Optimization of Removable Partial Dentures Using Digital Technologies

By Omar Alageel

Supervisor: Dr. Faleh Tamimi

Faculty of Dentistry

McGill University

Montreal, Quebec, Canada

April 2019

A thesis submitted to McGill University in partial fulfillment of the requirements for the degree of Doctor of Philosophy in Craniofacial Health Sciences

© Omar Alageel 2019

Dedication

I dedicate my thesis to my parents, brothers, and sisters for their endless encouragement and support throughout the course of this thesis. Also, I dedicate this thesis to my supportive wife, Faten, and to my sons, Faisal, Yousef, and Mohammed, for all the wonderful things they bring to my life.

Acknowledgment

First and above all, I praise to Allah Almighty for providing me with this opportunity and for granting me health, courage, and strength throughout life and my Ph.D. studies.

I would like to express my most sincere appreciation, gratitude, and respect to my supervisor, Dr. Faleh Tamimi, for guiding and supporting me over the years. I am grateful for his patience, encouragement, guidance and immense knowledge. The achievements during my studies would not have been possible without his support and feedback. I am very fortunate to have Dr. Tamimi as my supervisor who helped me form the initial to the final stage of this project, I can never thank him for what he has done for me.

I would also like to thank my thesis committee members Dr. Marta Cerruti, Dr. Jun Song, and Dr. Jean-Marc Retrouvey for their guidance, advice, and feedback.

I would like to thank my lab-colleagues and all the collaborators who helped me along the way. I would like to acknowledge Dr. Mohamed-Nur Abdallah for his help with this project especially in conducting the cell culture studies of this project. Thanks to my office neighbor (Dr. Alaa Mansour) for the general help, Ammar Alsheghri for the help with the mechanical analysis, Dr. Ahmed Al-Subaie for the help with the micro-CT test, Suliman Algezani for the help with the testing samples, Dr. Nida Ashraf for the help taking the satisfaction questionnaires to form the patients. I would like to acknowledge Dr. Eric Caron for helping me designing and processing the experimental samples for this project. Also, I would like to thank Dr. Emmanuel Nicolas and Dr. Marion Bessadet from Université Clermont Auvergne. Thanks to Dr. Jeffrey Myers and Ms. Linda Harrison for their technical support.

I am most grateful to my family: my wife, my parents, my brothers and sisters as well as my friends for their constant and unconditional support. Their encouragement helped me overcome any challenges and stay focused on my studies.

Finally, I would like to thank King Saud University (Riyadh, Saudi Arabia), Faculty of Dentistry of McGill University, 3DRPD Inc. (Montreal, Canada), and the Natural Sciences and Engineering Research Council (NSERC) Collaborative Research Development and Canada-Discovery grant for their financial support throughout this project.

Table of Contents

List of Tables

List of Figures

Abstract

Removable partial dentures (RPDs) provide cost-effective treatment for millions of partially edentulous patients worldwide. However, RPDs often present several complications that result in treatment failure and patient dissatisfaction. The main reasons for these problems are the inadequate quality of the RPD framework and the lack of precise guidelines for designing retentive RPDs. We hypothesized that laser sintering/melting technology and advanced algorithms can improve the quality and design of removable partial denture treatments. Accordingly, the aim of this thesis is to optimize the quality and the design of removable partial dentures by using digital technology.

Removable partial dentures (RPDs) are traditionally made using the casting technique, but recently, new additive manufacturing processes based on laser sintering/melting have been developed for the quick fabrication of RPDs metal frameworks at low cost. The objective of the first part of this thesis was to characterize and understand the mechanical, physical, and biocompatibility properties of RPD cobalt-chromium (Co-Cr) alloys produced by two laser-sintering/melting systems and compare them to those prepared using traditional casting methods. In this study, it was found that both laser-sintered/melted Co-Cr alloys fabricated by 2 different laser sintering/melting systems were more precise and showed higher hardness, yield strength, and fatigue resistance than the alloys prepared by the casting technique. This was due to the smaller grain size and higher microstructural homogeneity of the laser-sintered/melted alloys compare to the cast ones. In addition, both laser-sintered/melted and cast alloys had similar biocompatibility.

