 3.10 New digital workflow

4 Case of study

This section shows the dental prosthesis examined in this thesis with appropriate modifications to improve the structural performance.

It is important to introduce the concept of iteration between bone and implant. In recent years, dental implant prostheses (i.e. understood in its complexity or the whole prosthesis) have been used as surgical components in a wide variety of oral restoration scenarios, including single-tooth replacements or multiple implant-supported bridgeworks replacing various missing teeth. The success of the treatment depends on several biological and mechanical factors. Even when implants are fully osseous integrated, there is a risk of bone resorption in the peri-implant regions, which can be activated by bacterial infection or overloading due to masticatory forces. [25]

To date, the FEA analysis of the bone-dental implant system has been investigated by analyzing several types of prostheses; modelling only a portion of bone considered as isotropic material, despite its anisotropic behaviour; assuming in most cases complete osseointegration; considering compressive or oblique forces acting on the prosthesis; neglecting muscle forces and the bone remodelling process. Moreover, a large amount of research has been carried out on the application of the FEA method in dentistry. Most of the research has been focused on examining the implant mechanical behaviour during functional loads with the aim to assess the stress and strain fields in prosthesis components and bone at a macroscale level.

However, mechanical loading induces adaptive changes at a microscale level affecting bone homeostasis and remodelling. These changes significantly impact the shape of the stress and strain fields at the bone-implant interface.

Capturing these effects through numerical modelling may open a new wave of multidisciplinary approaches aiming to understand and control bio-integration and improve the long-term survival of dental prostheses. [21]

This project focuses solely on analyzing the mechanical physical phenomena in the dental prosthesis and not among implant and bone. The research part of this thesis regards the interpretation of the FEA simulations of a prosthesis most produced in the world of dental prosthetics, the crowns bridge.

In particular, the crowns bridge in this study is composed of four elements, three of which composed with a channel with T-Base inside, in surgical procedure it will be screwed with the endosseous screws, to have artificial teeth on patient’s gum.

In general, the crown bridge involves a connection either with abutment or T-Base with the implant or it is cemented on the natural stump.

It is very important to evaluate the fracture toughness (K_{Ic}) and flexural strength of material. Fracture toughness is expressed in units of stress times the square root of crack length (MPa$\sqrt{mm}$). [26]

Fracture toughness is a material property that describes its ability to resist the propagation of cracks. It is particularly important in engineering applications where materials are subjected to mechanical loading or stress cycles. Essentially, it measures how much energy a material can absorb before it fractures. Higher fracture toughness indicates greater resistance to crack propagation, making a material more reliable and safer under stress.

Instead, the flexural strength, also known as modulus of rupture, refers to the maximum stress a material can withstand before it breaks or fractures under bending forces. In the next sections are better explain these concepts with the FEA simulation.

4.1 Four-element crown bridge on T-Base (failed prosthesis)

The case of study is a four-element crown bridge made in zirconium dioxide (ZrO_2) on Figure 4.1 (a), which was broken over time.

It is a prosthesis of the partial upper right arch, so the teeth number are 13, 14, 15, and 16 which correspond to canine, first and second pre-molars and first molar. The piece has milled in a ZrO_2 disk. In particular, the ZrO_2 disk which was used is of 'Aidite Technology Co.' company and is the model named 'Superfect Zir SHT Preshaded', on Table 4.1 the technical specifications of the aforementioned ZrO_2 disk.

Table 4.1 Superfect Zir SHT Preshaded ZrO_2 disk technical specifications

Technical specifications	Parameters
Color	VITA 16 colors
Aesthetic	Super high translucency
Flexural strength	1000 MPa
Coefficient of thermal expansion	(10.5±0.5)*10 K
Chemical solubility	≤100µm.cm-²
Fracture toughness (K_{Ic})	≥158,11 MPa$\sqrt{mm}$
Sintered density	≥6.0g/cm³

Teeth number 13, 15, and 16 are on T-Base, they rest passively on the endosseous screws, and these latter are provided with the external hexagon geometry. Instead, the 14 leans on the patient natural gum.

ZrO_2 is a ceramic material ideal for dentistry because can be coloured to match teeth and it is translucent, mimicking the natural look of teeth far better than, for example, any crown with a gold base. The ZrO_2 works extremely well with glass ionomer cement. [27]

Before the prosthesis reaches the patient's mouth, the dental technician pours some porcelain layers, and the bonding process is very strong. It means that the porcelain is less likely to fracture away from a zirconia base with respect a prosthesis totally composed in porcelain.

This kind of prosthesis is optimal for a full crown restoration where the cosmetic is the principal purpose and extra strength, and durability are required. [28]

The biggest problem with crowns bridge is due to manufacture process, so rupture because some parts could be too thin to process. Moreover, the fatigue over time, cause of chewing cycles, which could be dramatic, especially on prostheses where the thickness is too thin but resistant enough to pass the processing stage.

Especially, the connector elements are subjected to the highest stress and Yndetech controls the thickness of every project, and if the prosthesis is below dimensional standards, in some parts, a document named 'release of responsibility' is signed to the customer requesting production. In this case of study, the connector item is the artificial tooth number 14, and the thickness of the prosthesis is controlled in every part showing an abnormality as will be seen below.

A FEA on this kind of prostheses can be very useful to study the stress and deformation phenomena over time in patient's mouth and understanding the degree of reliability and give a "shelf life" to the product. Also, FEA is involved, as seen in the Figure 3.10, previous the manufacturing step in the digital workflow, in order to assist and avoid breakage during processing and to study the stress and strain data before entering the patient's mouth.

So, this prosthesis is composed by crowns made of ZrO_2, T-Base made of titanium grade IV, and the composite resin to fill the space between T-Base and crowns. Subsequently, the prosthesis will be screwed with a fixation screw to the implants (i.e. the endosseous screw), and the holes covered with a suitable filler called "composite".

To be precise it is also calculated the total dimension of the prosthesis, that is 27,44 x 24,12 x 18,93 mm as seen on Figure 4.1 (b).

Another aspect possibly to be considered is the load transmission at the bone–implant interface (not concern this project), which is influenced by a variety of factors, including the length, diameter, form, and surface of the implants and prosthesis. Then the material properties of the implant and prosthesis, as well as the geometry, quality, and quantity of the residual alveolar bone. [29]

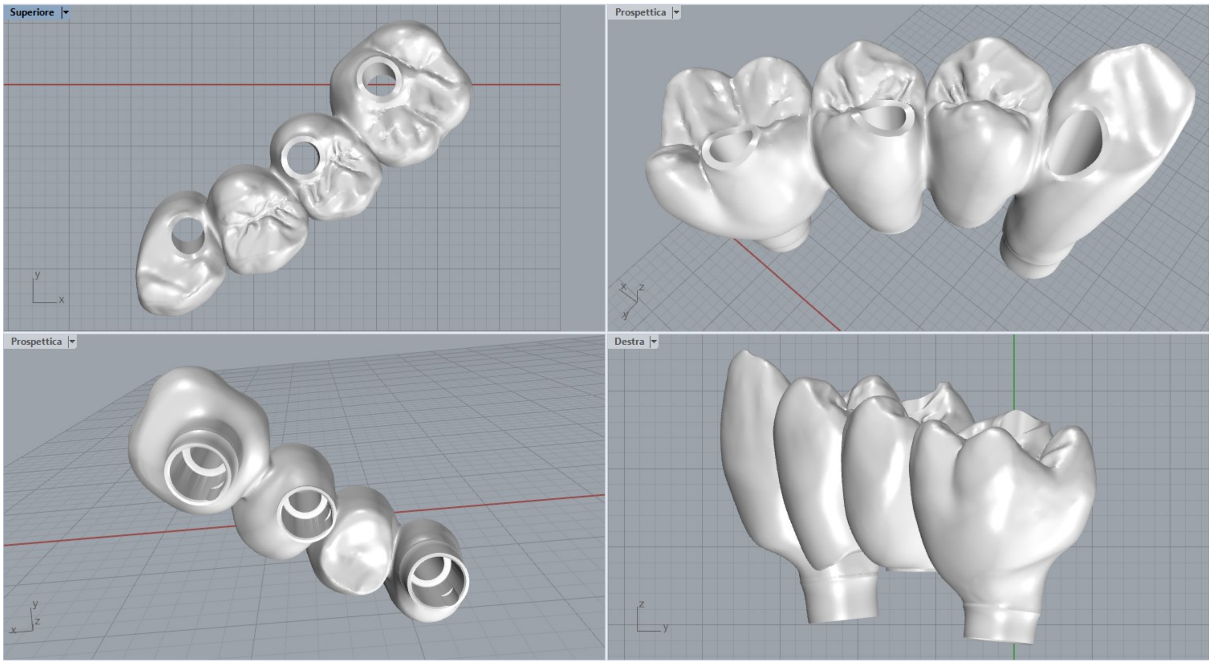


Figure 4.1 (a) Four-element crown bridge on T-base

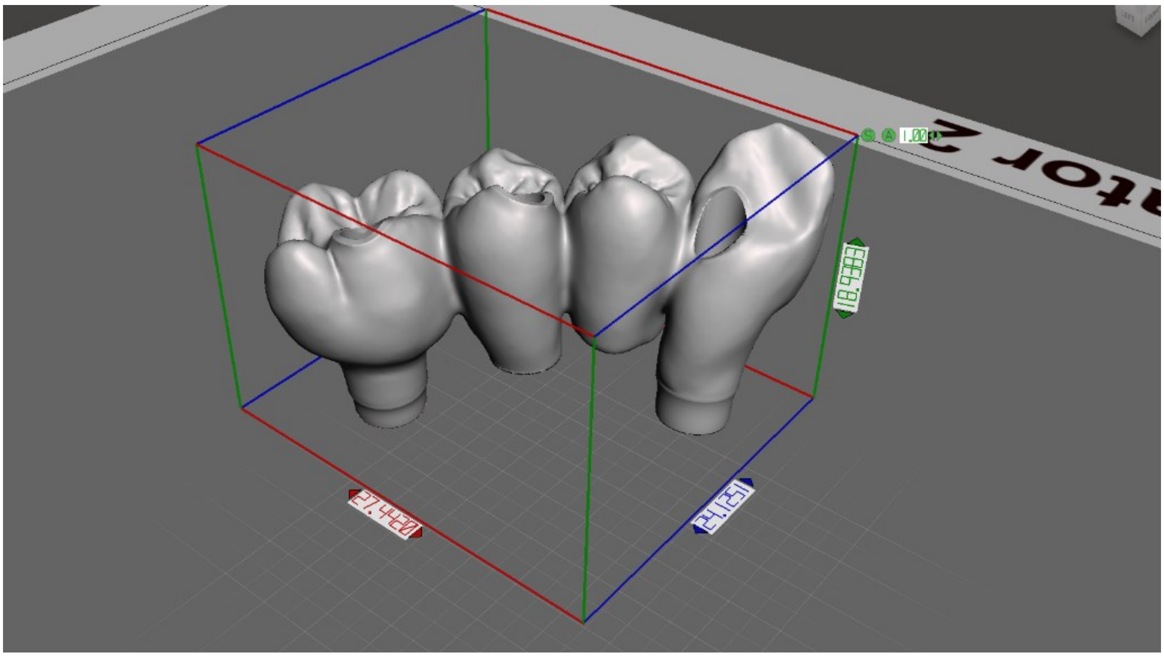


Figure 4.1 (b) Total dimension of prosthesis

4.2 Rupture analysis

Following medical advice from professional dentists, the principal problem of the prosthesis rupture, from a hypothetical point of view, regards the cementation. In Figure 4.2 (a) it is possible to see the photo of rupture of the prosthesis, it is also appreciated the breakages occurred on the right side of the molar and canine and a circular break extended upwards in the second pre-molar channel. Logically, it is very difficult to understand a posteriori the type of rupture and the possible start of the crack.

It seems the typical fatigue rupture, that is a type of failure that occurs after a large number of repeated or variable load cycles, even if the loads are less than the breaking load of the material under static load. It starts with the triggering of micro-crack on the surface of the material, which propagate as the load cycles progress. The cracks propagate stably to a critical size, at which point a sudden and catastrophic break occurs.

Fatigue rupture occurs with little or no visible plastic deformation, and has smooth surfaces with streaks. [30]

Figure 4.2 (a) Broken prosthesis

Also, an approximation of the break area with Rhinoceros 3D is possible to see on Figure 4.2 (b), it has been calculated, which are the following:

- Molar's rupture area = 23,63 mm^2;
- Pre-molar's rupture area = 25,00 mm^2;
- Canine's rupture area = 27,07 mm^2.

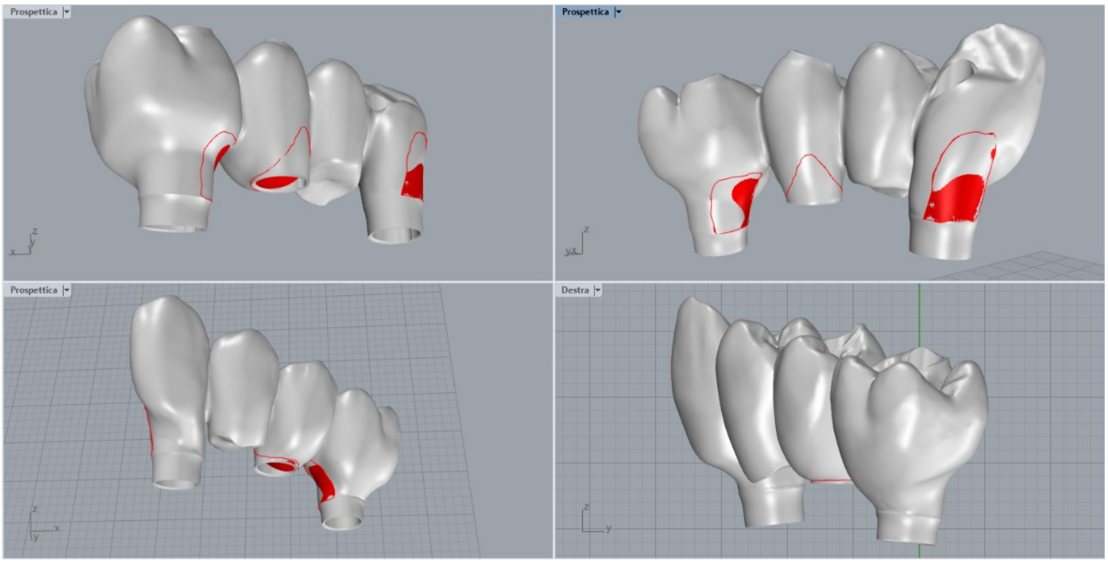


Figure 4.2 (b) Rupture areas

The crowns and T-Bases were subject to a mechanical cementation rather than adhesive because the filler among crowns and T-Base was a resin and not a cement, as for example the glass ionomer cement which would paste the T-Base to the prosthesis by transforming it a monoblock.

This causes a relative movement between the elements that becomes dangerous for the stability of the system over time. It imagines that the first rupture has happened into the resin and in this manner a gap between the crown and T-Base occured.

A relative movement was established between the T-Base and the crown and there was no longer the resin that acted as an absorbent material and as a result of greater efforts the crown broke, it needs to be considered the titanium is harder than zirconium dioxide material.

In this manner, there were three elements in motion, and the risk of breakage became greater than a prosthesis that would be a monoblock with the T-Base.

Another issue concerns the thickness of the lower walls of the prosthesis where the connection with the T-Base and the resin takes place. A thickness analysis is done on Rhinoceros 3D and it is shown in Figure 4.3, in particular the minimum values of the thickness at the base of the prosthesis is on average of 0,369 mm and the smallest part with a recess of 0,175 mm.

It is possible to see a thickness of 1 mm where there are links with the other elements. It is likely that as soon as the resin breaks, a "play" is created between the T-Base and the prosthesis, then it is very likely that the efforts generated are excessive for the integrity of the device.

In addition, another comment made with dental clinic professionals is that the attachments to the implant, so to the endosseous screw, should be at the same height for all three T-Base; thus, they must be levelled. In this implant there are external hexagonal geometries, which means a maximum disparallelism angle of 35 degrees, in order to have a passive insertion of the prosthesis into implants.

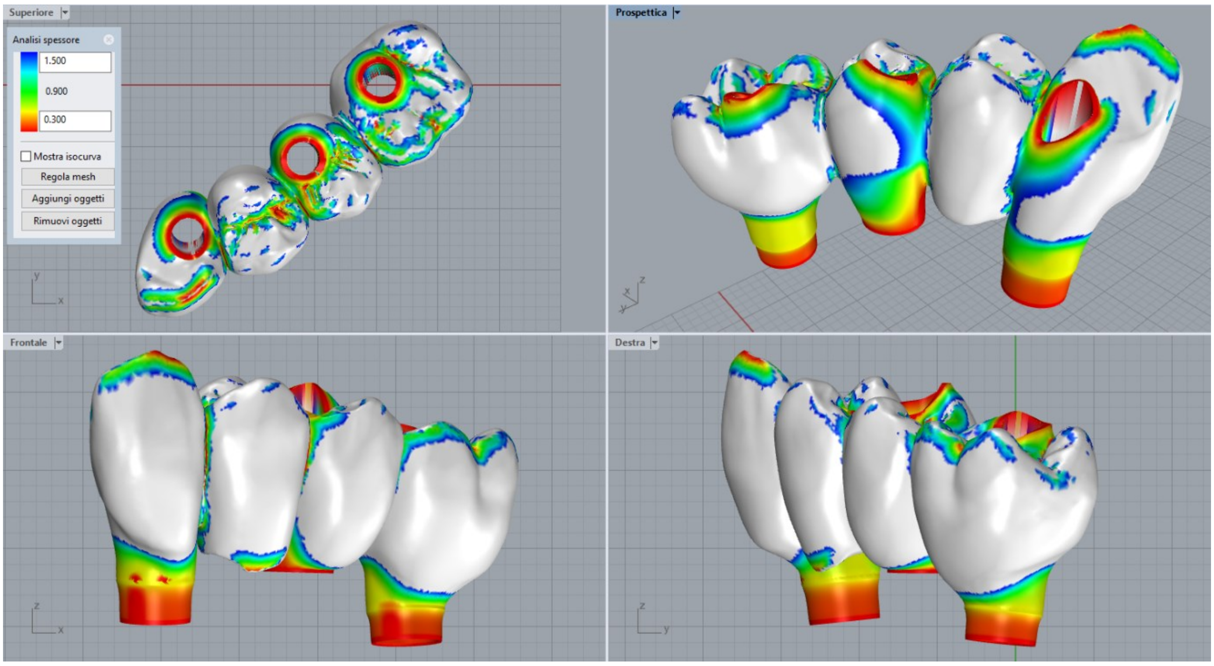


Figure 4.3 Thickness analysis of failed prosthesis on Rhinoceros 3D

For this purpose, an axis tilt analysis (Figure 4.4) has been performed on Rhinoceros 3D to prevent angles smaller than 35 degrees, because in that case insertion would not have been passive and further efforts would have been generated at the prosthesis base.

Taking as a reference the axis of the T-Base of the molar, the angles between the molar and the other T-Bases are calculated. Thus, 6,84 degrees is between molar and second pre-molar and 12,27 degrees between the molar and the canine.

Instead, the angles between the implants and the maxillary plane is 90,40 degrees for the molar, 91,68 degrees for pre-molars and 89,04 degrees for the canine. This last analysis only to have a complety view of the inclinations.

Hence, it can safely declare that the disparalleism is not the problem of the rupture and no additional effort complicates the structural situation of the prosthesis due to the inclination axes. Moreover, it is also important to assess, in according with purpose of this project, the magnitude of stress and strain values in the rupture areas.

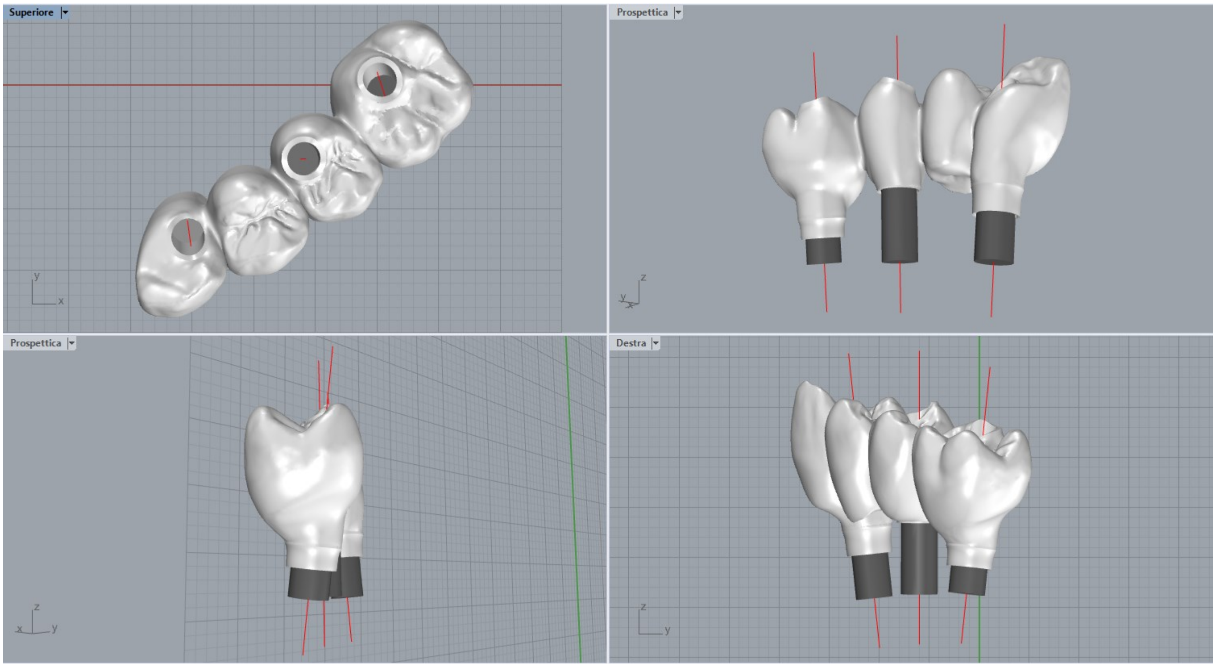


Figure 4.4 Axis tilt analysis on Rhinoceros 3D

Ultimately, the hypotetical cause of the break up is the association of three problems (from a clinical point of view):

- Mechanical and non-adhesive cementation (certainly the main problem);
- The thickness too thin of the cylinders at the base of the prosthesis (0,310 mm at the lower end);
- The prosthetic channels too long and not on the same plane.

Furthermore, several possible solutions to the problem, counteracting the break, can be applied. Firstly, from a clinical point of view the use of the adhesive cementation material between prostheses and T-Base so as to form a monoblock. Alternatively, level the channels of the prosthesis where the T-Base will be installed, so that they are all at the same height and eventually add intermediate abutment between the endosseous screw and T-Base. Logically, for the only cut prosthesis it is necessary to make additional processing in the implants. Whereas for the installation of abutment is necessary to follow the internal geometry of the prosthesis.