Designing a retentive RPD is very challenging due to the lack of precise guidelines for designing RPDs, and there are no guidelines available to determine the optimal RPD design for each form of partial edentulism. Thus, the objective of the second part of the thesis was to

determine the forces produced by food and clasps during mastication to develop an algorithm for predicting RPD retention and to help determine the optimal number of clasps. First, the forces that food exerts on acrylic resin teeth during simulated mastication and the retention forces provided by different clasps type (wrought wire, circumferential, and I-bar) were measured. The measured masticatory forces varied according to the type of tooth, occlusion, and food while the measured retentive forces varied according to the type of tooth and clasps. Using these measurements, an algorithm for predicting RPD retention and determining the optimal number of clasps was developed and validated experimentally with a sensitivity of 96%, a specificity of 100%, and an accuracy of 97%.

In order to clinically validate the new model for predicting the retention of RPDs and to improve the guidelines for RPD design, the objective of the third part of the thesis was to investigate factors related to the retention of RPDs that affect patient satisfaction. In a population of 75 patients treated with 107 RPDs, 67% of them were satisfied with their RPDs. Patients were more satisfied with RPDs in the maxillary arch, tooth-bounded, or retained by >2 clasps than with RPDs in mandibular arch, free-end saddle, or retained by ≤2 clasps. Patients were significantly more satisfied with RPD designs that were predicted by our algorithm to have sufficient retention. The mathematical model for predicting the RPDs retention showed a clinical specificity of 83% in predicting patient satisfaction that may help design better RPDs with more predictable treatment outcome.

In conclusion, digital technologies such as laser sintering/melting technology and advanced algorithms can improve the design, accuracy, and clinical performance of removable partial denture treatments.

Résumé

Les prothèses partielles amovibles (PPA) constituent un traitement rentable pour des millions de patients partiellement édentés dans le monde. Cependant, les PPA présentent souvent plusieurs complications qui entraînent un échec du traitement et l'insatisfaction du patient. Les principales raisons de ces problèmes sont la qualité inadéquate du cadre de la PPA et l'absence de lignes directrices précises pour la conception de PPA rémanentes. L'objectif de cette thèse est d'optimiser la qualité et la conception des prothèses partielles amovibles en utilisant la technologie numérique.

Les prothèses partielles amovibles (PPA) sont traditionnellement fabriquées à l'aide de la technique de coulée, mais récemment, de nouveaux procédés de fabrication additive basés sur le frittage au laser ont été développés pour la fabrication rapide de structures métalliques à faible coût. L'objectif de la première partie de cette thèse était de caractériser et de comprendre les propriétés mécaniques, physiques et de biocompatibilité des alliages PPA de chrome-cobalt produits par deux systèmes de frittage laser et de les comparer à ceux préparés à l'aide de méthodes de coulée traditionnelles. Dans cette étude, il a été constaté que les deux alliages de chrome-cobalt frittés au laser fabriqués par 2 systèmes de frittage au laser différents étaient plus précis et montraient une dureté, une limite d'élasticité et une résistance à fatigue supérieures à celles des alliages préparés par la technique de coulée. Cela était dû à la plus petite taille de grain et à la plus grande homogénéité microstructure des alliages frittés au laser par rapport aux alliages coulés. En outre, les alliages frittés au laser et coulés avaient une biocompatibilité similaire.

Concevoir une PPA rémanente est très difficile en raison de l'absence de directives précises pour la conception des PPA. De plus, il n'existe aucune directive permettant de déterminer la conception optimale de la PPA pour chaque forme d'édentement partiel. Ainsi, l'objectif de la

deuxième partie de la thèse était de déterminer les forces produites par les aliments et les agrafes lors de la mastication afin de développer un algorithme permettant de prédire la rétention de la PPR et d'aider à déterminer le nombre optimal d'agrafes.