Another opportunity to improve the prosthesis, less sophisticated is to increase the thickness where it is possible, in the channels where there will be the T-Base and cement so as to promote the durability of the device.

Moreover, guarantee the passivity of the prosthesis insertion, if the geometry is external hexagonal, the degree among axis must be below 35 degrees. If the geometry is internal hexagonal the degree must be below 12 degree, it is important to verify the axes inclination.

Another perspective, very important, to consider is the time; indeed, the prosthesis was requested by the customer on 8\11\2018 and the date of breakage is 30\01\2020 so by considering 10 days to produce and deliver the medical device, the prosthesis has lasted 1 year and 73 days.

Therefore, it is useful to understand how many chewing cycles the prosthesis has undergone, taking into account an average of 30 chewing per bite of food, as mentioned in the previous chapters.

Hence, if the patient has on average 10 pieces of food in a meal, they become 300 chewing cycles per meal. A normal person has 3 meals per day, so that are 900 chewing acts. Ultimately, for 1 year and 73 days that are 438 days the patient has performed 394.200 chewing cycles. A prosthesis of this type should have lasted at least 6 years and therefore 1.971.000 mastication. The case study prosthesis is far below the expectations of duration.

The rupture analysis will evolve as the thesis proceeds based on stress and strain data obtained from simulations.

From an enginnering poitn of view, in addition to all that has already been mentioned, to consolidate the various hypotheses, it needs to perform the static structural analysis of the prosthesis through the FEA, by studying stresses and deformations in the areas of rupture. The breakdown of the three T-Base channels should be found in manufacturing defects that would lead to the formation of micro-cracks. The so-called fatigue rupture is a type of failure that occurs after a large number of repeated or variable load cycles, even if the loads are less than the breaking load of the material under static load. The fracture mechanics in this case provides the crack slowly begin to propagate as the load cycles progress in a stable way, until it reaches a critical size. At that point the critical propagation of the crack can lead to a sudden and catastrophic break of the component. Hence, it is crucial to be aware of stress and deformation data, as shown in the next sections.

It can conclude this paragraph by saying that the hypothetical statements referred at the beginning examine the problem from a clinical point of view, while these last declarations, about the fracture mechanics, analyze the issue of the break from an engineering point of view based mainly on stress and deformation data, illustrated in the next paragraphs.

4.3 Pre-processing simulations workflow

The pre-processing process is fundamental to obtain a good result and doing a proper interpretation of physics phenomena.

As said, the static structural analysis was performed with ANSYS WORKBENCH R1 2024.

First, it was selected the toolbox called 'Static Structural', to make the structural analysis and simulations of the medical device.

The job was started with configuring the 'Engineering Data' section, it was opened, and a new material was created, it was the 'Zirconium Dioxide', and isotropic elasticity was implemented, where the Young's Modulus of 205.000 MPa and the Poisson's Ratio of 0,25 was defined. [31]

Furthermore, the geometry was loaded (as .Stl file), and it was corrected on Ansys 'Space Claim', the extension CAD software, with the command 'Check Facet' and consequently 'Auto Fix', in order to have a useful file for simulations.

Then, the project was ready to be processed on 'Mechanical' window and work effectively on simulations. It is very important put the correct unit of measure in this case in "mm, Kg, N, MPa, etc.". The material implemented at the beginning must be associated with the geometry parts, and controlling the stiffness behaviour was flexible.

The next step was the meshing of the geometry, by considering the 'Patch Independent' method, by starting from the outer geometry to inner with tetrahedrons, the minimum size limit is of 2 mm and with a medium span angle centre. The element quality has obtained an average of 0,76 with a standard deviation of 0,13.

At this point, it is necessary to select the areas related to boundary conditions. So, in this case the upper areas of the prosthesis where pressures will be established with the antagonist part of the dental arch.

The study was performed in two ways with different pressure configurations, one by considering vertical forces (i.e. perpendicular to occlusal plane) and the second with an outwards inclined force of 45°.

An additional boundary condition was placed at the base of the cylinders where the T-Base will be installed and where it will rest on the endosseous screw (in this case an external hexagon geometry), only in the circular area below the channel.

For each selected area a reference system was created to position the pressures and boundary conditions more precisely (Figure 4.5).

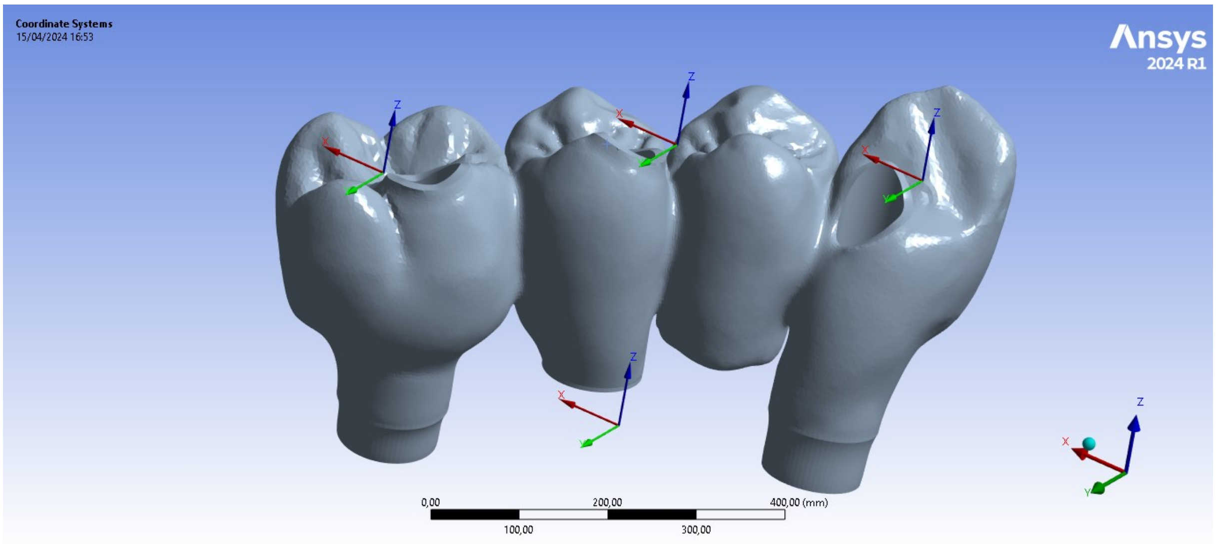


Figure 4.5 Coordinate system of selected areas

Then, it had to define the pressure values on each area of the teeth. First, it needs to know the areas, this task was done on Autodesk Netfabb 2024. So, an approximation of the occlusal tooth area (Figure 4.6) was obtained, and they are the following:

- Molar's Area = 98,15 mm^2;
- Pre-molars's Area = 109,78 mm^2;
- Canine's Area = 58,40 mm^2.

It was possible to calculate the pressure acting on each tooth with the forces perpendicularly to the area, so regarding only a vertical force. The pressure was defined as the force over area (P [MPa] = $\frac{F\,[N]}{A\,[\mathrm{mm2}]}$) and considering the force values which is 150 N for the molar, for both pre-molars are 85 N and for the canine is 60 N, for healthy chewing, in according with the Table 3.1.

The following results were obtained for the vertical forces:

- Molar's Pressure = $\frac{150\,N}{98{,}15\,\mathrm{mm2}}$ = 1,52 MPa;

- Pre-molar's Pressure = $\frac{85\ N}{109{,}78\ \text{mm2}}$ = 0,77 MPa;
- Canine's Pressure = $\frac{60\ N}{58{,}40\ \text{mm2}}$ = 1,02 MPa.

When it was considered the inclined forces of 45 degree, the pressure formula changes (P [MPa] = $\frac{F\ [N]\ cos45°}{A\ [\text{mm2}]}$), because pressure is a quantity that measures the action of the force acting perpendicularly on a surface; thus in this case the force has two component, the perpendicular one and the parallel one to the surface. It must be considered only the perpendicular one (i.e. by computing the Cosine of 45°), because the parallel one "slips" on the surface. Hence, the following results were obtained:

- Molar's pressure = 1,08 MPa;
- Pre-molar's pressure = 0,54 MPa;
- Canine's pressure = 0,72 MPa.

The values are slightly lower than the vertical forces.

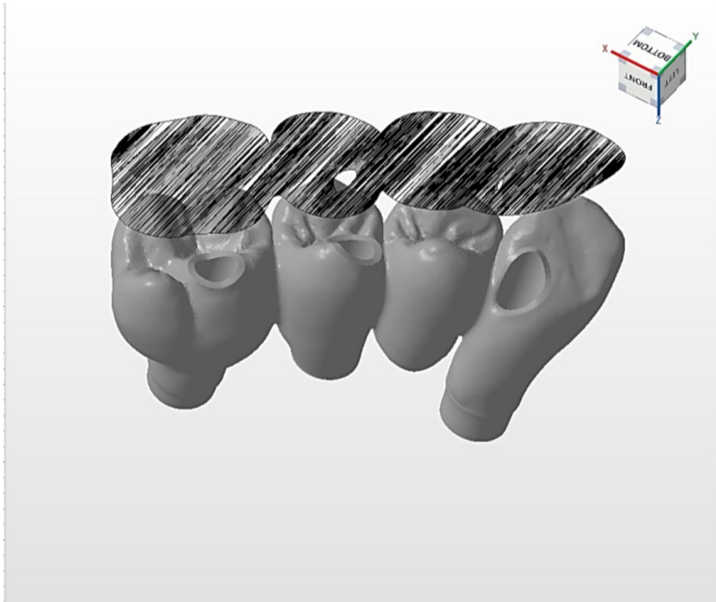


Figure 4.6 Approximation of each tooth area with Autodesk Netfabb 2024

Furthermore, the above-mentioned pressure values were implemented for specific areas, and as it can see on Figure 4.7 (a) the red arrows represent the pressures in vertical forces configurations which acts in the purple areas which were previously calculated. In the bottom, blue coloured represents the three fixed supports of the prosthesis.

The Figure 4.7 (b) represents the pressures on purple selected areas with the red arrows that represent the outwards tilted forces of 45°.
As already said, the pressure is always perpendicular to the surface so, however for a better understanding the inclined forces are represented.

Always, the blue areas are the fixed support of the device. This step is denoted 'Static Structural' on Ansys, and it defines the boundary conditions, and the configuration of the physical quantities present. The simulations were carried out under two conditions, first considering a normal chewing, and secondly using the values of the MVBF, both values can be taken from Table 3.1. [32]

The pressures with vertical forces implemented on simulation by considering the MVFB were the following:

- Molar's pressure = 5,62 MPa;
- Pre-molars's pressure = 4,01 MPa;
- Canine's pressure = 3,87 MPa.

Instead, the pressure with inclined forces of 45° implemented on simulation by considering the MVFB were the following:

- Molar's pressure = 3,97 MPa;
- Pre-molar's pressure = 2,89 MPa;
- Canine's pressure = 2,73 MPa.

This was done to assess the under-force behaviour of the prosthesis under normal and extreme conditions. Also, because many subjects may be affected by a pathology called bruxism or involuntary tightening of the jaws, resulting in teeth grinding and rubbing, between them, of the upper and lower dental arches causing a pressure of the occlusal plane certainly much greater than normal chewing. [33]

Which leads to a greater stress and wear of the prosthesis, an investigation also in this aspect must be done to appreciate the differences and the maximum effort suffered by the prosthesis.

A resume of all pressure values is in Table 4.2 depending on the different pressure values of normal chewing or MVBF.

Table 4.2 Pressure values used for vertical forces and inclined forces in each configuration

Simulation	Molar (MPa)	Pre-molars (MPa)	Canine (MPa)
Case of study vertical force	1,52	0,77	1,02
Case of study inclined force of 45°	1,08	0,54	0,72
Case of study MVBF vertical force	5,62	4,01	3,87
Case of study MVBF inclined force of 45°	3,97	2,89	2,73

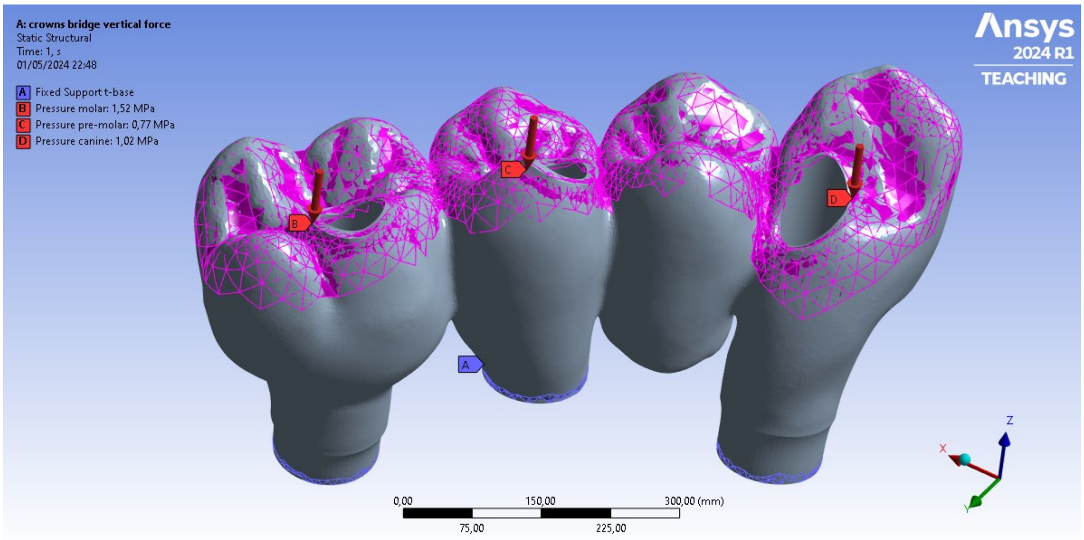


Figure 4.7 (a) Boundary conditions of the pressures with vertical forces

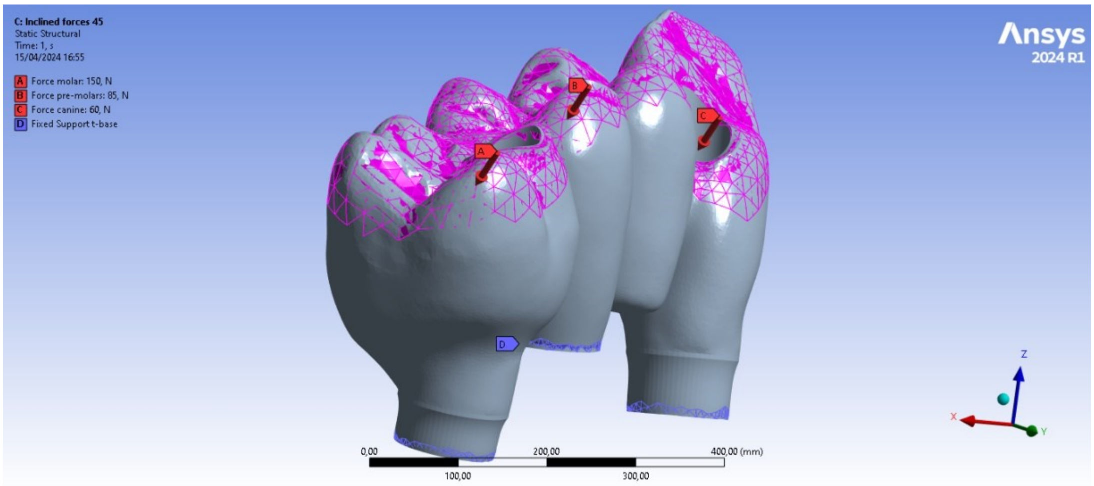


Figure 4.7 (b) Boundary conditions of the pressures with outwards 45° tilted forces

The last step is the implementation of the ‘Solutions’ commands to obtain the finite element simulations. The fundamental solution to study is the ‘Equivalent stress (von Mises)’ and ‘Total Deformation’.

According to the von Mises yield criterion, a given material will not begin to fail if the maximum value of the distortion energy does not exceed the distortion energy to require yielding the material in a tensile test. Equivalent stress of Von Mises is a value that is used to determine if a given material will begin to yield, where a given material will not yield if the maximum von Mises stress value does not exceed the yield strength of the material.

From a practical perspective, von Mises stress simply enables engineers to understand the performance of a complexly loaded part with a simple uniaxial or multiaxial tensile test.

It is important to underline this concept that will later serve, von Mises equivalent stress is a measure used to assess the state of stress in a material under complex three-dimensional load. It’s used to determine if a material subjected to a combination of multiaxial stress will start to fail.

Von Mises equivalent stress is useful because it reduces a complex three-dimensional stress state to a single scalar value, which can be directly compared with the yield stress of the material, in this case it is used for the flexural strength or the fracture toughness of the zirconia disk.

The von Mises criterion establishes that the material will yield (i.e. it will begin to deform plastically) when the equivalent stress of Von Mises reaches or exceeds the yield stress of the material, that is, there would be a ductile rupture, but the engineers should take into account any possible defects on the object.

It is widely used in structural analysis and engineering design because it provides a simple and effective measure to evaluate the strength and safety of materials under complex loads, it is a concept used in material mechanics to determine whether a material subject to a set of complex stresses will enter the yield. [34]

Since this thesis want to propose a solution to the problem of breakage, various geometric and dimensional changes through CAD software, such as Hypsocad or Rhinoceros 3D, had adopted; after analysing the case of break-up already discussed in chapter 4.2. Hence, re-propose the simulation to the finite elements in these modified prostheses by examining and comparing data with the same conditions with the case of study.

Therefore, different cases have been considered, the first involves increasing the thickness of the cylinders connecting the crown to the endosseous screw, as well as the housings for the T-Base.

The second concerns a geometric change, that is to cut the prosthesis and then level the housing of the T-Base all at the same height and evaluating in another simulation the insertion of an intermediate CoCr (by defining this other material on Engineering data) abutment placed between the cut prosthesis and the implant.

In this case the fixed support must be put in two points of the abutments; the first where will go the contact between the abutment and the endosseous screw (i.e. as for the others), and the second one in the internal beat of the prosthesis.

The only difference, in terms of pre-processing considering the abutment solution, is to define the three contacts between the abutments and the prosthesis (Figure 4.8). The dental prosthesis was defined as 'Contact bodies' (i.e. the deformable body) and the abutments as 'Target bodies' (i.e. the rigid body).

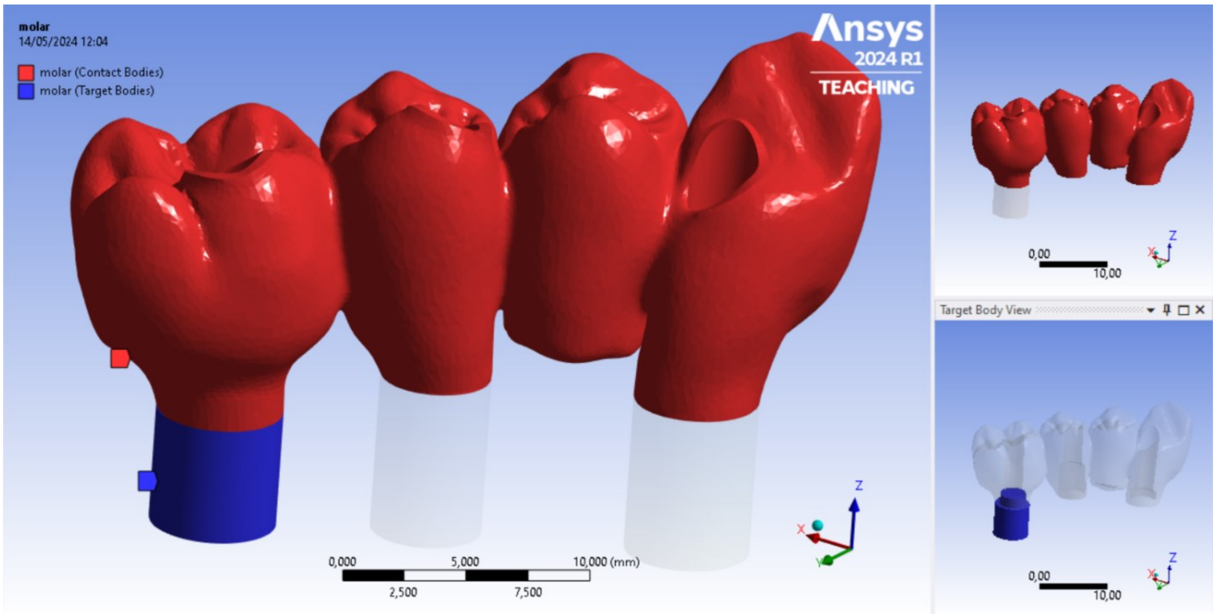


Figure 4.8 Contacts between prosthesis and abutment

Moreover, a fourth simulation that does not directly concern the case of study but can be useful to better understanding the interpretation of stress and deformation values obtained in the previous simulations, and if they the reliability of the method. So, the validation prosthesis is a transversal project to test the FEM, and an extreme case was developed.

Indeed, a crown bridge on the natural stump of the canine (i.e. number 13) and the lateral incisor (i.e. 12) has been modified "ad hoc", in such way some areas of the prosthesis were undersized and therefore would be unreliable from a structural point of view, and they probably would not have resisted the masticatory load. On Table 4.3 the pressure values of the validation method taken in the same manner as the case of study.

Table 4.3 Pressure values used for the validation method

Simulation	Canine (MPa)	Lateral incisive (MPa)
Validation method vertical force	2,24	3,38
Validation method inclined force of 45°	1,58	2,40
Validation method vertical force (MVBF)	5,06	5,24
Validation method inclined force of 45° (MVBF)	3,58	3,71

4.4 Thickened four-element crown bridge on T-Base

During the dimensional analysis, it was observed that the thickness of the cylinders that act as housing for T-Base were undersized (i.e. on average 0,369 mm) and therefore do not comply with the minimum technical specifications to cover this function.

The Hypsocad CAD software had increased the thickness to an average of 1,150 mm, because 0,600 mm is the gold standard for parts in contact with the T-base in a cemented prosthesis. As it is possible to see in Figure 4.9, where a further dimensional analysis was performed on Rhinoceros 3D, the middle part of the cylinders has an increased thickness with respect Figure 4.3.

It is logical to think that increasing the size of an object also increases physical performance. Furthermore, the concept was to obtain an object that reflects the gold standard for the thickness in which a break would take place very difficult, it also needs to weigh the thickness size so as not to change the adjacent gums and teeth too much.

Hence, it was done in order to obtain a further study in which it is possible to compare stress and deformations in the critical points in which the breakage had been happened, so that to be able to value the gap between the two models. To ensure that the data are as comparable as possible, the simulation was done in the same way as the case of study (failed prosthesis).

The simulations were performed both considering pressures with vertical forces and an inclined ones of 45° outward, in healthy chewing and MVBF.

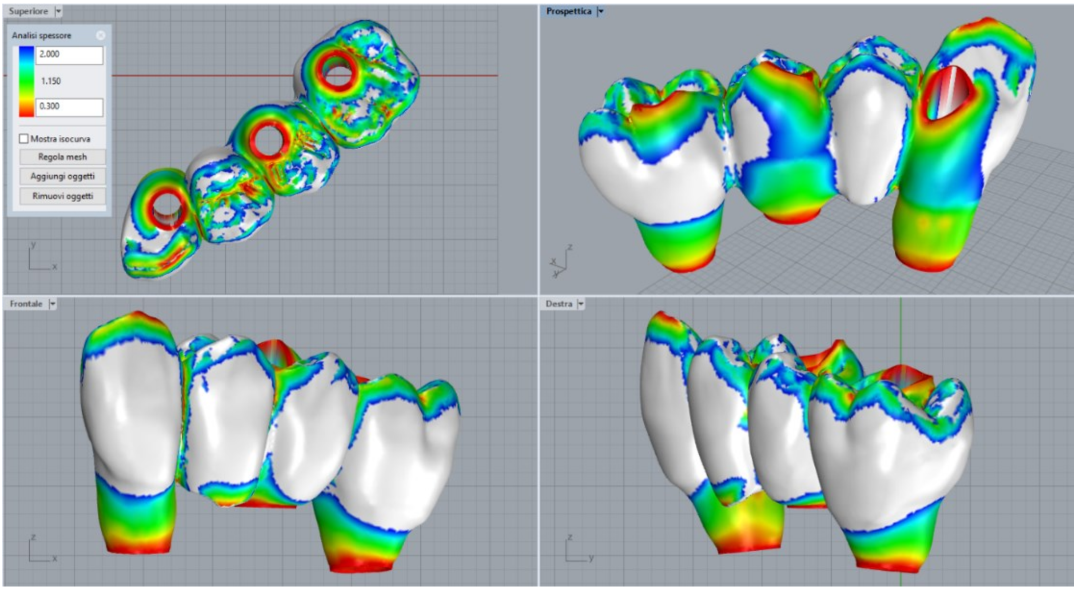


Figure 4.9 Thickness analysis of thickened prosthesis on Rhinoceros 3D

Then, the FEA analysis was performed by implementing 'Total deformation' and 'Equivalent Stress (Von Mises)'.

In Chapter 5 it is possible to see, respectively the Equivalent Stress of von Mises and the Total deformation data, with the graphic annotations which reflect the values of the physical phenomenon studied at specific points (i.e. in nodes of tetrahedrons) both by considering the perpendicular and inclined forces acting on prosthesis.

It is obvious that the deformation as well as the stress are less than the failed prosthesis, for the purposes of the thesis understand how these values are differentiated.

For completeness are also represented tables, where there are the values considering the outward inclined forces of 45°. Moreover, the MVBF scenario, always by examining the Equivalent Stress of von Mises and the Total deformation in vertical forces, and inclined forces configurations. From an economic point of view, this solution could be advantageous for the customer, because the cost of design and production remains the same.

4.5 Cut and with abutment four-element crown bridge on T-Base

In the clinical aspect, one way to improve the prosthesis was to level the bases of the molar and canine channels at the height of the second pre-molar channel, to guarantee more structural stability (Figure 4.10). One solution could be to put as many intermediate abutments as T-Base channels, it serves to fill the gap between the prosthesis and implants.

Initially, the simulations were performed considering only the cut prosthesis and then of the latter with the abutments, but it makes sense to talk about these cases together in this chapter because the cut prosthesis anyway would need abutments. Except that the implants and/or crowns are raised so that the prosthesis is levelled with the occlusal plane of the upper dental arch. The solution with only cut prosthesis would entail more time to set the implant at the right height to level the occlusal plane with the entire arch, or to modify the geometry of the prosthesis. So, this solution is advantageous from a structural point of view, but the prosthesis or implants must be completely revised and brings up times and costs. In this thesis it was not done because it would take the oral scans of the patient to redesign the prosthesis cut.

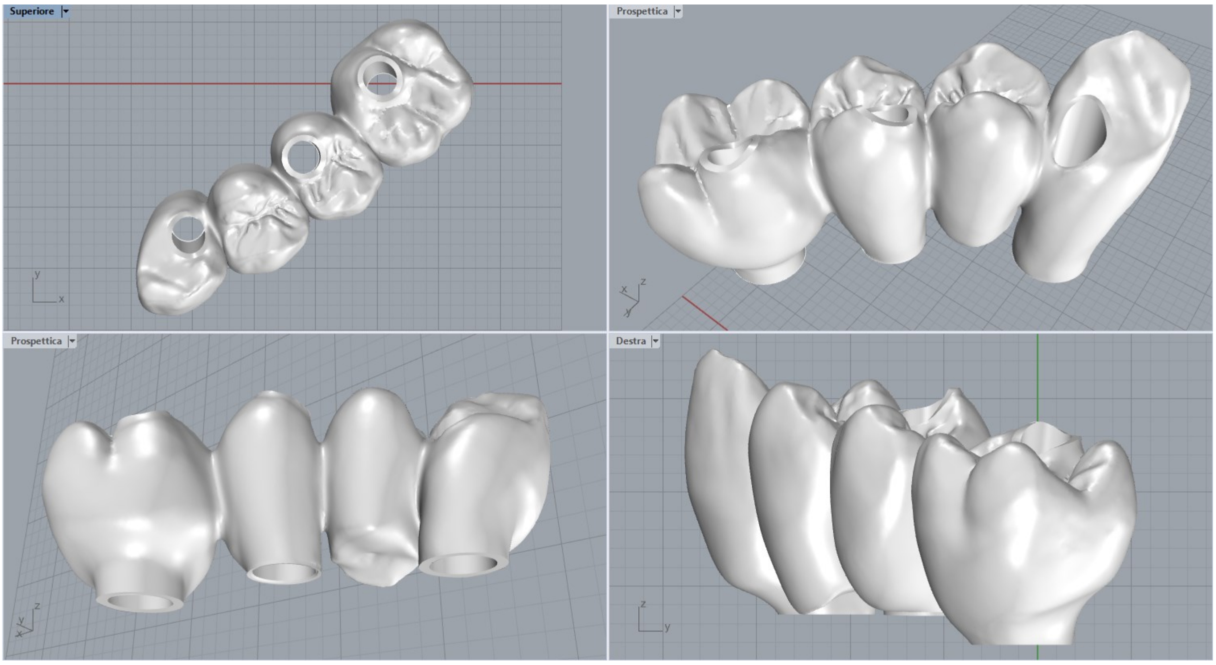


Figure 4.10 Cut prosthesis

The prosthesis was designed on Rhinoceros 3D, as it can see on Figure 4.10, the thickness at the cylinders base augment only by cutting the channels, and so it is very advantageous because a problem related the rupture regards the low thickness of the T-Base housing.

At this point the clinician can decide whether to increase the height of the implants (i.e. the endosseous screws), or the crowns, otherwise puts abutments that act as a bridge between prosthesis and implants. In the case of the second option, further design was considered sequentially.

On Figure 4.11 (a) the cut prosthesis with the abutment interposed, which were designed with Rhinoceros 3D. Instead, on Figure 4.11 (b) the sectioned view of the molar, where it is possible to see better the coupling among the prosthesis and abutment, that follow the exact internal geometry of prosthesis.

The material with which abutments would be made is CoCr, and they would be manufactured with SLM technology. The dark grey items are the abutments that touch with the prosthesis and with the connections of the endosseous screw.

Therefore, the simulations also in this combination were executed to understand if this type of combination, totally feasible from the company Yndetech, would turn out advantageous or not. On this occasion a shared reasoning with the production department was necessary to understand the best solution for materials and production techniques.

Instead, for the customer, from an economic point of view, it is logically disadvantageous because to complete the prosthesis would need two production processes, one to produce the zirconium dioxide prosthesis and another to produce the abutments in CoCr, that are much more expensive in terms of materials.

Moreover, in this case the design phase would require more time and so more costs because the engineer will also have to decrypt the prosthetic connections which will go in contact with the implants.

However, it can be said that the prosthesis with abutment has demonstrated improvements in mechanical performance for stress and deformation, which means it works wonderfully in extreme situation. In fact, higher value of size crack, so lower stress with respect those in normal chewing for failed prosthesis, were detected. This implies that is a suitable device also for people with bruxism. Clearly, it is the best solution but also the most expensive.

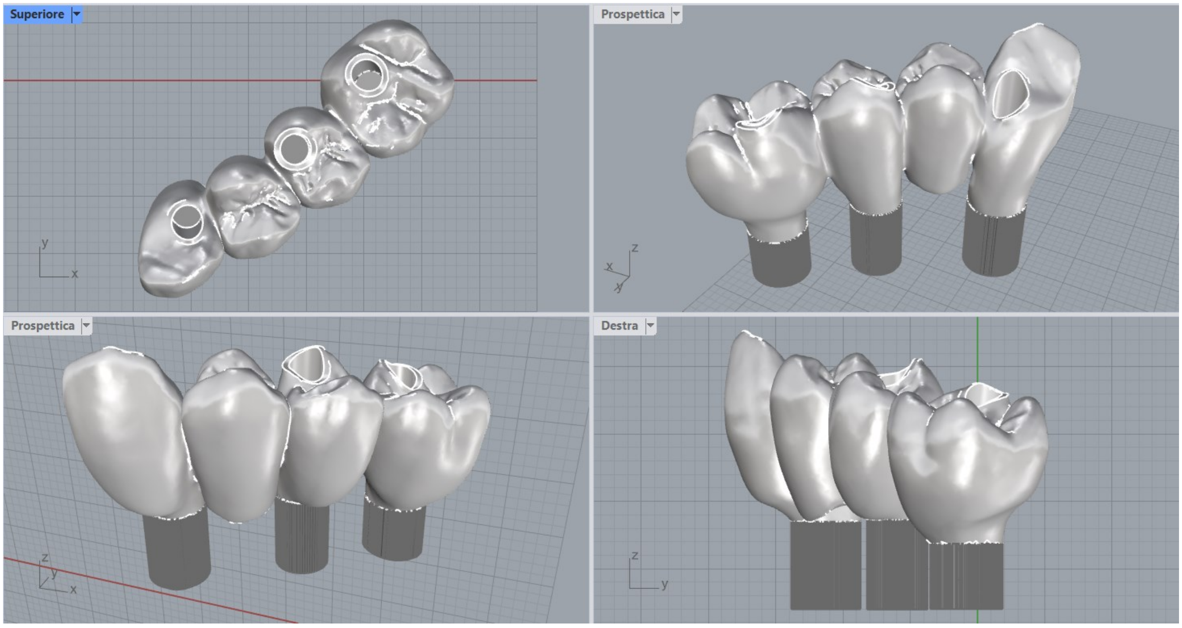


Figure 4.11 (a) Cut prosthesis with abutment

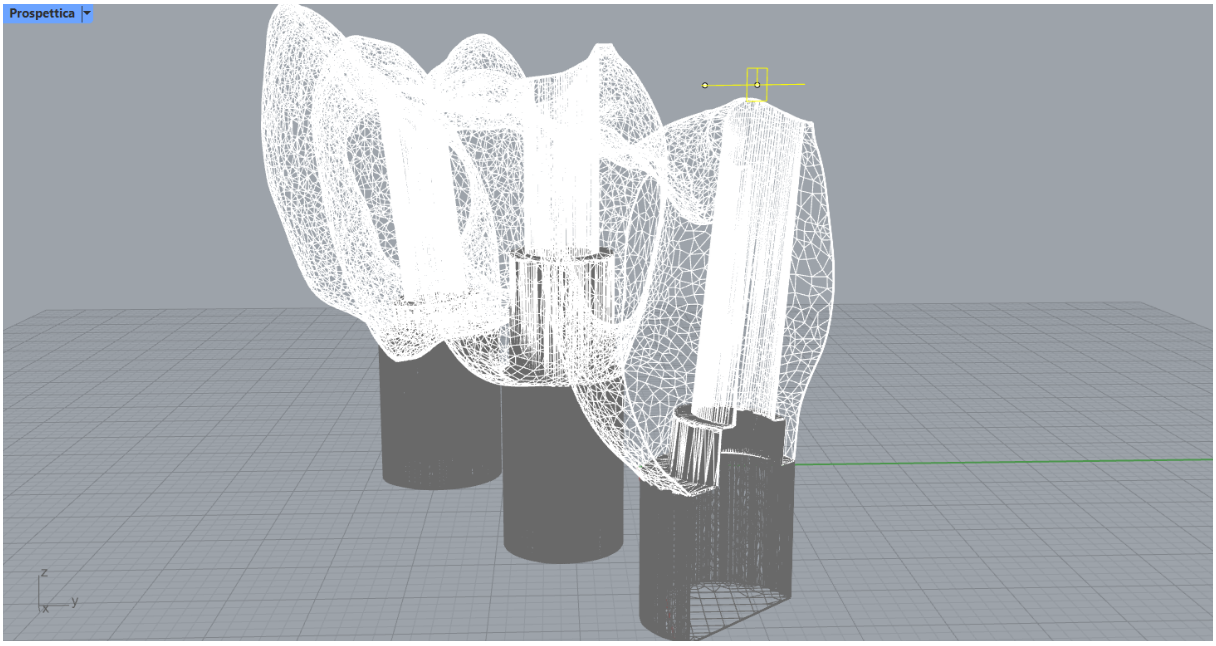


Figure 4.11 (b) Section plane of the Molar

4.6 Validation method

A transversal project was made, so the work of this thesis involves a limited number of simulations to validate the FEM for the company Yndetech, and therefore usable in the R&D department.

Therefore, to understand if the direction of the experiment was right and the results obtained in the various simulations were reliable, an extreme case of a teeth prosthesis made "ad hoc" was designed.

The device concerned is a prosthesis on the natural stump of the tooth number 13 (i.e. canine) and number 12 (i.e. lateral incisive) that is represented on Figure 4.12 (a), where is clear the connective element among teeth is too small in terms of area magnitude, in fact the area is 2 mm^2.

Instead, in Figure 4.12 (b) has been resized the thickness of palatal surface up to 0,200 mm, that is a very low value, far below the gold standard. These values represent an extreme case that will surely be dangerous for structural integrity during chewing and manufacturing.

As it can see on Figure 4.12 (b) a thickness analysis was done and the red value regards a thickness which start from 0,190 mm to 1,000 mm for the blue colour.

The palatal surface is completely red because the file was modified and reduced on Hypsocad, in order to obtain a very low value of thickness to test. The results of simulation in Ansys Workbench R1 2024 regards as for the previous cases the equivalent stress of von Mises and total deformation, which will be compared with the simulation of the case of study; the pre-processing was made in the same manner of failed prosthesis simulations, to have the most likely and comparable data to the case study.

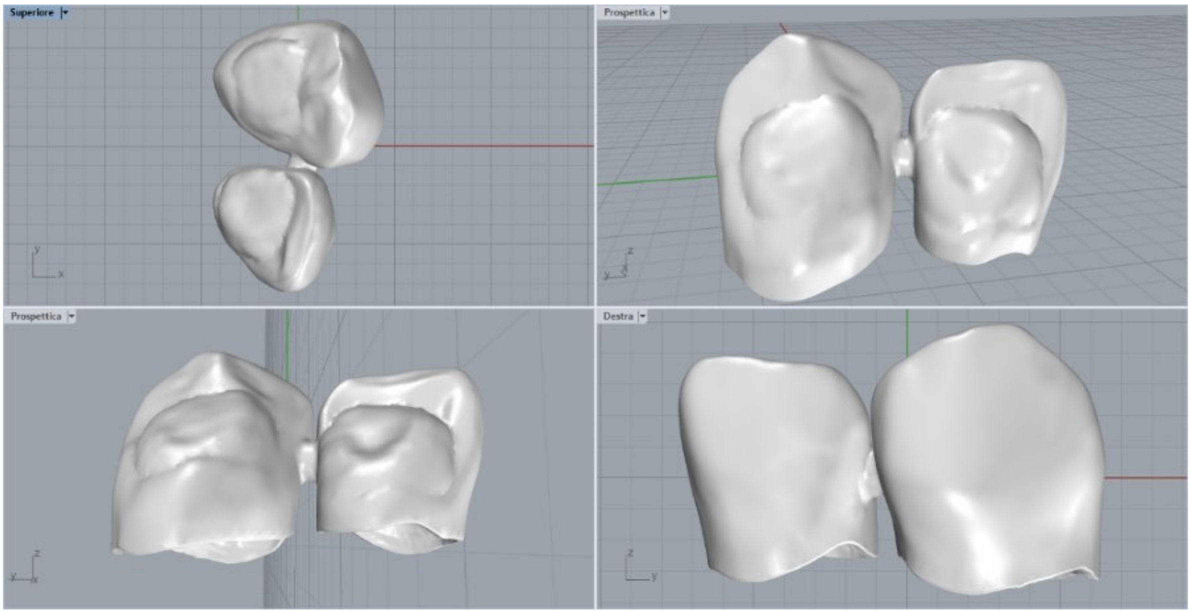


Figure 4.12 (a) Validation prosthesis with low area of Bridge element

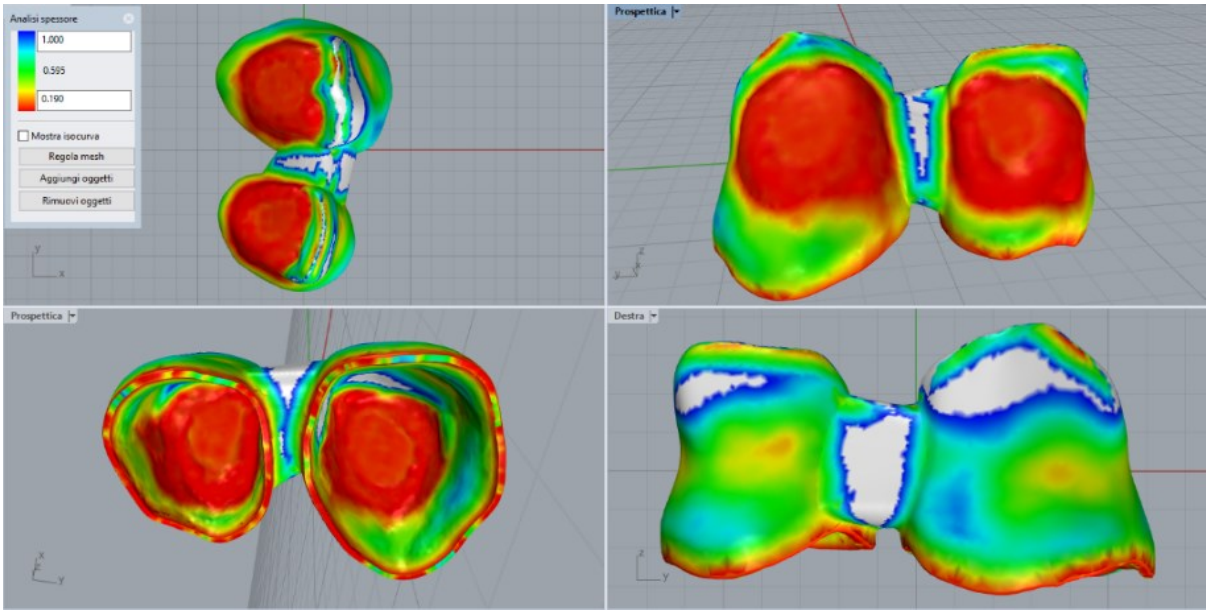


Figure 4.12 (b) Thickness analysis of the validation prosthesis with undersized palatal surface on Rhinoceros 3D

5 Results

In the following paragraphs, the numerical results of the FEA simulations of the failed prosthesis which represents the prosthesis that had broken, and the results of the FEA simulations made on the elements modified to remedy the problem, and validation prostheses.

Remembering that the rupture of the ZrO_2 prosthesis occurred at the bottom of the cylinders that connect the prosthesis to the endosseous screw through the T-Base canals. It is possible to note from the photos (Figures 4.2 (a) – (b)) that the molar breakage has happened on the right side as well as for the canine, while for the second pre-molar the break seems to be circular on the entire base of the crown.

FEA, as said, is a numerical method used to solve engineering problems, it is widely used in various engineering disciplines to predict the behaviour of designs and optimize performance before physical prototypes are built. It allows engineers to study complex systems under different physical conditions and make informed design decisions.

The long-term success and predictability of implant-supported restorations largely depends on the biomechanical forces (i.e. the load bearing and the stress on surfaces) acting on implants and the surrounding alveolar bone in the mandible or maxilla. The aim of this study is to investigate the biomechanical behaviour and understand the reason for such a premature failure. The prosthesis in question is the four-element crown bridge on T-Base, and therefore several modified cases, supported by endosseous implants, under simulated masticatory loads, in the context of different loading schemes, using a three-dimensional finite element analysis (3D-FEA). [29]

5.1 Post-processing results of all FEA prostheses simulations

This chapter aim to visualize and subsequently interpret the results obtained, also called the post-processing step of all FEA simulations. This involves generating contour plots, stress distributions, deformation distributions and any physical phenomenon useful to be understood to improve the final product.

Remember that von Mises stress is a criterion for determining if a given material will yield or fracture, and to obtain a single scalar stress value.

To follow several tables with the values of Equivalent stress of von Mises and total deformations for the different cases studied, 'in primis' (the most important one for the thesis project) the values of the failed prosthesis.

It is useful to recall that the stress values were taken for different types of force depending on the type of tooth, also the simulations were conducted in the normal chewing scenario (e.g. during meals) and for the MVBF (e.g. either bruxism phenomena or intense voluntary bite). [35]

On Figures 5.1 (a) and (b) it is possible to see the equivalent stress of von Mises data labelled on the failed prosthesis, at certain critical points, so in some parts of the prosthesis, respectively for the pressures by considering the vertical forces and with the inclined forces of 45°. These reference points will later be used to take the others and compare them from all FEA simulations done, for more accurate interpretations of results.

Instead, on Figure 5.2 the equivalent stress of von Mises values of the thickened prosthesis. The Figure 5.3 refers to the stress values of the cut prosthesis only and onto the Figure 5.4 the stress data of the cut prosthesis with also the abutment. Then, in Figure 5.5 the sectioned view of the canine where it can see the stress value at the interface between prostheses and abutment, only in this scenario it was taken the stress values also in the interface among the prosthesis and CoCr abutments. Moreover, in Figure 5.6 is represented the total deformation distribution of the failed prosthesis.

At last, on Figures 5.7 (a) - (b) the graphical visualizations of stresses and deformations of the validation prosthesis used to confirm the CAE method, by examining the undersized palatal surface. The critical points of stresses and deformations were taken at the most dangerous areas, above and below the medical device.

The figures, apart from the case study, are depicted in conditions of normal chewing and the pressures with perpendicular forces to the surface, the red label represents the maximum value. This because it would not be useful to illustrate additional images since what is important are the numerical data.

Indeed, Tables 5.1, 5.2, 5.3 and 5.4 group the equivalent stress of von Mises values by analyzing the various simulations of the case of study, and of the various transformations performed to improve the broken prosthesis, respectively.

Pressures with vertical forces and with inclined forces of 45 degrees were considered, and two types of mastication were examined, the normal one and the MVBF (always in the same pressure configurations).

As mentioned above, the data were taken at some reference points, that are the following: in front, sideway, behind and inside the T-Base channels in each tooth (i.e. molar, pre-molar and canine). To clarify and simplify the comparison of the data, the maximum values and the averages of the given physical dimension have also been shown in the tables.

Finally, the Table 5.5 includes the values of the total deformation of just one critical point in each tooth which was broken, and the maximum deformation of each simulation.

Even in this case the scenarios are normal chewing and MVBF, and the pressures at vertical and inclined forces of 45 degrees were valuated.

Instead, the Table 5.6 contains the pressure and deformation data of the FEA made on the validation prostheses by considering only the critical points, so an interval between minimum and maximum stresses and their averages is contained on table. For consistency with the cases studied before, it was analyzed the normal chewing and MVBF in vertical and inclined forces of 45 degrees, as well as for the total deformation of the validation prostheses.

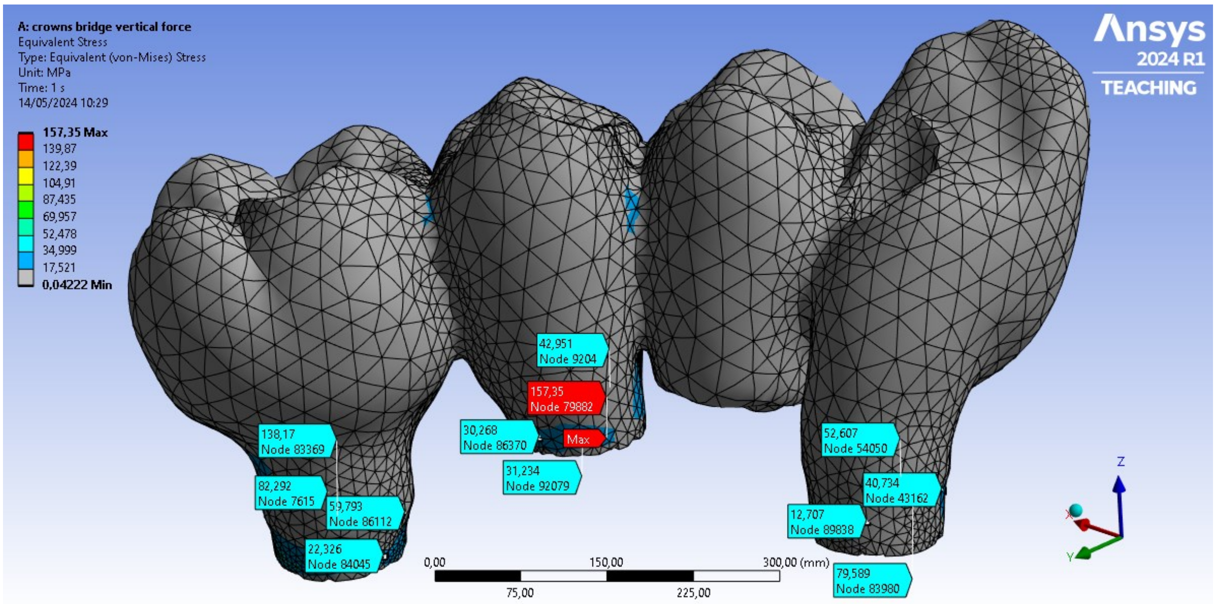


Figure 5.1 (a) Equivalent von Mises stress values of the failed prosthesis for vertical forces

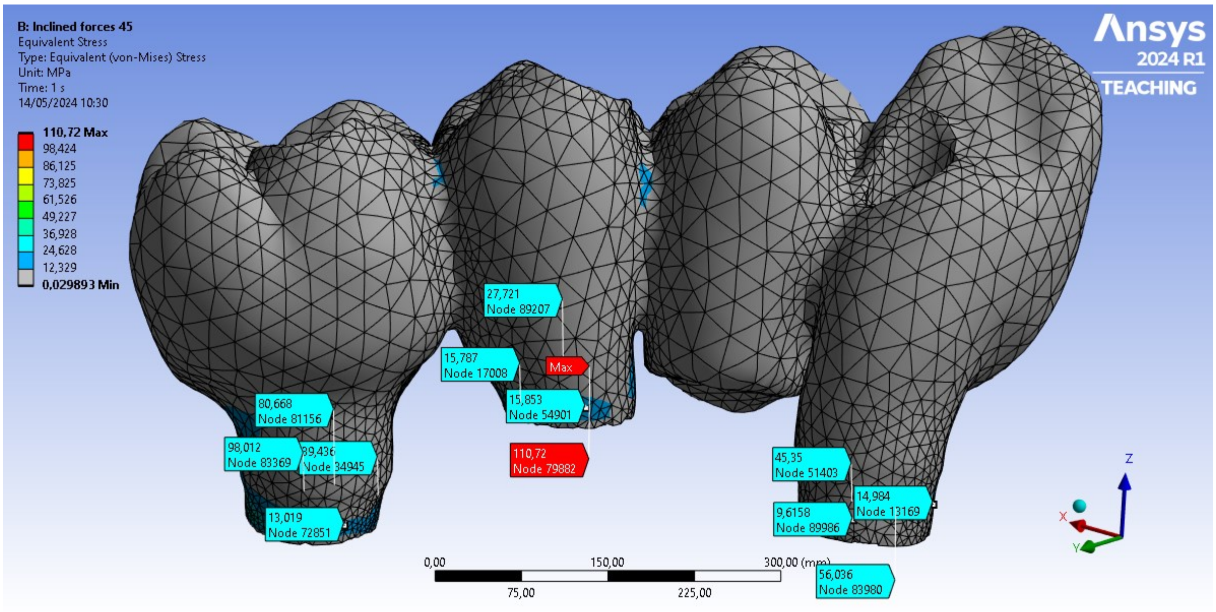


Figure 5.1 (b) Equivalent von Mises stress values of the failed prosthesis for inclined forces of 45°

Table 5.1 Equivalent von Mises stress results of the failed prosthesis

	Molar (MPa)	Pre-molar (MPa)	Canine (MPa)		Molar (MPa)	Pre-molar (MPa)	Canine (MPa)
Vertical forces				Vertical forces			
Normal chewing				MVBF			
Max.157,35 MPa				Max.722,62 MPa			
Average 9,50 MPa				Average 38,49 MPa			
Front	22,33	31,23	12,70		114,46	104,40	48,08
Sideways	59,80	30,27	40,73		213,44	121	125,93
Behind	82,29	157,35	52,60		296	722,62	247,13
Inside	138,17	42,95	79,58		536,90	197,88	196,01
45° Inclined forces				45° Inclined forces			
Normal chewing				MVBF			
Max.110,72 MPa				Max.517,06 MPa			

Average 6,72 MPa				Average 27,34 MPa			
Front	13,02	15,85	9,61		80,32	86,34	34,85
Sideways	39,43	15,78	14,98		160,94	57,80	121,02
Behind	80,66	110,72	45,35		188,20	517,06	152,83
Inside	98,01	27,72	56,03		380,56	131,72	148,26

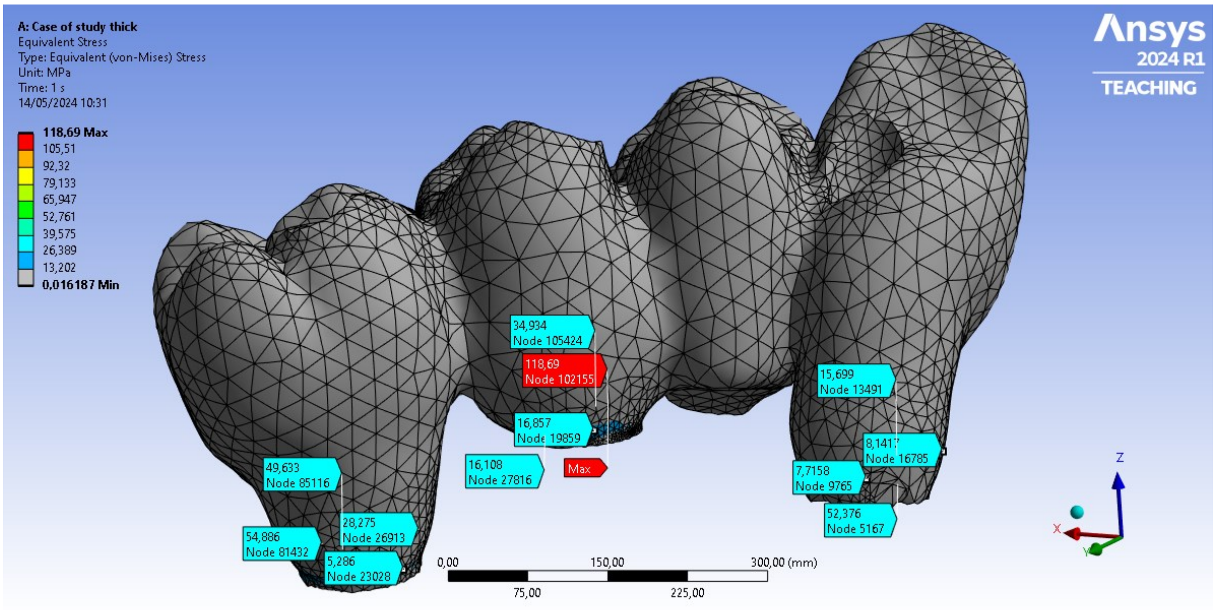


Figure 5.2 Equivalent von Mises stress values of thickened prosthesis

Table 5.2 Equivalent von Mises stress results of thickened prosthesis

	Molar (MPa)	Pre-molar (MPa)	Canine (MPa)		Molar (MPa)	Pre-molar (MPa)	Canine (MPa)
Vertical forces				Vertical forces			
Normal chewing				MVBF			
Max.118,69 MPa				Max.556,47 MPa			
Average 6,42 MPa				Average 26,92 MPa			
Front	5,28	16,86	7,72		14,36	56,38	23,56
Sideways	28,28	16,11	8,14		60,08	66,22	74,76
Behind	54,89	118,69	15,70		155,82	556,47	180,89
Inside	49,63	34,93	52,38		88,75	128,36	393,04

45° Inclined forces				45° Inclined forces			
Normal chewing				MVBF			
Max.83,48 MPa				Max.398,62 MPa			
Average 4,53 MPa				Average 19,16 MPa			
Front	2,93	9,96	1,77		4,75	22,76	9,13
Sideways	9,66	6,47	5,74		14	13,20	48,03
Behind	38,93	83,48	9,88		37,73	398,62	122,90
Inside	36,30	23,42	26,41		55,88	71,20	47,28

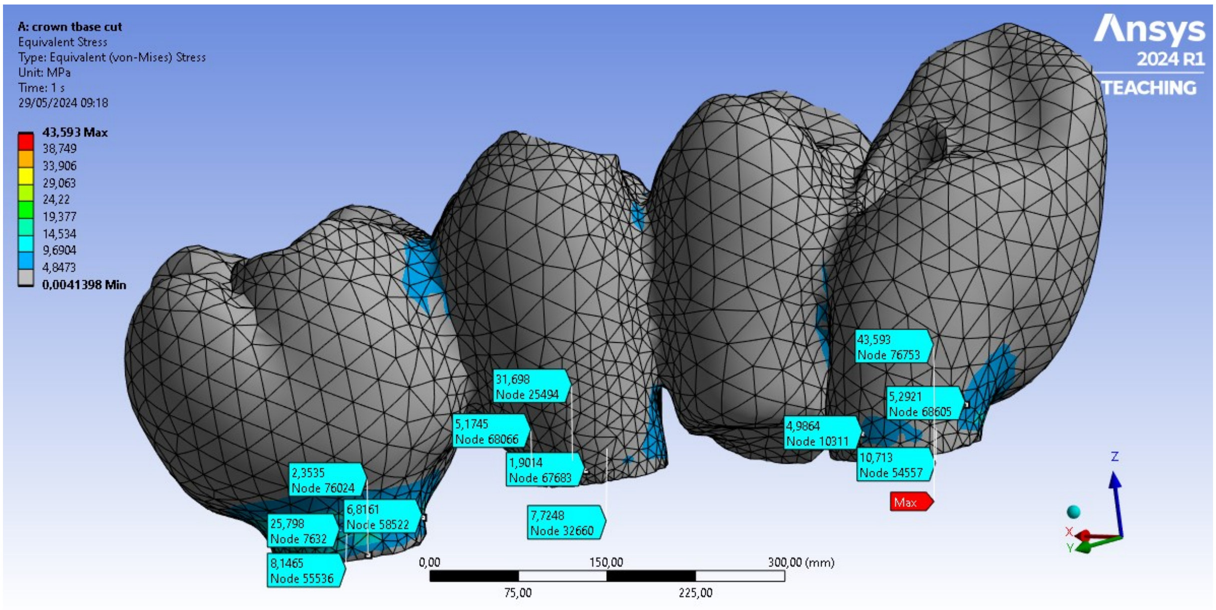


Figure 5.3 Equivalent von Mises stress values of cut prosthesis

Table 5.3 Equivalent von Mises stress result of cut prosthesis

	Molar (MPa)	Pre-molar (MPa)	Canine (MPa)		Molar (MPa)	Pre-molar (MPa)	Canine (MPa)
Vertical forces				Vertical forces			
Normal chewing				MVBF			
Max.43,59 MPa				Max.186,79 MPa			

Average 4,06 MPa				Average 17,45 MPa			
Front	2,35	1,90	4,98		18,92	6,64	21,20
Sideways	6,81	5,17	5,29		50,56	19,33	26,44
Behind	25,80	31,70	43,59		106,91	183,17	186,79
Inside	8,15	7,72	10,71		25,17	22,09	36,95
45° Inclined forces				45° Inclined forces			
Normal chewing				MVBF			
Max.30,69 MPa				Max.132,97 MPa			
Average 2,87 MPa				Average 12,44 MPa			
Front	3,83	0,96	4,26		5,56	4,73	17,82
Sideways	6,08	5,03	4,92		12,27	11,23	25,97
Behind	24,59	26,24	30,69		31,08	131,58	132,97
Inside	7,07	4,73	6,66		31,14	18,78	25,90

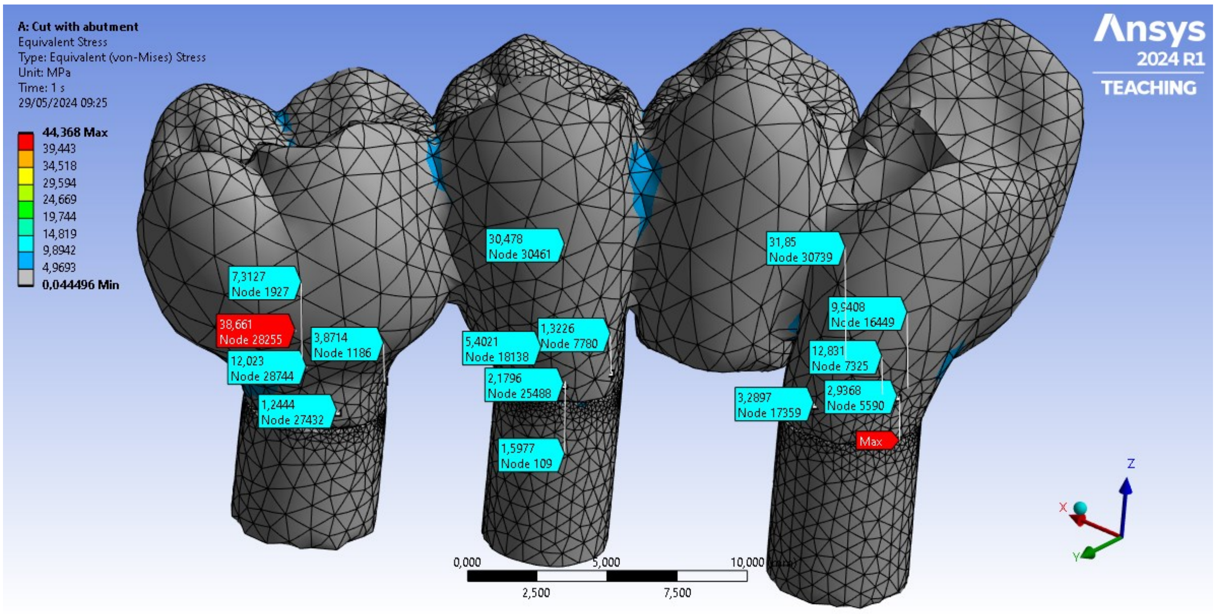


Figure 5.4 Equivalent von Mises stress values of cut prosthesis with abutment

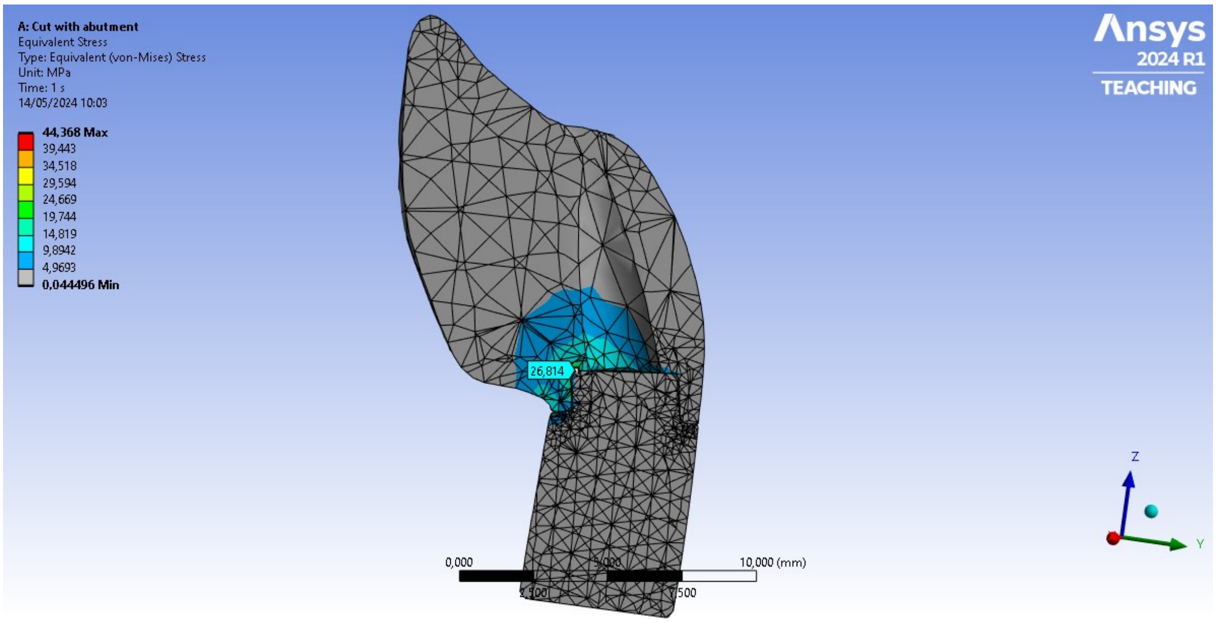


Figure 5.5 Canine section plane with equivalent von Mises stress value at the interface

Table 5.4 Equivalent von Mises stress result of cut prosthesis with abutment

	Molar (MPa)	Pre-molar (MPa)	Canine (MPa)		Molar (MPa)	Pre-molar (MPa)	Canine (MPa)
Vertical forces				Vertical forces			
Normal chewing				MVBF			
Max.38,66 MPa				Max.147,63 MPa			
Average 2,60 MPa				Average 10,86 MPa			
Front	1,24	2,18	3,28		8,97	10,99	13,41
Sideways	3,87	1,32	2,93		9,84	8,36	7,53
Behind	12,02	5,40	12,83		46,29	27,03	72,21
Inside	7,31	1,59	9,94		8,42	3,94	5,22
Interface	38,66	30,48	26,81		147,63	142,19	140,18
Abutment Max.44,37 MPa				Abutment Max.182,12 MPa			
45° Inclined forces				45° Inclined forces			
Normal chewing				MVBF			

Max.27,44 MPa				Max.104,52 MPa			
Average 1,84 MPa				Average 7,73 MPa			
Front	1,75	1	2,54		3,30	6,07	8,15
Sideways	1,71	0,82	1,42		4,04	5,19	12,14
Behind	8,53	3,80	12,01		14,81	18,03	36,03
Inside	4,97	1,04	1,14		4,58	3,17	3,73
Interface	27,44	17,80	22,41		104,52	101,86	99,96
Abutment Max.31,27 MPa				Abutment Max.129,23 MPa			

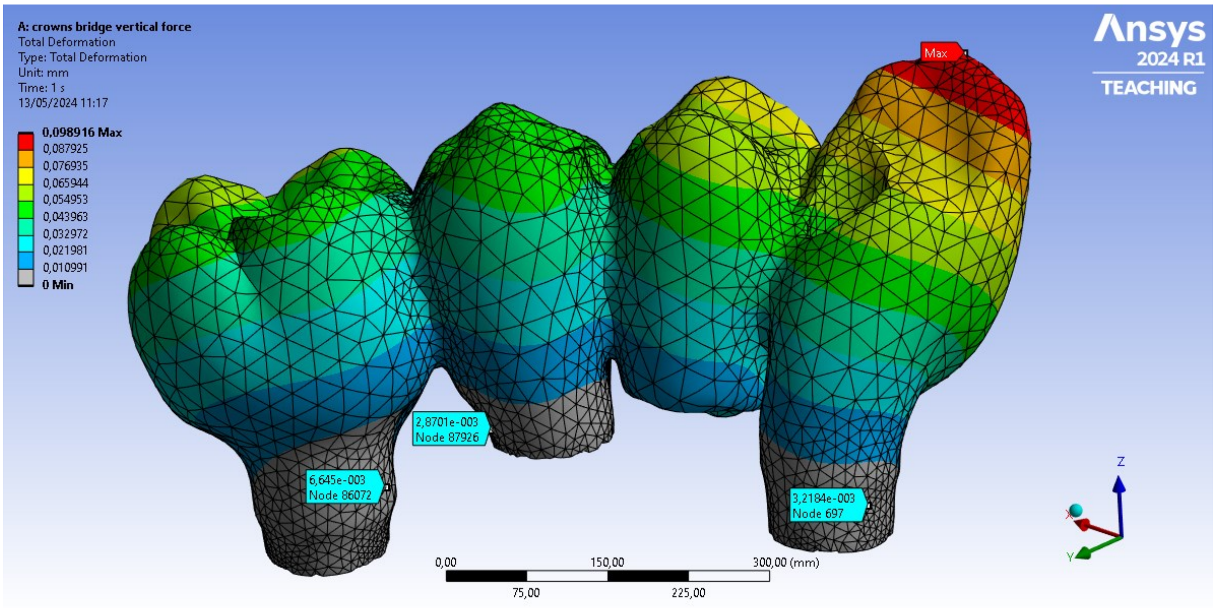


Figure 5.6 Total deformation values of the failed prosthesis

Table 5.5 Total deformation results of all FEA simulations

	Max.	Molar (mm)	Pre-molar (mm)	Canine (mm)		Max.	Molar (mm)	Pre-molar (mm)	Canine (mm)
Vertical forces					Vertical forces				
Normal chewing					MVBF				
Failed	0,0989	0,0064	0,0028	0,0032		0,43	0,0254	0,0067	0,0076

Thickened	0,0526	0,0017	0,0015	0,0019		0,22	0,0053	0,0023	0,0065
Cut	0,0262	0	0,0001	0		0,11	0,0027	0,0005	0,0010
Abutment	0,0009	0	0	0		0,00 40	0	0,0030	0,0001
45° Inclined forces					45° Inclin ed forces				
Normal chewing					MVBF				
Failed	0,0696	0,0037	0,0036	0,0019		0,30	0,0150	0,0123	0,0092
Thickened	0,0370	0,0017	0,0009	0,0011		0,16	0,0030	0,0066	0,0040
Cut	0,0185	0	0	0		0,08	0,0020	0,0003	0,0021
Abutment	0,0007	0	0	0		0,00 28	0	0,0002	0

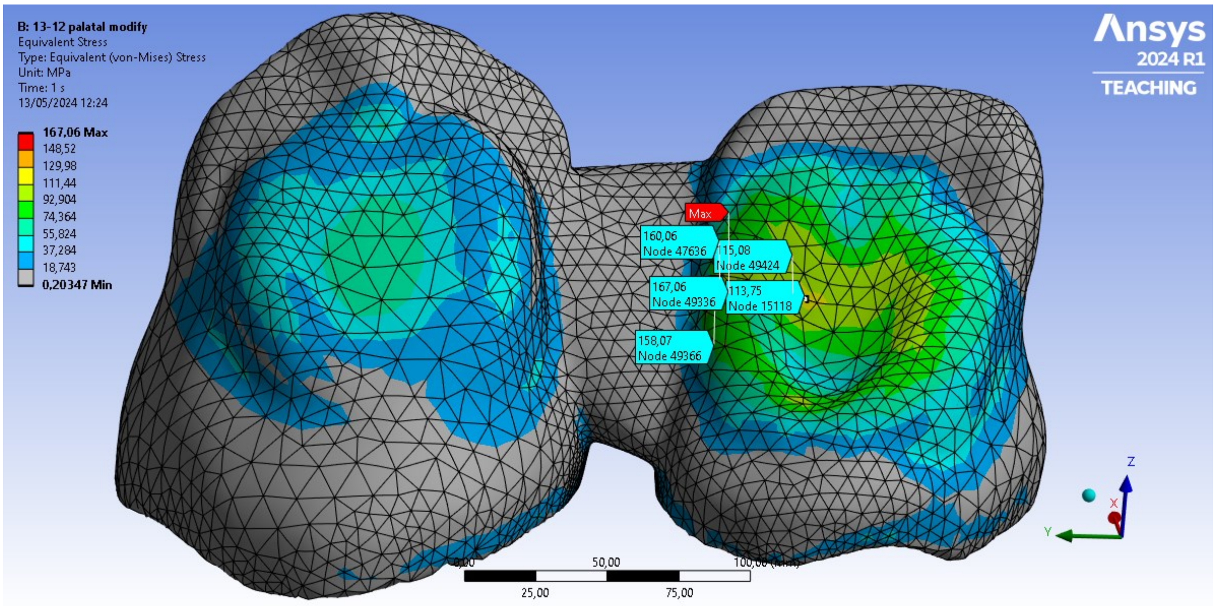


Figure 5.7 (a) Equivalent von Mises stress of the critical points in validation prosthesis

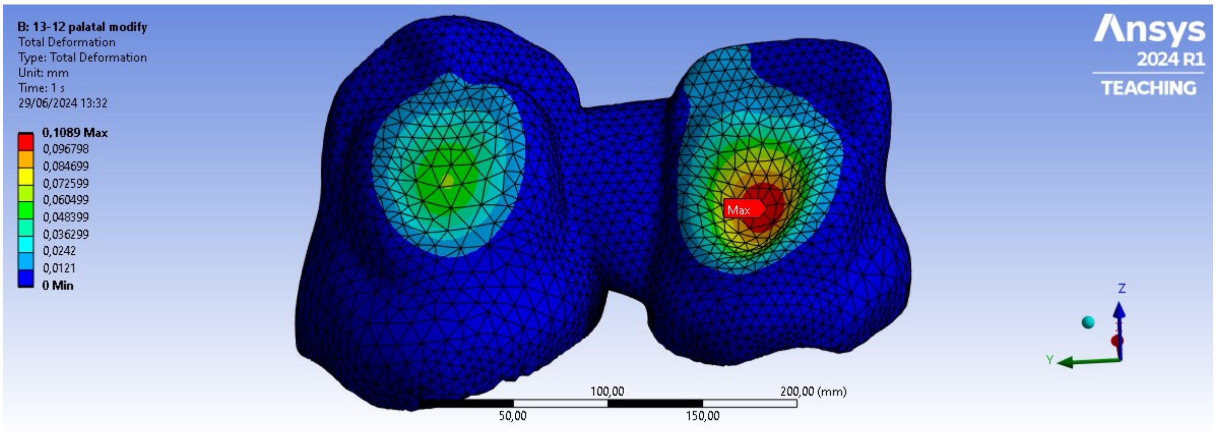


Figure 5.7 (b) Total deformation of the validation prosthesis

Table 5.6 Equivalent von Mises stress and total deformation data of validation prosthesis

	Undersized bridge element (MPa)	Resizing palatal surface (MPa)		Undersized bridge element (MPa)	Resizing palatal surface (MPa)
Vertical forces			Vertical forces		
Normal chewing			MVBF		
Critical points	6,30 – 12,33	113,75 – 167,06		15,86 – 17,43	202,04 – 259,06
Average (MPa)	8,34	14,47		15 ,27	26,46
Tot. Def. (mm)	0,0052	0,1089		0,0099	0,1687
45° Inclined forces			45° Inclined forces		
Normal chewing			MVBF		
Critical points	6,15 – 11,04	80,76 – 118,62		10,70 – 12,65	143,05 – 183,40
Average (MPa)	5,91	10,25		10,81	18,73
Tot. Def. (mm)	0,0038	0,0773		0,0070	0,1194

5.2 Discussion

In this chapter it wants to discuss the results obtained from the various simulations. Dental prostheses are subjected to high stress and FEM simulations, by considering some mechanical properties, gives further certainties in ensuring structural integrity and safety.

Logically, starting from the failed prosthesis and then evaluating the data of the cases in which changes have been made, which it has been discussed in previous chapters. Clinical hypotheses must be validated through engineering evaluation and assessed for benefits to the client and the company.

The expected results include a detailed map of the stress and strain distributions for all type of prostheses, to give an answer to the problem, and by finding various solutions to the problem.

As already said a FEA simulation is a powerful tool for structural analysis, the FEA simulations will provide a detailed understanding of the structural response of the prosthesis under masticatory load, helping to improve the design and safety of such medical devices.

The main mechanical properties to detect breakage are the following: fracture toughness (K_{Ic}) and the flexural strength. The K_{Ic} is directly attributable to the toughness of a material. The toughness is a general measure of a material's ability to absorb energy and deform without fracture, while K_{Ic} specifically quantifies the propagation resistance of an existing crack or fracture. In other words, fracture toughness is a specific aspect of toughness, focused on the material's ability to withstand the growth of a stress fracture.

But first, factor K_I, also known as 'stress intensity factor', it is also called 'Mode I', and it describes the distribution of stress at the peak of a crack in which the load is applied orthogonally to the crack plane, and the opening is perpendicular to the crack; namely tensile load is perpendicular to crack surface, and mode I indicates crack opening under the perpendicular load. 'Mode II and III' are for slip and tear fracture respectively but are not considered in this project, these situations consider shear stresses inside and outside the crack plane. For a dental prosthesis where it wants to study fractures it is generally correct to consider the Mode I stress intensity factor. This is because the predominant fracture mode in these cases is usually the opening, that is tensile forces perpendicular to the crack plane tend to open the fracture. However, there may be situations where modes II and III may also be relevant, depending on the specific geometry and loads applied to the dental implant and the degree of complexity of the problem, and the accuracy it wants to achieve. For this project it may be enough to rely on K_{Ic}.

Hence, K_{Ic} is a parameter that represents the limit value of K_I above which unstable crack propagation occurs, when the equation $K_I \geq K_{Ic}$ is satisfied, the crack advance irreparably to leading to fracture of the material. Instead, if it is kept $K_I < K_{Ic}$ the material is safe, and the crack will not spread.

K_{Ic} is named 'critical stress intensity factor', which is a measure of the material's resistance to fracture when a crack is present, and the grown of crack propagation suddenly becomes rapid and unlimited.

It is a material property that describes its ability to resist the propagation of cracks. So, it is particularly important in engineering applications where materials are subjected to mechanical loading or any stress.

It is an intrinsic property of the material, which depends on its chemical composition, microstructure, and the possible presence of defects or impurities. Essentially, it measures how much energy a material can absorb before it causes fractures through crack spread. Higher K_{Ic} indicates greater resistance to crack propagation, making a material more reliable and safer under stress.

K_{Ic} is measured in units of energy times unit area (i.e. MPa$\sqrt{mm}$); the stress, evaluated in MPa multiplied by the square root of the crack length, needed to propagate a crack in the material which causes a rupture. It characterizes the resistance of the material to crack propagation.

It is a key parameter in fracture mechanics, widely used in several engineering field to assess the integrity and durability of materials under stress. Be aware of the fracture toughness is crucial in materials selection for engineering applications, where materials are frequently exposed to high stress and potential for crack propagation.

It shows how a specific stress acts on the square root crack length itself; so, this unit indicates the amount of pressure needed to propagate a crack in prosthesis's material, with a specific length.

It is important to emphasize that K_{Ic} is a material property associated with dynamic conditions. The project involves static analysis, and the K_{Ic} is used to calculate the critical lengths of cracks, with the general formula shown below, which identify the points at risk of propagation of cracks under a given stress. The critical crack lengths thus become indicators of danger for the fracture.

Instead, the flexural strength, also known as 'modulus of rupture', is a measure of a material's ability to resist deformation under bending, it is measured in unit of pressure (MPa).

Flexural strength indicates the maximum stress a material can withstand before it breaks or fractures when subjected to bending forces. Flexural strength is a critical parameter in material selection for structural applications, as it helps engineers assess the material's suitability for withstanding bending loads and resist failure. So, the maximum fracture stress is then recorded as flexural strength. The fracture mechanics of overcoming flexural strength are presented as a ductile fracture, so followed by plastic deformation. The dental prostheses undergo flexor loads and therefore it is correct to evaluate also this mechanical property. [36]

However, particular attention must be paid to K_{Ic} values because it indicates the limit value for fatigue breakages, which are dangerous because they are immediate and unexpected, moreover it could also be much lower than flexural strength.

In dental prosthetic, K_{Ic} is one of the most important mechanical properties regarding the materials used for dental designs, to evaluate the safety of the prosthesis. In fact, during the production of such medical devices can occur micro-fractures that subjected to high stresses due to masticatory load could be lethal for structural reliability.

Furthermore, subjected to a load that varies in a range, the repeated stresses of the loading cycles over time produces a considerable fatigue inside the material which could lead to the breakage.

As already mentioned, the fracture can occur unexpectedly even if the value of K_{Ic} is far below than flexural strength (i.e. the maximum fracture resistance value). Therefore, the causes are not due to internal forces but to a small superficial crack that with the time, and the masticatory loads, widens until the total abrupt and sudden collapse. [37]

Should be considered the pressures values on each tooth both for normal chewing and for the MVBF, and the number of chewing cycles that led to the rupture and the technical characteristics of the material used, as already illustrated on Table 4.1.

Remember that the simulations presented reproduce a single mastication and just out of the factory (i.e. the best condition for the prosthesis), which means that a stress-life simulation was not made. The number of cycles (394.200) can explain quantitatively the failure of the prosthesis.

The technical parameters of the ZrO_2 considered more (shown in Table 4.1) are the following:

- $K_{Ic} \geq 158{,}11\ \mathrm{MPa}\sqrt{mm}$;
- Flexural strength = 1000 MPa.

The maximum value of stress recorded in FEA simualtions is lower than the flexural strength of the ZrO_2, so this explains why there was not cleavage rupture but a fatigue rupture due to repeated loads; for this reason, it is more logical to dwell on the K_{Ic}. If a material has a $K_{Ic} \geq 158{,}11$ MPa$\sqrt{mm}$, it means that can tolerate the presence of a certain size of defects without fracture; about 0,319 mm. Therefore, if the K_I reaches or overtakes the K_{Ic} value (i.e. $K_I \geq K_{Ic}$), the material can no longer stop the propagation of the crack and this spreads rapidly, by leading to the catastrophic fracture.

It is necessary to make a posteriori reasoning, since from the simulations only stress values and deformations are known but not any defects. Therefore, it is necessary to introduce the general formula of the K_I, it can be applied using K_{Ic} (i.e. $K_I = K_{Ic}$), and it is the following:

$$K_{Ic} = Y\sigma\sqrt{\pi a}$$

- Y is a geometric factor that depends on the shape and orientation of the defect (a useful simplification is equal to 1, which is a good approximation for this project. But for a more precise analysis, especially for complex geometries and specific load conditions, it would be advisable to use tables with various values of Y);
- σ is the stress applied;
- a is the critical length of the defect.

Knowing σ and the limit of K_{Ic} it can find the critical crack length (a) values inside the prosthesis to evaluate if there may have been a break. At least for maximum stress in the prosthesis. In an initial approach, the critical crack length values were calculated by considering only the maximum stress values applied for each configuration, because it is logical that if stress decreases the size of the crack increases, and it is more difficult to have a dangerous environment which would lead to a failure. If the critical size of cliques increases it is more unlikely to reach it. The data can be viewed on Table 5.7.

Table 5.7 Critical crack length values at the maximum stress applied

Normal chewing		
	Max. Stress (σ) (MPa)	Critical length (a) (mm)
Failed	157,35	0,321
Thickened	118,69	0,564
Cut	43,59	4,18
Abutment	38,66	5,32

MVBF		
	Max. Stress (σ) (MPa)	Critical length (a) (mm)
Failed	722,62	0,015
Thickened	556,47	0,025
Cut	186,79	0,228
Abutment	147,63	0,365

From this data can be deduced that if the defect turns out to be smaller than a then the material does not fracture under the applied stress, on the contrary if it proves to be greater with respect the crack would spread unlimited up to the rupture under the stress applied.

Focusing on the case of study data it is plausible a break for the failed prosthesis because considering a normal chewing, with a maximum stress of 157,35 MPa found behind the pre-molar, the critical length is of 0,321 mm, this last computed with the K_{Ic} formula. Even with the configuration of MVBF in the case of study with a stress of 722,62 MPa, a critical length of 0,015 mm is achieved. If, hypothetically, there was a clique of 0,321 mm and a stress applied of 157,35 MPa, this would have brought the K_I to the K_{Ic} and would have led to a fracture instantly.

It is likely that the values between 0,015 and 0,321 mm may be the size of a manufacturing defects in ceramics. So, it can make a hypothetical reasoning, if there was a defect either greater or equal of 0,015 or 0,321 mm under a load of 722,62 or 157,35 MPa respectively, the prosthesis surely would have gone into crash fracture. Instead, if the stress was lower or critical crack length higher with respect the previous values, it would have resisted to masticatory loads.

It must be said that a phenomenon representative of this thesis is fatigue, it is a mechanical phenomenon of progressive degradation of a material subjected to loads that vary over time, and that can lead to the fatigue breakdown of the material even if it falls within the limits of elasticity. Indeed, by processing the data of the failed prosthesis, it is likely that during production were created cracks close to the standard dimension of zirconia defects or it is likely that after about 394.200 chewing cycles, the size of the micro-structural cracks exceeded or have reached values between 0,015 and 0,321 mm, which with the above-mentioned stresses, caused the T-Base elements to break. Hypothetically, at the beginning, it could have occurred an unexpected fracture behind the pre-molar and then created an imbalance of the entire medical device, causing much greater stress in the most vulnerable areas.

As it can see on Figure 5.8, the graph of maximum stress for normal chewing (column to left) and for MVBF (column to right) with the orange line that depict the critical crack length values for all cases.

The trend of critical crack length is very positive going towards maximum stress values of improved cases (i.e. thickened, cut and with abutment prostheses) for normal chewing and less but always growing for MVBF. By considering the normal chewing for the cut and with abutment prostheses the chances of breakage are minimal because the a values are very high, whereas the thickened prosthesis has a "border line" a value.

Clearly for MVBF the values of a drops dramatically but it is comforting to see that even for the maximum possible stress, in the cut prosthesis with intermediate abutment the values obtained are higher than those of normal chewing in the case of broken prosthesis. It means the probability of breakage, for prosthesis with abutment, is almost null in both configurations. Instead, for the maximum stress data of failed, thickened and only cut prosthesis the values of a is dangerously low for the MVBF.

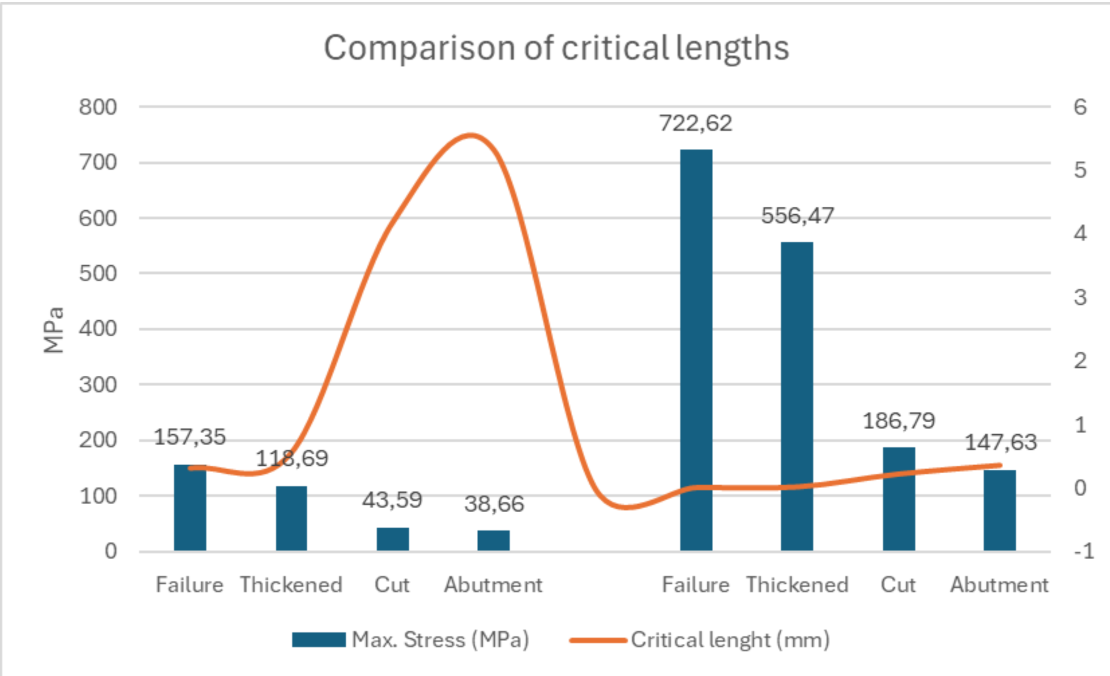


Figure 5.8 Comparison of critical crack lengths for normal chewing and MVBF for all cases

In general, to produce dental prostheses in zirconium dioxide it wants to obtain a value of the crack size of 0,100 mm, which means a maximum applied stress of 282,14 MPa, it is largely exceeded in the MVBF scenario for the broken and thickened prosthesis.

However, defects such as cracks, porosity, inclusions, or surface imperfections can easily fit within this dimensional scale in a range of [0,100 - 0,400] mm (it depends on the type of production and finishing), thus affecting the quality and structural integrity of the material produced. [38]

This means that it is possible to make a quantitative analysis, by considering a critical length range of defects of [0,100 - 0,400] mm, the limit stress range is [282,14 – 140,99] MPa. The concept is: while not knowing if there are defects in the prosthesis but knowing the dimensional range of ZrO_2 or material used, the design must ensure that the stress values applied remain outside the calculated stress range, by implementing a conservative design. The larger the size of the cracks and the less stress will be needed to propagate them and vice versa. The best durability is achieved with very small defects and low stress values, resulting in high critical crack lengths. [39]

This is a posteriori argument because it does not know any crack size at the time of the production of the prosthesis. It must also consider that values of the defect lengths close to a range, by regarding the maximum stresses applied can represent a problem because with the chewing cycles these can increase and going towards rupture.

Without information on the size of the cracks, it is not possible to make a precise risk assessment, but it is possible to recognize that some stress values compared to the K_{Ic} could represent a risk for the structural integrity of the prosthesis, by considering the standard dimensions of the defects for the type of material and processing.

As said, the critical length values increase considerably with the improvements made for the prosthesis, namely the increasing of thickness, the levelling, and with the insertion of intermediate abutments. This means that the probability of breakage due to a crack or defects decreases because it should be larger than standard dimension and, in that case, would be suitable for putting on the patient.

With higher stress values, it is more likely that the dental prosthesis will break, because a decreases and fractures become more plausible due to microstructural defects that may occur within the element.

However, without knowing the actual size of the cracks, no definitive answer can be given. For a precise assessment, a detailed analysis of the cracks present would be necessary in post-production phase.

It is also logical that the forces with which the pressures were calculated can be angled in any direction but also in accordance with several scientific articles; to have reliable results it was necessary to have perpendicular and 45° inclined forces with respect the occlusal plane.

Moreover, the points where stress values were taken are almost the same. It is impossible to have the same nodes for every prosthesis because they are in the order of one hundred thousand nodes.

All these values are of fundamental importance in the analysis of the failed prosthesis, and those of the consequent geometric variations to obtain valid alternatives to the problem.

It has been said in previous chapters that the prosthesis was held for 438 days, which means that the rupture was not immediate but begins with the initiation of micro-crack on the surface of the material (due to defects), that propagate with the progress of the load cycles up to a limit dimension that with a certain stress, K_I reaches K_{Ic}, and there were critical events in fracture mechanics, namely 'critical crack propagation' that causes the breakage.

Before going into the various simulations performed on the case study and the suitably modified prostheses, it is interesting to see on graphs how the maximum and average stress values decrease through the various cases by regarding the normal chewing, as seen on Figures 5.9 (a) - (b).

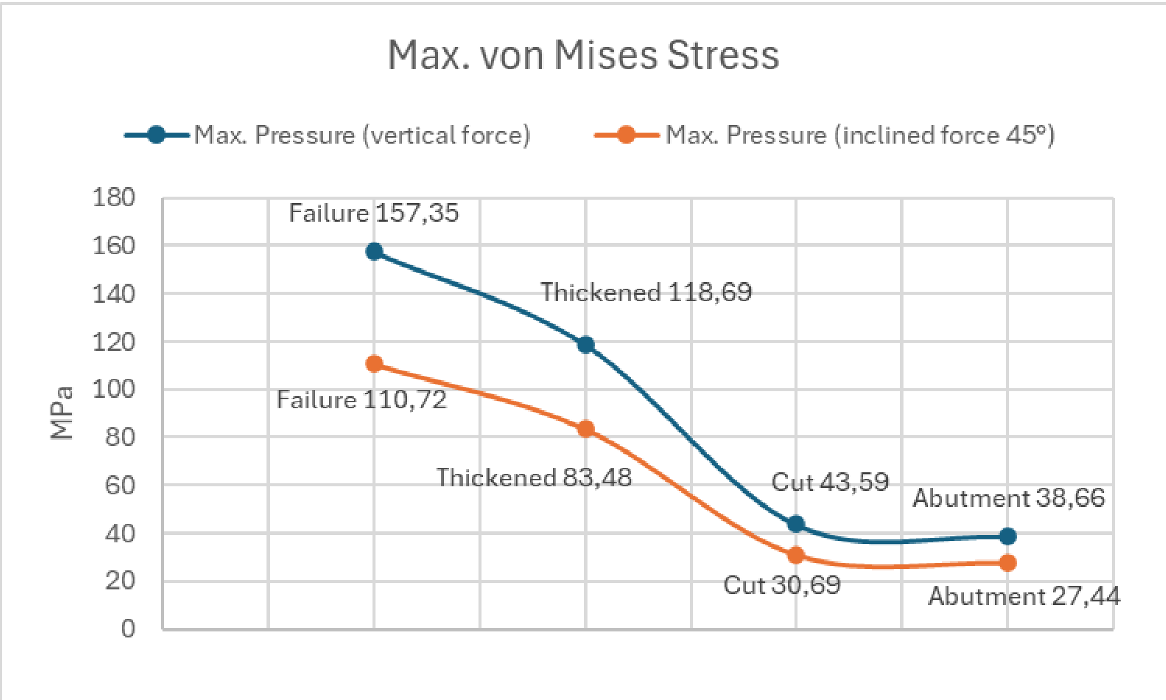


Figure 5.9 (a) Maximum von Mises stress for all cases in normal chewing scenario

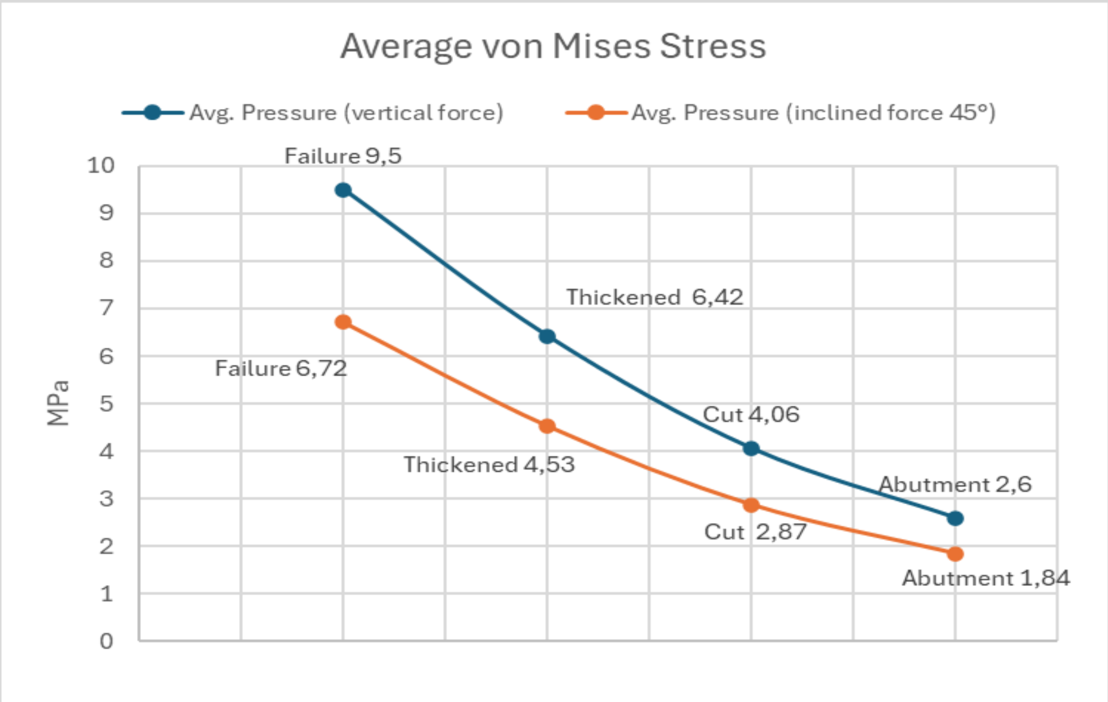


Figure 5.9 (b) Average von Mises stress for all cases in normal chewing scenario

It is noted that, both for the maximum stress values and for the average found, the stress decreases considerably towards the amended cases. Thus, a list of the percentages of decrease between the maximum stress of the failed prosthesis and the modified ones:

- 24,58% stress decrease between the failed prosthesis and thickened one;
- 72,32% stress decrease between the failed prosthesis and cut one;
- 75,45% stress decrease between the failed prosthesis and cut with abutment one.

Clearly there is a considerable advantage in using the cut prosthesis and that with abutment, but this also means increasing costs because surely there will be a need for metal abutments or additional design performance. The thickened prosthesis could be a solution, since Yndetech does not want to increase the price of the prosthesis and then guarantee acceptable performance, in according with the customer; this is the simplest solution. It is certainly lost in structural safety because there is a decrease in stress of 63,28% between the thickened prosthesis and that cut.

The prosthesis, in which the T-Base channels have been levelled all at the same height, is interesting since many different solutions can be adopted, the one considered in this thesis has used abutments in CoCr, but it is also possible leave it thus increasing the lengths of the implants or the device itself, in fact simulations just of the cut device was performed. The abutments are very resistant and suitable for this type of prosthesis at four-element, and the tensive values decreases further with respect the cut prosthesis.

Moreover, among cut prosthesis and the prosthesis with abutment a stress decrease of 11,31% is achieved. The averages data of von Mises stress for all cases confirm the inferences just made.

Instead, on Figure 5.10 (a) it is possible to view the maximum von Mises Stress in MVBF scenario. The percentage of stress decreasing are the following:

- 22,99% stress decrease between the failed prosthesis and thickened one;
- 74,14% stress decrease between the failed prosthesis and cut one;
- 79,59% stress decrease between the failed prosthesis and cut with abutment one.

Compared to normal chewing, it notes that despite the MVBF scenario is certainly more aggressive, the cut prosthesis and the prosthesis with abutment have higher decrease in stress, by valuating in percentage terms. The same cannot be said between the failed and the thickened ones where the decreasing stress has worsened slightly than normal chewing.

Instead, between the thickened and the cut prosthesis there is a decrease in stress of about 66,45% which is higher with respect the normal chewing, in fact a trend steeper can be appreciate on Figure 5.10 (a). These data confirm the strong structural stability and safety of the cut prosthesis, and that equipped with abutment. Also, the reduction in stress is higher between the only cut prosthesis and that fitted with the abutment with respect the normal chewing; indeed, it has a value of 20,97%.

The Figure 5.10 (b) shows the values of the average stress of von Mises in MVBF scenario looking that all cases, the trend of the curves are in according with the maximum values and so, also in this case they confirm the behaviour of the different kind of prostheses under MVBF loads.

Moreover, the values of 45° inclined forces are lower because it considers only the perpendicular component of the force while the parallel one does not because it would slide on the surface.

Some comments about the configurations, this project wants to give more attention to the normal chewing scenario because it is the principal activity during the days performed by the patient. Instead, the MVBF not only represent a pathological condition as an involuntary gesture (e.g. the nighttime bruxism) or a voluntary gesture, but also it has not the same frequency of the normal chewing cycles.

Therefore, it is important and convenient to explore all the possible circumstances that the prosthesis would address, but MVBF certainly has less weight from an experimental point of view with respect the normal chewing.

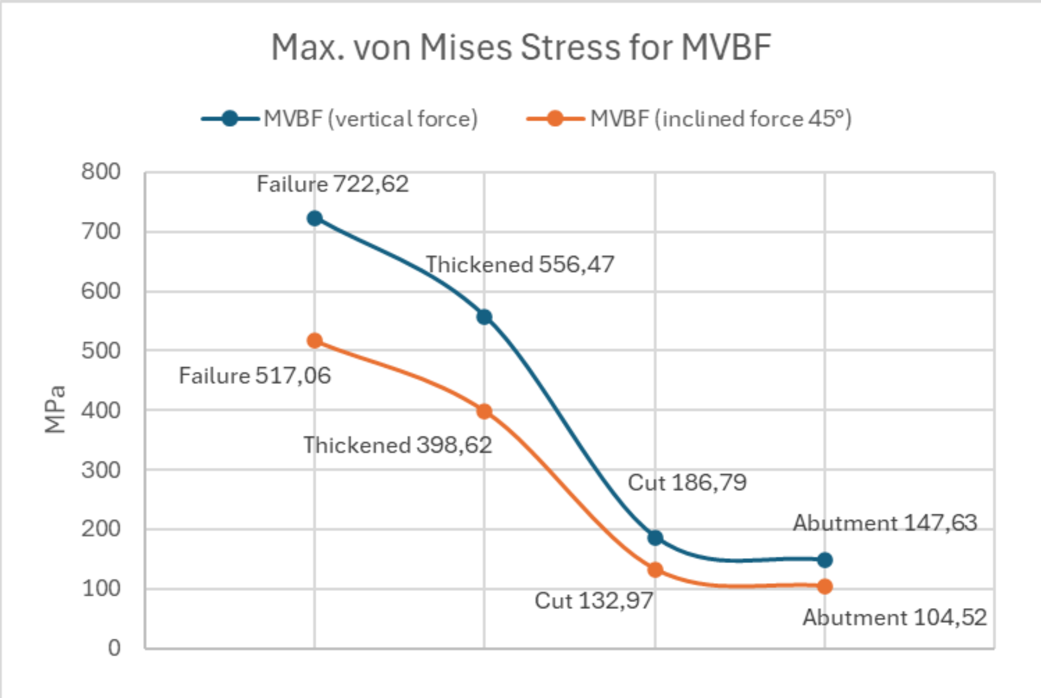


Figure 5.10 (a) Maximum von Mises stress for all cases in MVBF scenario

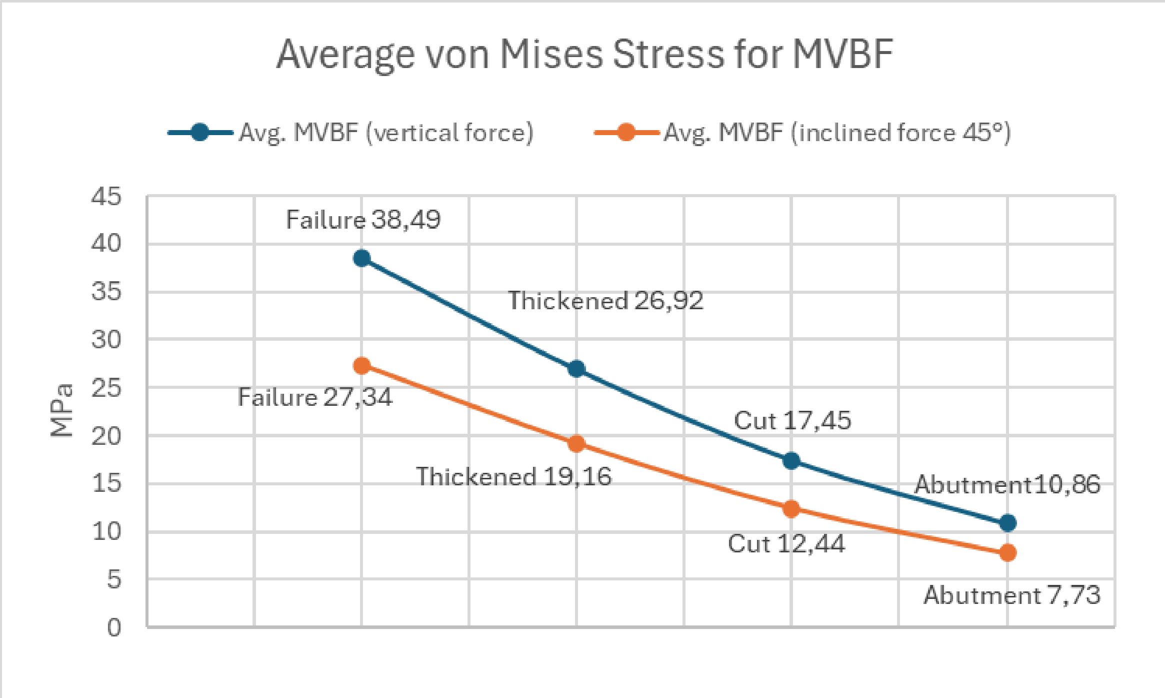


Figure 5.10 (b) Average von Mises stress for all cases in MVBF scenario

Some considerations on total deformation of the prosthesis in each different case. Deformation values (unit of measures in mm) were taken in the rupture areas, then at critical points. In Figure 5.11 (a) the graph of the deformation distribution in critical areas for all cases. Instead, on Figure 5.11 (b) the graph of the deformation in MVBF scenario. The points taken are only one for each T-Base channel, directly inside the break areas. Hence, the reference points are in that areas which are subject to breakage for the molar, pre-molar, and canine.

The graphs represent the total deformation for all simulations by considering only the vertical forces and not those inclined, because this interpretation of data can be enough to understand the trend of this quantity, among other things the trend with inclined forces would be the same.

It is possible to see that in normal chewing, the highest deformation values are present in the three T-Base channels of the failed prosthesis, which is coherent with the analysis of the rupture of the case of study.

The data decrease with the modified cases and for the cut prosthesis only a deformation on pre-molar occurs the others are null. Then, for prosthesis with abutment the deformations are noticed null for each tooth.

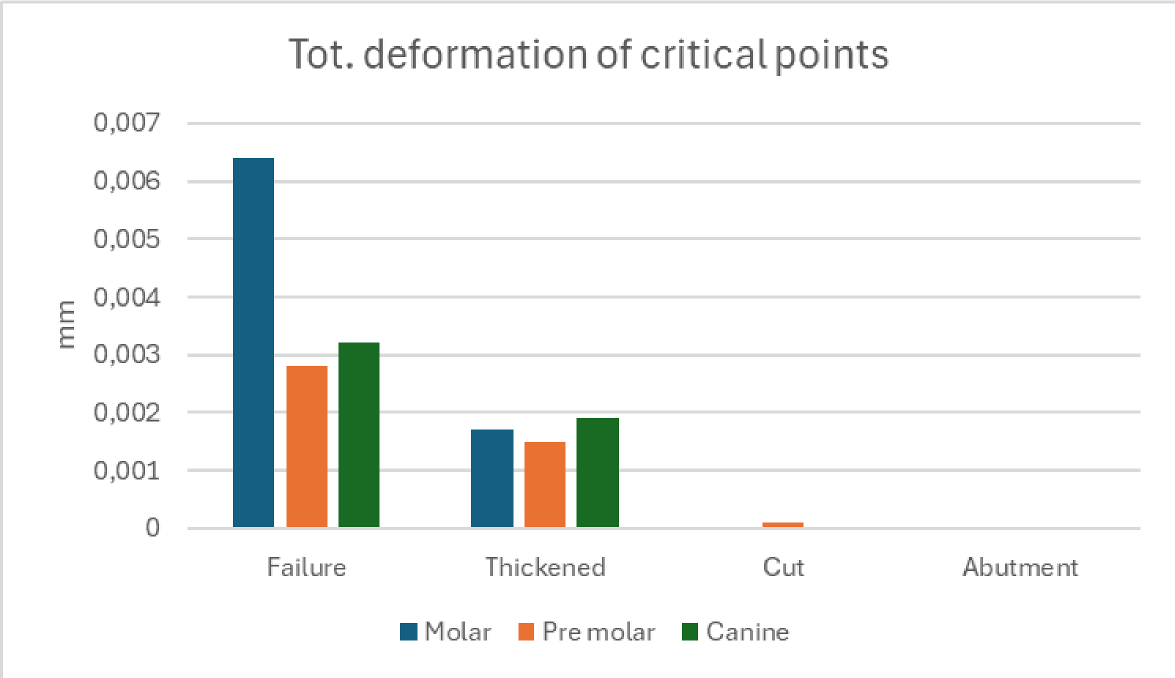


Figure 5.11 (a) Total deformation of critical points for all cases in normal chewing

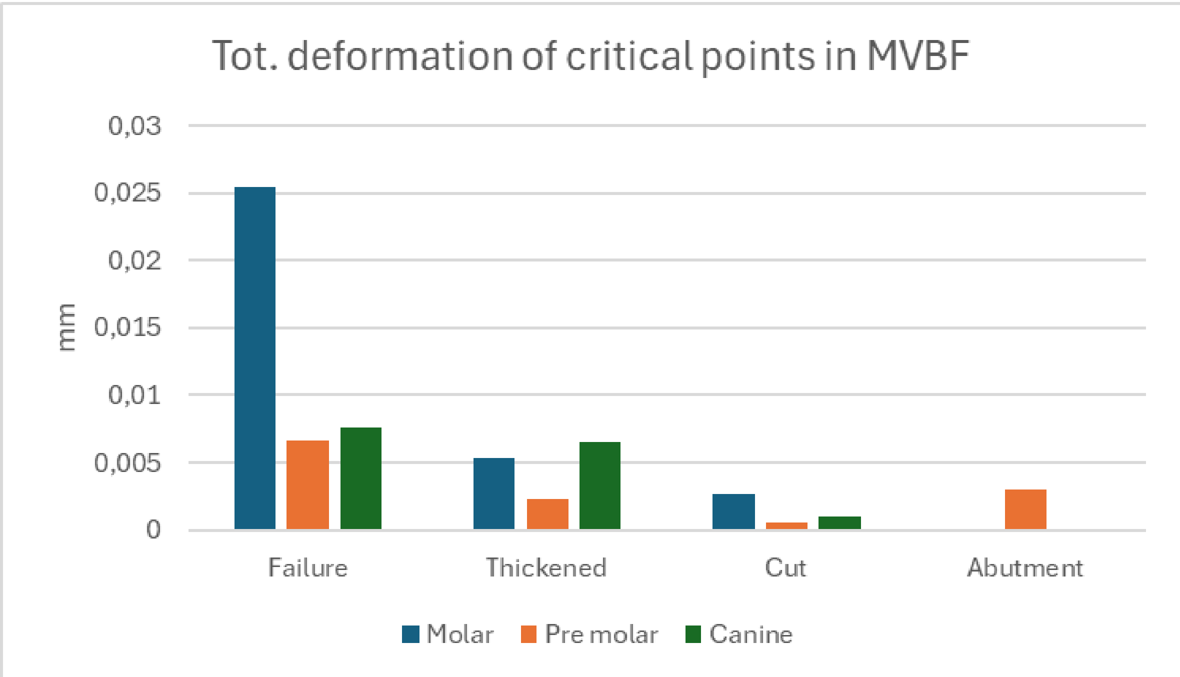


Figure 5.11 (b) Total deformation of critical points for all cases in MVBF scenario

Logically, in MVBF scenario, the deformation values are higher with respect the normal chewing and in all four cases the deformations are present. The failed, as before, shown the greater values in all three teeth, whereas the thickened are lower.

After, it is reassuring because from failed prosthesis the deformations drop until almost all null for the prosthesis with abutment. This last has higher deformation value with respect that simply levelled, speaking about the pre-molar, in fact the pre-molar canal has higher value with respect that of cut prosthesis. This may be because the prosthesis rests on CoCr abutment that is a material much harder than ZrO_2, and because it is the middle items among the others T-Base canals, where stress is conveyed more. Also, in normal chewing this phenomenon is negligible but with MVBF values could be an explanation.

5.2.1 Data analysis of the failed prosthesis

In this section it wants to comment the numerical data obtained on the simulation of the failed prosthesis.

Table 5.1 contains Equivalent stress of von Mises values of the failed prosthesis, protagonist of this thesis.

As already abundantly explained, the simulations were carried out in four phases, that are, with pressures in the normal chewing condition in which the forces were first perpendicular to the occlusal surface and then inclined forces of 45 degrees (taking only the perpendicular component). For consistency with the data, by regarding the MVBF scenario, the same pressure configurations.

Moreover, the stress data are taken in four areas for each T-Base channel of the prosthesis (i.e. molar, pre-molar and canine), that are the following: in front, sideways, behind and inside.

For normal chewing by considering both forces configurations, the highest stress values in the front parts occurs on the pre-molar, instead on the sideways the highest are found on the molar.

If the behind areas are considered the biggest values of stress is always on the pre-molar (where for vertical forces, there is the biggest value of the simulation).

On molar there are also the maximum stress values if the inside parts of canals are considered. On Figures 5.12 (a) – (b) the graphs of the data distribution of von Mises stress by considering each tooth area, and so each T-Base channel tooth. The figures are in normal chewing scenario, with pressures in a perpendicular forces configuration and inclined at 45°, respectively.

As regards the MVBF, the data are depicted on Figures 5.13 (a) – (b) in each pressure configuration, the positions of maximum value are almost the same of healthy condition. In the front the maximum stress values occur in the molar (for vertical forces) and pre-molar (for inclined forces). On the sideways and on the inside areas the molar contains the highest stress values, and for the behind parts, the pre-molar, which has the highest stresses of all simulations.

The maximum values of total deformation for the failed prosthesis occurs on the tip of the canine, but it is obvious that it is interesting see on breakage areas, so the maximum deformation on the T-Base channel is found on the molar for all four force configurations.

Since the prosthesis was worn for 438 days and undergoing at least 394.200 cycles under masticatory loads, it cannot rely on an initial fracture in a scenario where MVBF was the main configuration. Under this hypothesis, the handwork had to be of high quality or have defects with a critical length below 0,015 mm, considering the value of 722,62 MPa as maximum stress (as already seen in table 5.7), at least initially. Or it should also consider that the configuration with the MVBF never happened, it would be necessary to investigate if MVBF could occur in the patient.

Since von Mises's equivalent stress never reaches flexural strength, even in the MVBF scenario, the prosthesis would never have undergone ductile rupture; therefore, the thought of this project is aimed at sudden fatigue breakages due to the achievement or overcoming of critical length of any micro-manufacturing defects, after several masticatory cycles, mainly by considering the normal chewing setup.

Also, remember that MVBF condition is extreme, that is due to a medical condition such as bruxism, or as a voluntary gesture. It is however studied to analyze the structural integrity in a further scenario.

It is better to focus on normal chewing, because as already mentioned the fracture is fatigue type and not brittle for cleavage, so the crack had to be smaller than 0,321 mm. If it was greater than 0,321 mm the prosthesis, under normal chewing condition, would have broken instantly considering that with an applied stress of 157,35 MPa would have reached the K_{Ic}.

The prosthesis can tolerate defects up to about 0,321 mm in length without fracture in the immediate by considering the normal chewing. The stress values vary according to the position at different points of rupture and the force configurations, but taking the maximum value it is possible to have an estimate of the maximum critical length that would fracture the material as soon as the patient would perform a chewing.

However, the other stresses lower than the maximum one would have a critical crack length which would be excessively large of the normal crack and therefore it would have no sense to study in those points. But after the rupture of a piece of prostheses would follow in each critical point of the prosthesis an increase in tensile states.

Given these considerations, it can deduce what could have happened to the broken prosthesis. In Section 4.2, 'Rupture analysis' several break-up scenarios have been explained. Based on the FEA simulation this project can give a response to a possible break due to numerical data already abundantly illustrated.

It can be concluded that from an engineering point of view, the stresses for the broken prosthesis at some points reach some values, that by calculating the a, these last fall within the dimensional scale of critical lengths for ZrO_2, and thus in the stress range. So, it is explainable a fatigue fracture.

The rupture occurring over time suggests that the size of the crack was, just worn, surely less than the 0,321 mm and that as the masticatory cycles progressed, it would have reached the critical length for which equalling or exceeding the K_{Ic}, the prosthesis brakes irremediably, in several places.

For example, the maximum stress in the molar is within the canal and is 138,17 MPa, which means that the minimum length should be 0,416 mm, and it is slightly out of range of standard cracks that would cause concern.

Even in the canine, considering the maximum stress of 79,58 MPa, can be computed a crack of 1,256 mm as minimum size, which it would not be fractured. All this means that in the pre-molar and almost in the molar there are such tensions that by calculating the critical dimensions of the defects, these fall within the dimensional scale of a normal clique formed in the production of zirconia prostheses. Which is a real problem for the integrity of the medical device.

For completeness, critical lengths data are reported in Table 5.8 in the normal chewing and MVBF scenarios, only for vertical force pressures, to give an idea of the aforementioned dimensional quantity for the main case study of this thesis. These values were computed with the general formula of fracture toughness.

Logically, to implement a conservative design, it would also need tools aimed at detecting any defects present in the medical device during the post-production phase

Table 5.8 Critical crack length values for different areas of the failed prosthesis

	Molar (mm)	Pre-molar (mm)	Canine (mm)		Molar (mm)	Pre-molar (mm)	Canine (mm)
Normal chewing				MVBF			
Front	15,940	8,140	49,310	Front	0,610	0,731	3,440
Sideways	2,220	8,680	4,970	Sideways	0,170	0,542	0,500
Behind	1,170	0,321	2,870	Behind	0,090	0,020	0,130
Inside	0,416	4,310	1,250	Inside	0,030	0,200	0,210

From the Table 5.8 the minimum length in the molar is inside the T-Base channel as well as in the canine. In the pre-molar, as already discussed at length, is in the back area. The value is “border line” for the inside part of the molar, and not so worrisome in the canine. For MVBF they are located below the threshold inside and behind the molar and behind the pre-molar, in other positions, though slightly, they fall within the dimensional range of standard defects during the production of such medical devices in ZrO_2. In Figure 5.14 the trend of the critical crack lengths for normal chewing (left) and MVBF (right) described in Table 5.8, it can be seen as for the MVBF the values are minimal.

But it must be said that, when the fracture occurs at one point of the prosthesis, the entire safety is lost, creating greater efforts at the other weakest points. The fracture, as already mentioned, seems to be on the right side exclusively for the molar and the canine, instead for the pre-molar following a circular break but still poured more on the right.

The most widely accepted hypothesis could be that breaking in some part of greatest stress (presumably behind the pre-molar), created tension zones on the right side of each channel. These problems added to those described in previous paragraph have contributed to the failure of the prosthesis. The most credible hypothesis from a mechanic approach is of a sudden fatigue fracture, due to thousands of chewing cycles. A brittle or ductile fracture can be safely excluded, as the maximum recorded stress does not approach the breaking modulus.

It is important to remember that clinical malfunction hypotheses, previously reported, have helped to understand the problem and create alternative solutions. However, these need to be complemented by an engineering perspective, which means it is important to provide numerical data that offers an objective view of the structural situation.

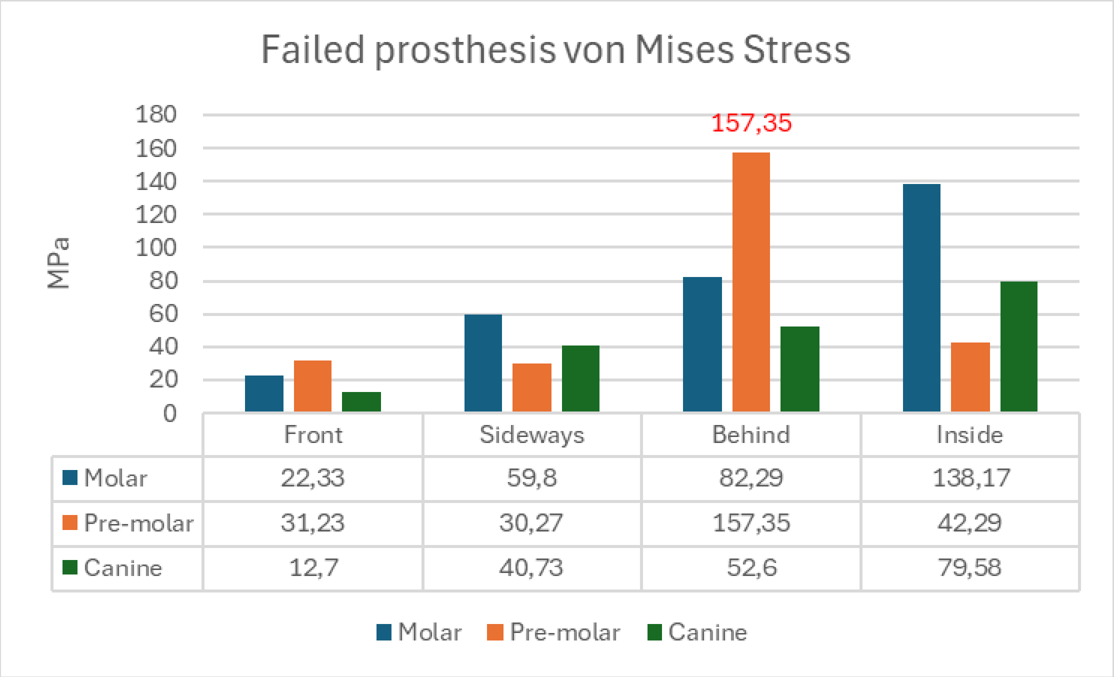


Figure 5.12 (a) Stress of von Mises of the failed prosthesis in normal chewing configuration

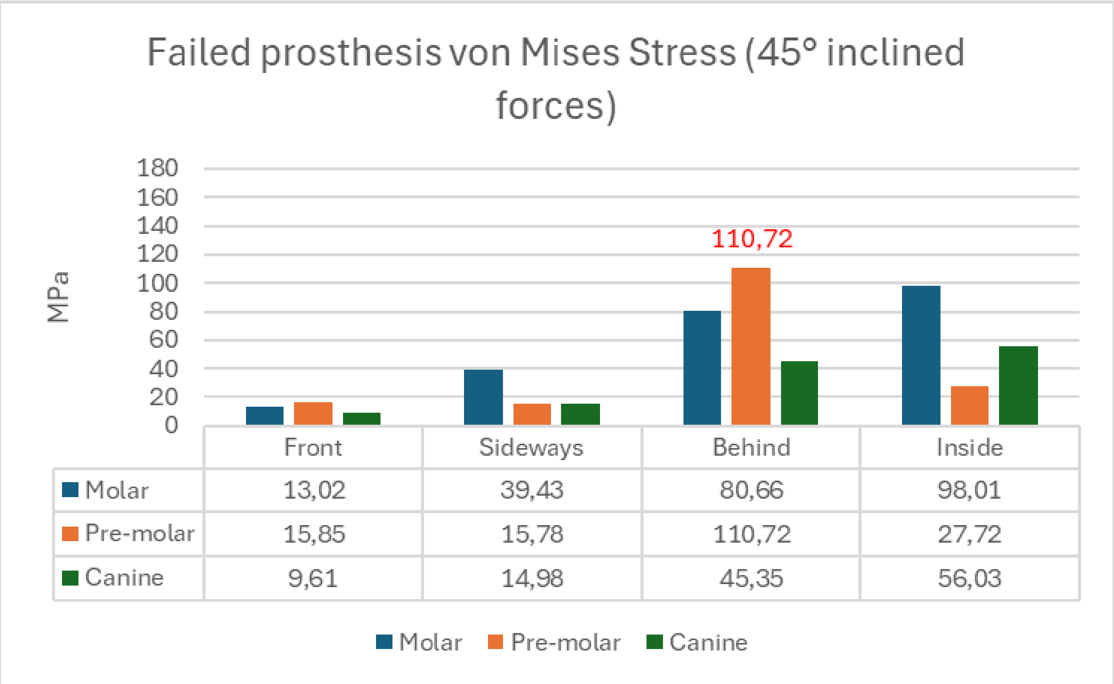
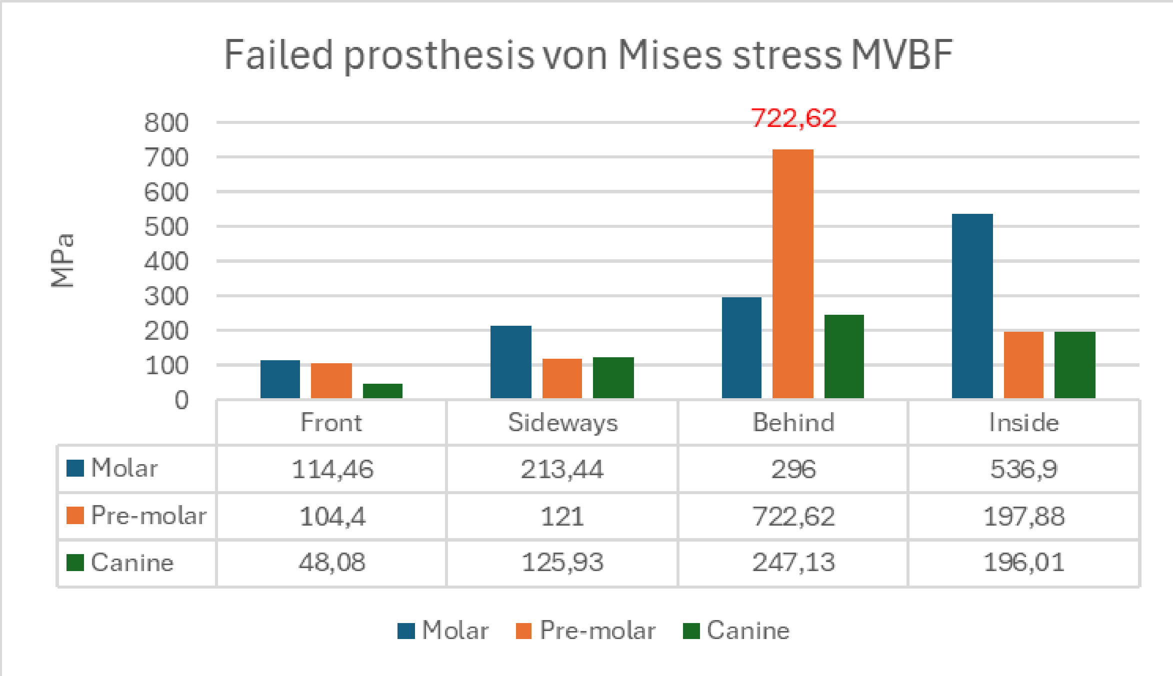
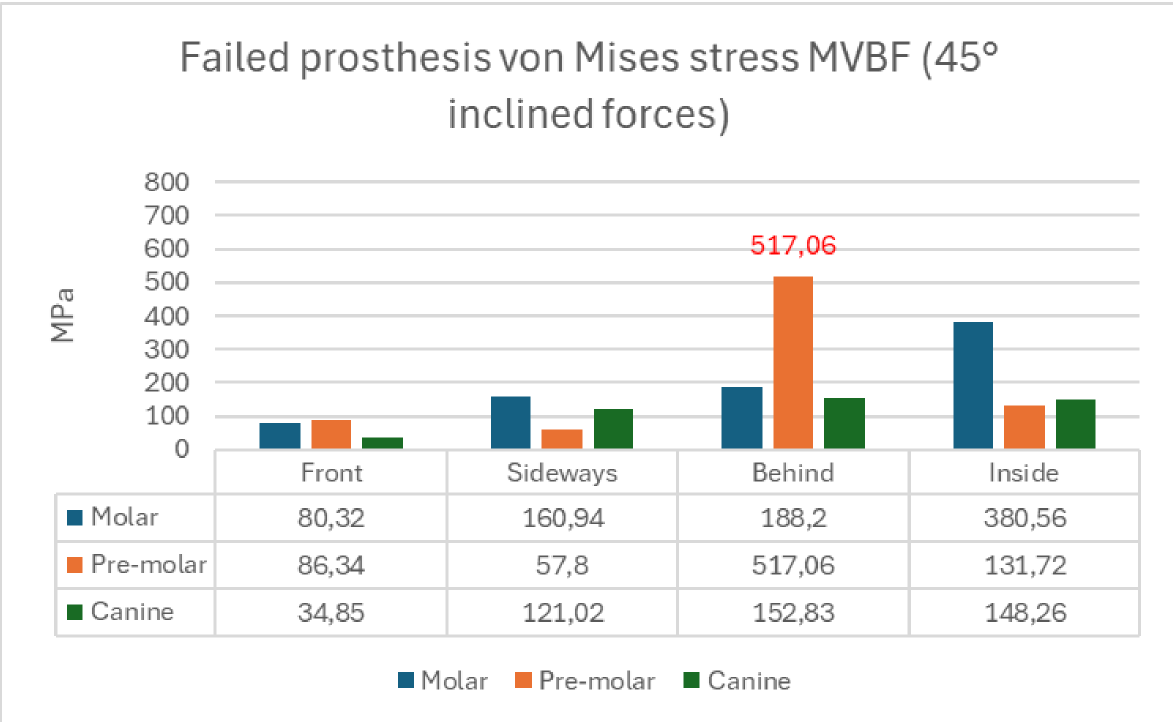


Figure 5.12 (b) Stress of von Mises of the failed prosthesis in normal chewing configuration (45° inclined forces)



	Front	Sideways	Behind	Inside
Molar	114,46	213,44	296	536,9
Pre-molar	104,4	121	722,62	197,88
Canine	48,08	125,93	247,13	196,01

Figure 5.13 (a) Stress of von Mises of the failed prosthesis in MVBF configuration



	Front	Sideways	Behind	Inside
Molar	80,32	160,94	188,2	380,56
Pre-molar	86,34	57,8	517,06	131,72
Canine	34,85	121,02	152,83	148,26

Figure 5.13 (b) Stress of von Mises of the failed prosthesis in MVBF configuration (45° inclined forces)

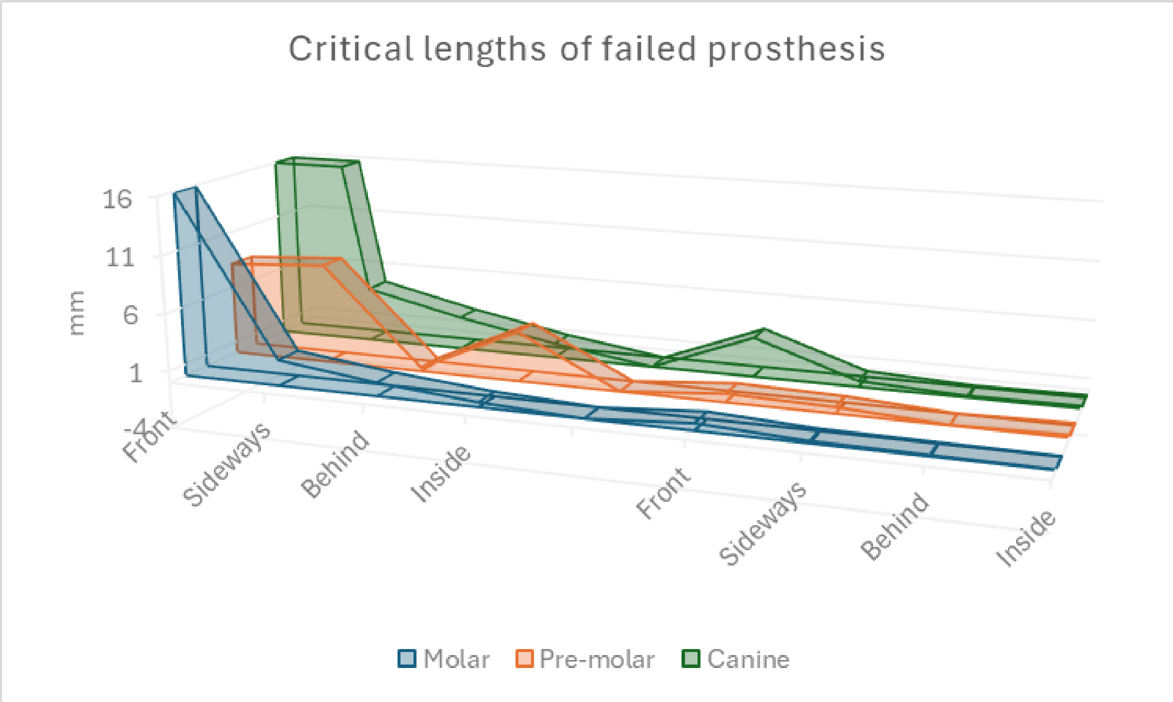


Figure 5.14 Critical crack lengths of failed prosthesis in normal chewing and MVBF configurations

5.2.2 Data analysis of the thickened prosthesis

As said, the first change was made by increasing the thickness of the T-Base channels to assume an increase in the safety of the device. In fact, as seen in Figure 5.15, the tensile values for normal chewing with vertical forces, which are all lower than the failed prosthesis, previously analyzed. On the left the failed prosthesis data while on the right the thickened prosthesis data.

As regards the normal mastication, maximum values are noted in the pre-molar for the front and rear parts. In the side part the highest value for molar and for the inner wall the canine (for vertical forces) and the molar (for inclined forces). It is already possible to see a concrete decrease in numbers.

In the MVBF configuration, the front and back areas have the maximum stress values in the pre-molar with a large numerical difference. In the lateral zone the canine has the maximum, and in the internal zone the canine (for the vertical forces) and the pre-molar (for the inclined forces) also here with great difference of values. The MVBF and normal chewing in vertical forces have the same position value except for the sideways which in MVBF is on canine. Instead, for inclined forces a discrepancy is always in lateral part (canine), and on inside wall (in pre-molar).

In this simulation, there is already a substantial difference in the stress data for the MVBF. Apart from the pre-molar which still has significant data, the others present a significant decrease, which is reassuring from a structural point of view and confirms the predictions for this case.

In addition, from a design point of view, the critical crack sizes before breaking rise to 0,564 mm (i.e. an increase of 75,70% than failed device) for the normal chewing, which brings this value out of range for standard zirconia defects; therefore, a break would occur more difficult. If the MVBF is considered the critical length is always low value of 0,025 mm.

The total deformation value remains maximum at the tip of the canine but in critical areas the highest value is in the canine for normal loads with vertical forces and in the molar for the inclined forces. Instead, for the MVBF in the canine for vertical forces and in the pre-molar for inclined forces.

It would be insignificant to illustrate a table with the values of a in the various areas of the prostheses, because they are higher than failed prosthesis, and it can rely only on the value related to the maximum stress. An estimate of the growth rate of the critical size of the cracks is about 75,70%.

From an economic point of view, such a solution could be advantageous for the customer, would ensure greater stability than failed prosthesis and an equal cost of production.

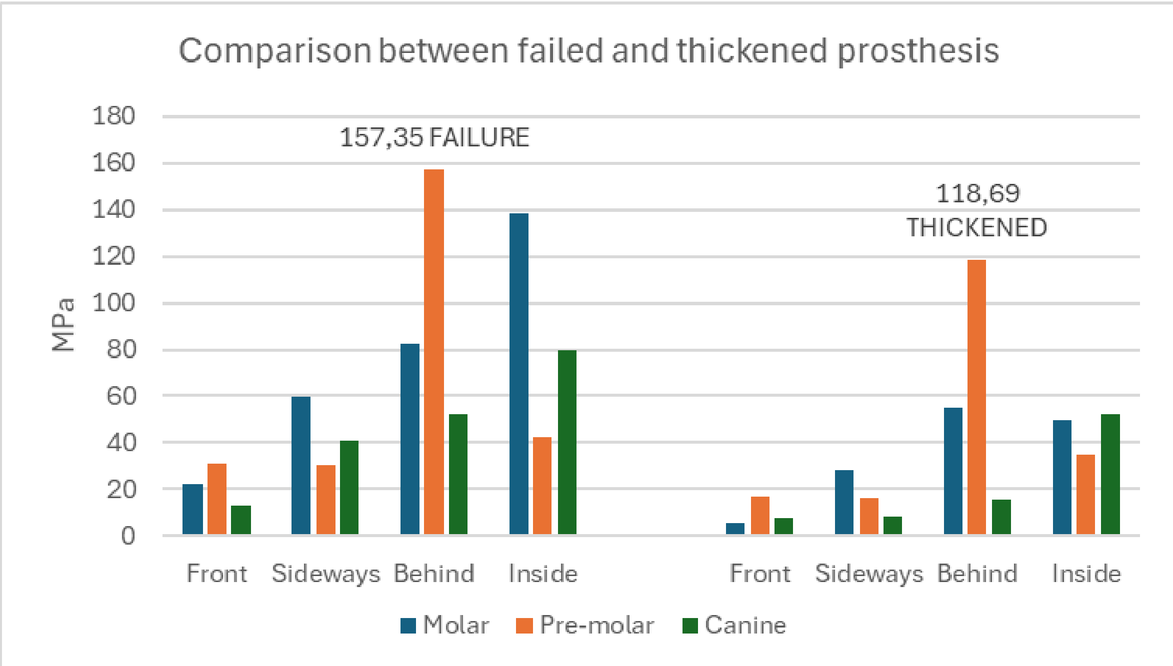


Figure 5.15 Comparison of stress data of the failed prosthesis with the thickened one

5.2.3 Data analysis of the cut prosthesis

The cut prosthesis was designed to improve the distribution of masticatory loads and reduce the points of excessive stress. By analyzing how the levelled prosthesis handles the pressures with respect to the thickened prosthesis and the broken one is of great importance for this project. The geometry of the cut prosthesis was modelled using advanced CAD software, but this solution should involve additional processing on crowns or implants to fill the gap created. Previously, the load conditions included vertical, then inclined forces at 45 degrees.

It is expected that the levelled prosthesis will show a decrease in stress more uniform than the thickened prosthesis, and therefore with significant reductions in stress concentrations at critical points.

FEA simulation results will provide valuable guidance for further improvements in prosthetic design for Yndetech.

Note that the greatest stress in the front is in canine, whereas sideways is in the molar canal. Instead, in the back and inside, they find in the canine canal, such considerations apply to normal chewing in both configuration of forces unless in the inclined force for inside part on the molar, so it is possible to see on Figure 5.16 the comparison between the three cases analyzed up to now, that is failed, thickened and cut prosthesis.

A gradual decrease in the stress distribution is observed, only slightly higher stress values are noted in the cut prosthesis than thickened one in the back of the canine in all pressure configurations (slightly lower of broken prosthesis). In normal chewing, the maximum values are also in canine for the front and inside part (for vertical forces), whereas in molar for the lateral area and into inner wall in molar T-Base canals (for inclined forces).

If the focus goes on the MVBF in both strength configurations, it has the same results for normal chewing in vertical forces and different for inclined forces in lateral areas which in this case the higher value is on canine. Among MVBF situations, the maximum numbers are in the same position except for the lateral and inside areas. In vertical forces are on molar and canine respectively, whereas for inclined forces on canine and molar respectively.

There is a general decrease in stress on critical areas of about 63,28% between the thickened prosthesis and cut one, and 72,32% compared to that broken. It is amazing how a simple change has such a structural impact.

This means that the critical crack length increases by a dimensional order in the standard range of zirconia manufacturing defects, to a minimum number of 4,18 mm, that is 641,84% compared to the thickened case.

Remember that with these stresses applied in this kind of prosthesis it would come to break overcoming a clique of this size, which is unthinkable such a defect in production, because surely it would not exceed the quality control.

This value undoubtedly brings security to the structural integrity of the dental prosthesis, proving that the assumptions of the geometric changes previously made were accurate.

Moreover, even with the MVBF the maximum stress applied is little higher than that of the broken prosthesis under normal chewing conditions. This means that it falls within the range of the minimum crack sizes produced during the processing of zirconia, which makes this prosthesis safe even in such extreme conditions, in fact the critical length of the crack in this scenario is of 0,228 mm (remember the range of zirconia defect on dental prosthesis [0,100 – 0,400] mm).

As for the total deformations, they are null for the molar and the canine and a minimum value in the pre-molar, regarding the normal chewing with the occlusal force entirely perpendicular. While in the inclined force configuration the deformations are all zero. With the MVBF total deformations are further reduced compared to the thickened case, so a percentage decrease of about 50% with respect the thickened case, and about 74,42% than failure case. Also in this case, it does not illustrate a table with the values of a in the various areas of the prosthesis, and an estimate of the growth rate of the critical size of the cracks is about 1201,56%. It can be concluded that the cut prosthesis is the mid-range device; advantageous from a structural point of view with contained costs of design but also higher with respect the thickened medical device.

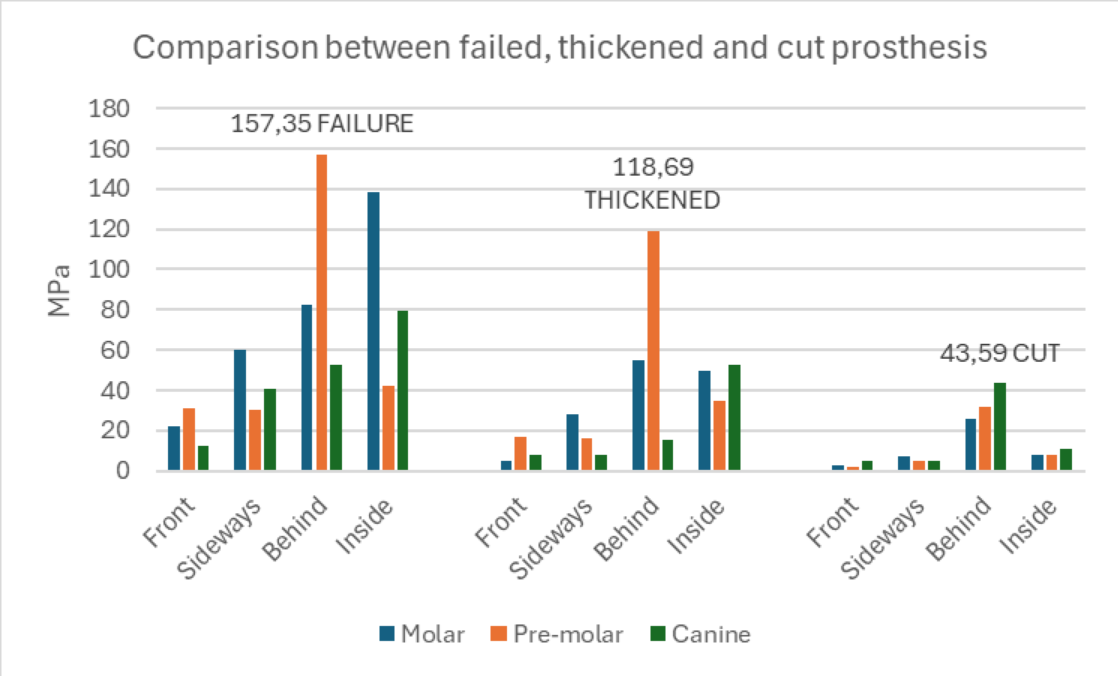


Figure 5.16 Comparison of stress data of the failed prosthesis with the thickened, and cut ones

5.2.4 Data analysis of the cut prosthesis with abutment

As the latest change for the geometric part and with the novelty of adding of some medical devices. This type of prosthesis is designed to improve the distribution of masticatory loads using intermediate supports, knows as abutments, that connect the prosthesis to the underlying dental implant structures.

The FEA simulation of this case showed promising results, by revealing significant improvements in mechanical performance and consequently in the service life of the prosthesis.

This study aims to assess the effectiveness of the intermediate abutment prosthesis in distributing the masticatory loads more evenly than traditional prostheses. Then to identify the structural and functional benefits offered by the intermediate abutments, with particular attention to the reduction of stress concentrations and the increase of the stability of the prosthesis.

The abutments are made in CoCr and would be inserted into the channels of the T-Base, following the exact geometry inside them in such way to put the correct conditions at the boundary, as illustrated in the pre-processing section.

As in all cases, in order to have useful data to compare the simulated load conditions, the configurations are always the same, by repeating, pressures with vertical forces and inclined ones at 45 degrees, reproducing realistic scenarios of mastication, in a healthy and extreme chewings. On Figure 5.17 the comparison between all prosthesis in normal chewing scenario by considering the pressures with only perpendicular forces.

The highest stress values occur in the canine for the front, the inner wall and the posterior part, while for the lateral part the molar has the highest value. In this case, the contact surface between the prosthesis and the abutment, called 'interface', was also studied, in which the maximum value is for the molar, for normal chewing at vertical and inclined forces. For tilted forces, the highest stresses are in the same place unless for the inside part which is in molar and not on canine. In the interface area of the medical device, it is possible to see quite high stress values, a plausible explanation could be the interaction of the ZrO_2 prosthesis with the abutments in CoCr, which of course is a much harder material (depending on the alloy that is chosen it has a fracture toughness ranging from 632,4 to 1581 MPa$\sqrt{mm}$) and that could create greater stress under load. [40]

For the inclined forces the greater values are found in the same points except for the inner part where it is in the molar.

Talking about the MVBF condition, the values have a very small gap compared to those of normal chewing, unlike other simulations. The positions of the maximum stresses are in the same points as normal chewing except for the inside part which is, in vertical forces, in canine for normal chewing and for MVBF in molar. Instead, for inclined forces the difference occurs in sideways, where it is in the canine for MVBF and the molar for the healthy chewing. Among the MVBF configurations the values are in the same position except the lateral part, one in molar and the other in canine.

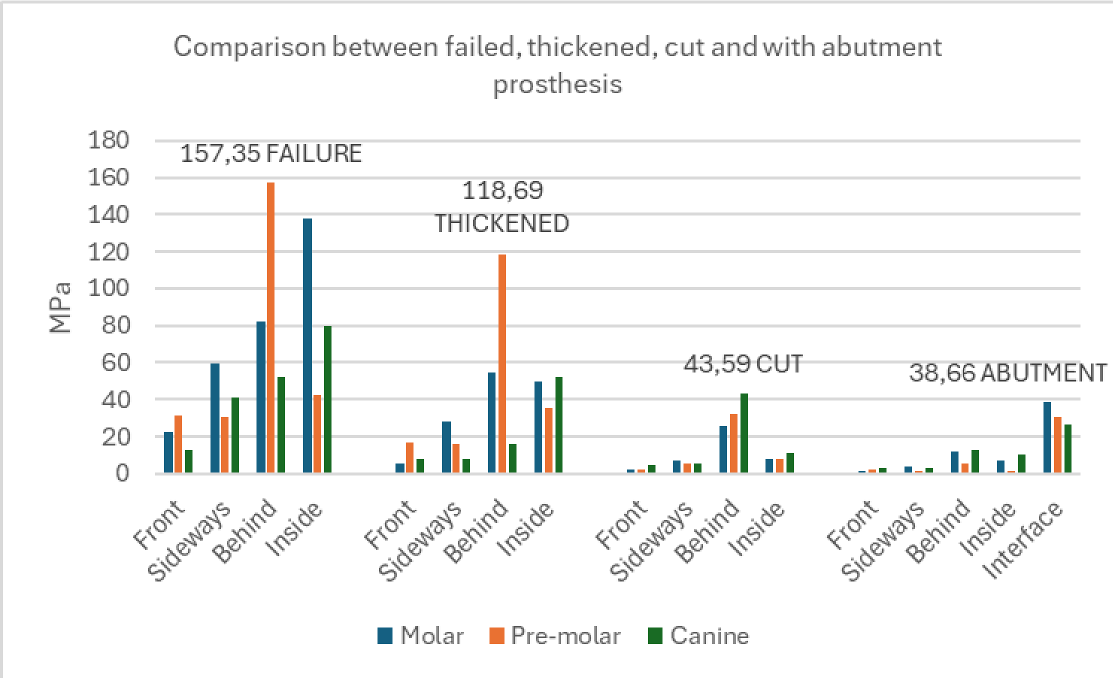


Figure 5.17 Comparison of stress data of all simulations

As can see from the graph, overall values are much lower than the simple prosthesis cut, except for the pre-molar in the anterior area and onto interface areas. The simulation showed comforting and expected results, which showed a more balanced distribution of forces, and a reduction in stress concentrations at critical points. Moreover, in this FEA simulation the total deformation was finalized. The results are clear, for the normal chewing the deformations are zero in the areas of rupture in all scenarios; but also considering the MVBF there are zero deformations for the molar and the canine and a small value for the pre-molar. It is a surprising result, that even with MVBF there are zero values, it is clearly important from the structural point of view and the reliability of the prosthesis. Data indicate that with the use of intermediate abutments not only improves the stability of the prosthesis but also increases its longevity, reducing the risk of fractures and other forms of structural damage. An estimate of the growth rate of the critical size of the cracks is about 1558,57%, in fact it is 5,32 mm for normal chewing. Surprisingly, in MVBF the value of the maximum stress is lower, and so the critical crack length higher than those of failed prosthesis (i.e. 147,63 MPa and an a of 0,365 mm).

Finally, the prosthesis with abutment shows greatest improvements in performance, it works wonderfully in extreme condition, it is suitable also for people with bruxism, and the use of abutment augments its stability and longevity. Clearly, it is the best solution but also the most expensive.

5.2.5 Data analysis of the validation prosthesis

A crucial step in this process is the validation of analytical methods, to achieve this goal, a specially designed modified prosthesis has been developed to validate and optimize the FEA method on that project, without the necessary time to test on other failed models.

This new prosthesis incorporates structural changes (as said on section 4.6) aimed at improving the accuracy of FEA simulations and ensuring that the results obtained are representative of real conditions. [41]

The modified prosthesis has been designed with specific structural alterations that facilitate FEA analysis. The FEA simulation of the modified validation prosthesis aims to verify the accuracy and reliability of the FEA method in predicting the mechanical performance of dental prostheses, which have discussed before.

The mode of the simulation must be the same as for the other tests. Starting from the implementation of material, boundary conditions, the masticatory loads and finally studying the same physical quantities, that are 'Equivalent stress of von Mises' and 'Total deformation'.

Starting from von Mises's stress for normal chewing and MVBF it is possible to note that, considering the prosthesis in which the total area of the element connecting the two crowns has been severely reduced (i.e. 2 mm^2), the stress remains low.

On the other hand, in the case of the undersized palatal area, can be seen almost high values. For example, under masticatory load in normal chewing condition, the maximum stress obtained, with completely vertical force at the occlusal zone, is of 167,06 MPa, which means having a limit length of crack of 0,284 mm. For the MVBF, a maximum of 259,06 MPa that corresponds to 0,118 mm. The average stress for the healthy chewing and MVBF in both pressure configurations shows a trend that goes hand in hand with maximum stress.

The stress values are similar those of the failed prosthesis that is the case study of the thesis, for normal chewing of the broken prosthesis, the values are similar (i.e. 157,35 MPa and 0,321 mm) and bearing in mind that the validation prosthesis would lead to structural damage, it can be concluded that the broken prosthesis is a "border line" case. However higher values are observed for the extreme changes made on this prosthesis, as expected. Comparing the results of the FEA simulations of this modified prosthesis with the experimental data obtained, ascertain that the stress values obtained are higher than those of the failed prosthesis, it gives us positive feedback on the robustness of the FEA simulation. It is useless to talk about other configurations with inclined forces since they give lower stress values.

Some last considerations on total deformations, the comments may be analogue to stress, that are, higher values are found in the case of the prosthesis in which the thickness of the palatal area has been changed, while lower deformations data regarding the simulation performed in the case of area undersized in the middle of the crowns.

It is interesting to observe that the deformation values taken in critical areas of palatal surfaces modified are increased by about 10,11% (Figure 5.18), with respect the maximum value of deformation of the failed prosthesis, by considering the configuration in normal chewing at pressures with perpendicular forces.

This increase in deformation may be an indication, it may specify unusual elastic behaviour of ZrO_2, which may mean a higher probability of breakage. Thus, an augment in the risk of rupture, as wanted to demonstrate with these simulations, as well as consistency with stress distributions. Indeed, such a prosthesis would suffer a rupture much more easily than any other prosthesis, in fact this kind of crowns would never pass quality control and a change would be necessary.

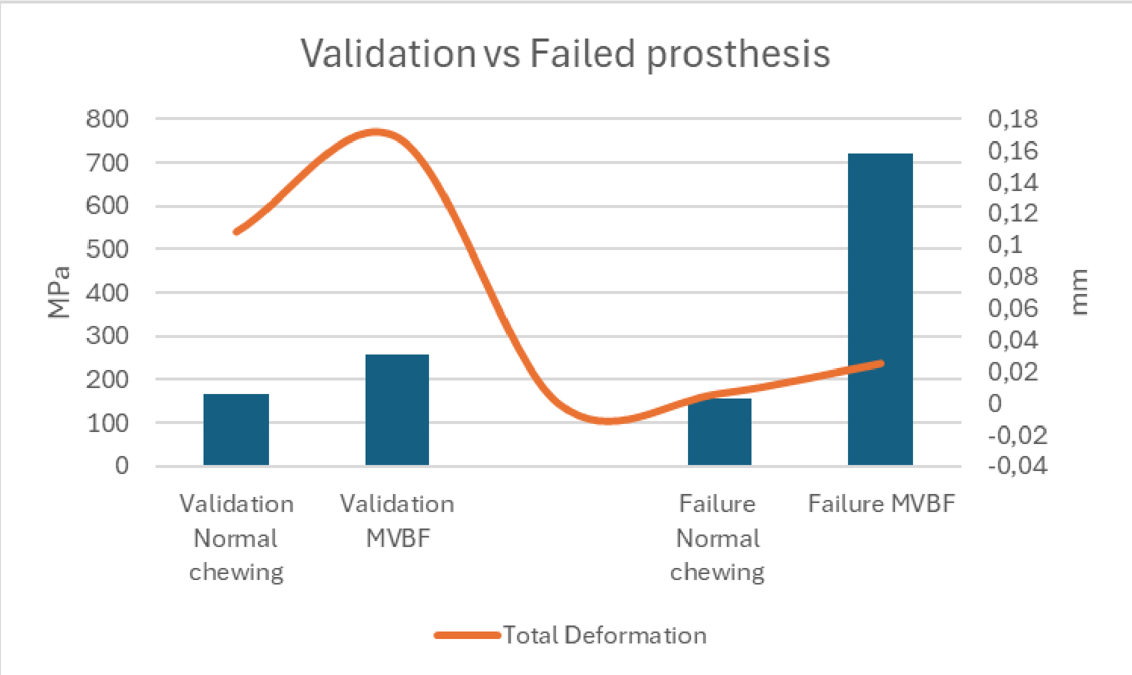


Figure 5.18 Comparison between maximum stress and Tot. deformation of Validation and failed prosthesis

It is appreciated in Figure 5.18, a comparison of stresses between the validation prosthesis in the palatal area on the left, creating a thin layer of zirconia, and the previously studied broken prosthesis on the right. Noting similar stresses except for the MVBF that in the broken prosthesis is much greater. The orange line represents the trend of total deformations, it can be viewed totally different deformations data, as said before an increase of 10,11% with respect the failed prosthesis.

This indicates an almost certain rupture of the validation prosthesis because a high degree of deformation by considering similar or worse lower stress values compared to the broken prosthesis data, assumes a great flexibility of the material while not being a characteristic of zirconium dioxide, that means a certain breakage.

Whereas low strain values for higher stress, in the case of the failed prosthesis with MVBF, indicates a resilience of the material before breaking.

This transversal project of the validation method confirms three aspects. The first, the assumption made for the failed prosthesis by considering the stress data obtained, which is, the prosthesis has arrived at a fatigue break following approximately intense masticatory loads, is correct (even if it does not reach the value of flexural strength).

In fact, the stress values of the broken prosthesis are slightly lower than these, and since it was assumed mechanical properties completely equal to the previously simulated prostheses (i.e. $K_{Ic} \leq 158$ MPa$\sqrt{mm}$ and the flexural strength = 1000 MPa), similar values are found, remembering that this prosthesis would probably break at the beginning, while the failed prosthesis last over time. It notes that in this case it has in normal chewing 167,06 MPa that exceeds 158 MPa, which turned out to be risky.

Also, critical lengths fall within the range of dimensional defects of zirconia prostheses (i.e. [0,100 – 0,400] mm), what it wants to highlight is the similarity in the stress and critical crack length data obtained, however greater or worst in the validation test. Thus, it is also very important detect further defects on the prostheses.

The second is that during the simulations of dental prostheses, although it seemed not very useful, the numerical data of the deformations are significant, they are indicators of structural failure, provided that minimum values must be observed. In fact, even if we have similar stress values in test validation, the values of the deformations change totally indicating a flexibility of a hard material such as ZrO_2 unlikely.

The third is that FEA simulation is a safe and reliable method to study and analyze the structural behaviour of a dental prosthesis.

Hence, structural changes are expected to improve the accuracy of the results obtained in the other tests. Indeed, the validation prosthesis has been a good intuition that leads this project to confirm the hypotheses thought on the other simulations, and since this prosthesis will break almost certainly having dimensional thresholds below the standards, even the deformations result of the quantities to be considered. Furthermore, structural changes are expected to improve the accuracy of the results obtained in the other tests. [42]

The validation test of two-element crown, which is a prosthesis cemented on a natural stump, was very useful from an engineering point of view, saving time on further simulations of broken prostheses.

Moreover, since the stress and deformations data are similar to the broken prosthesis of case of study, it means this last is a "border line" case.

6 Conclusion

The FEA performed on a broken prosthesis, followed by several simulations on modified devices, provided valuable results for the improvement and validation of dental prosthesis design techniques.

The simulations showed significant increases in the mechanical performance of the modified prostheses, confirming the robustness and effectiveness of the FEA method used. [43]

FEA simulations have demonstrated a remarkable ability to accurately predict the mechanical behaviour of this kind of medical devices, to numerically examine stresses and deformations potentially dangerous for structural integrity. Moreover, FEA become a good tool for a static structural analysis either pre-production or to understand the reasons after the fracture event.

It can be concluded that the dental prosthesis related to the case study has probably broken due to several clinical causes, first the mechanical cementation has contributed to a relative movement of three elements (i.e. prosthesis, T-Base and resin), secondly the T-Base channels were not all at the same height, which led to an imbalance of loads.

In addition, the thickness of the T-Base housing was too thin, which leads to the study of stresses in these critical areas by highlighting high stress values for ZrO_2 prosthesis, and by taking account of the $K_{Ic} \leq 158$ MPa$\sqrt{mm}$ and flexural strength = 1000 MPa. The rupture could not be by cleavage but by fatigue, due to thousands chewing cycles which has led to reach the critical length of the micro-structural defects potentially present in the channels, leading to the inevitable rupture of the prosthesis. It is essential to know the fractural strength and especially the fracture toughness, which were proven to be the most important quantities to consider. Also, the critical size of defects (a) was proved to be a key parameter for the mechanical investigation in this project. Hence, the clinical aspects were confirmed and enriched also from an engineering point of view by analyzing stress and deformation data.

The idea of structural and geometry changes that have led to performance improvements could be implemented in future prosthetic designs, so that improving their strength and durability.

Furthermore, confirmation of the robustness of the FEA method allows designers to use this tool with greater confidence, knowing that it can accurately predict the performance of prostheses under real loads.

Future directions for Yndetech could include the execution of further experimental studies to continue validating and refining the FEA model, ensuring that it remains a reliable method for prosthetic design. Then, the engineer must be able to translate the observed improvements in FEA simulations into clinical applications by testing modified prostheses.

First, Yndetech could develop an 'ad hoc' CAE software by applying FEM and implementing it on Hypsocad or create the program from new to study their products. It would be useful to develop a FEA software for the dental prosthetics and implantology fields with some tools supplied to the company as for example the optical scanners or external medical imaging devices as CT or MRI of dental studies or hospitals. It could only perform static structural analysis and possibly the stress-life study of dental components subjected to small, medium and maximum intensity masticatory loads over time.

Therefore, as a consequence the FEM must enter in clinical and industrial practice of Yndetech's activities or dental studies that collaborate with Yndetech, particularly in the field of implantology and dental prosthetics.

From a point of view regarding the mechanical properties of materials, if it is known the K_{Ic} and the flexural strength of the ZrO_2 or the prosthetic material used, the engineers could always opt for a conservative design of the medical device. In the absence of crack size data, high safety factors could be used to ensure that the structure can withstand loads well below the theoretical K_{Ic} limit.

Moreover, in conjunction with the FEA could be useful regular inspections to identify and measure cracks in the material. One hypothesis could be the introduction of non-destructive techniques such as radiography, ultrasonography or magnetic tests can be used for this purpose. Hence, test material samples to determine the presence and size of cracks. So that the engineer could apply an appropriate safety factor in the design to consider uncertainties regarding cracks and other defects.

To conclude, Yndetech, through the FEA, has a detailed view of internal stresses and deformations, under mouldable masticatory loads. It is a powerful tool for understanding the causes of breakage or detecting any critical points in pre-production phase, so that developing solutions to improve the design. In conclusion, FEA simulation has proven to be a powerful and reliable tool for the design and optimization of dentures.

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