Premièrement, les forces que les aliments exercent sur les dents en résine acrylique lors de la mastication simulée et les forces de rétention fournies par différents types de crochet (fil forgé, circonférentiel et barre en I) ont été mesurées. Les forces de mastication mesurées variaient selon le type de dent, l'occlusion et la nourriture, tandis que les forces de rétention mesurées variaient selon le type de dent et de fermetures. À l'aide de ces mesures, un algorithme permettant de prédire la rétention de la PPA et de déterminer le nombre optimal de crochets a été développé et validé expérimentalement.

L’objectif de la troisième partie de la thèse était d'examiner les facteurs liés à la rétention des PPA qui affectent la satisfaction du patient. Dans une population de 75 patients traités avec 107 PPA. Les patients étaient plus satisfaits des PPA de l'arcade maxillaire, des dents attachées ou retenus par> 2 crochets que des PPA de l'arc mandibulaire, de la selle libre ou de ≤2 crochets. Les patients étaient significativement plus satisfaits des concepts de RPD prévus par notre algorithme pour une rétention suffisante. Le modèle mathématique permettant de prédire la rétention des PPA a montré une spécificité clinique de 83% dans la prédiction de la satisfaction des patients, ce qui pourrait aider à concevoir de meilleures PPA avec des résultats de traitement plus prévisibles.

En conclusion, les technologies numériques telles que la technologie de frittage laser et des algorithmes avancés peuvent améliorer la conception, la précision et les performances cliniques des traitements de prothèses partielles amovibles.

Contribution of Authors and Statement of Originality

This thesis consists of one book chapter and three manuscripts in the publication prepared by the candidate as the first author. A summary of the work originality and a statement of the involvement of the candidate and the co-authors are described below for each of these manuscripts.

Chapter 3 (Book Chapter):

Title: Fabrication of dental restorations using digital technologies: techniques and materials.

Authors: Omar Alageel, Berge Wazirian, Balqees Almufleh, Faleh Tamimi.

A chapter of the book "Digital Restorative Dentistry" that is submitted, and it will be published by Springer Nature.

Contribution: O Alageel wrote the additive manufacturing part as well as reviewed and edited the subtractive manufacturing part of the book chapter. B Wazirian wrote the subtractive manufacturing part of the book chapter. B Almufleh reviewed the additive manufacturing part of the book chapter. F Tamimi reviewed and edited the entire book chapter.

Originality: This book chapter reviews the main systems of fabrication dental prostheses by digital technology. This chapter includes production processes, dental applications, available materials and equipment, and the advantages and limitations of digital technology.

Chapter 6 (Manuscript 1):

Title: Removable partial denture alloys processed by laser-sintering technique

Authors: Omar Alageel, Mohamed-Nur Abdallah, Ammar Alsheghri, Jun Song, Eric Caron, Faleh Tamimi.

© 2017 Wiley Periodicals, Inc. *Journal of Biomedical Materials Research Part B: Applied Biomaterials, 106B: 1174–1185, 2018.*

Contribution: O Alageel designed and performed the experiments in the study, analyzed the data and wrote the manuscript. MN Abdallah helped with designing and performing some experiments include hardness micro-CT and cell culture experiments, and he also reviewed the manuscript. A Alsheghri helped with analyzing the mechanical tests and he reviewed the manuscript. J Song helped with designing and reviewed the data of the mechanical tests. E Caron designed and processed the testing specimens. F Tamimi conceived the study, designed the experiments, analyzed and reviewed the data, and co-wrote the manuscript.

Originality: All content of this work is original. This study was designed to characterize the mechanical, physical, and biocompatibility properties of two different laser-sintered RPD Co-Cr alloys and compare them to the cast RPD Co-Cr alloys. This study is the first study comparing the two laser-sintering methods (SLS and DMLS) for processing RPD Co-Cr alloys with the traditional casting method. Moreover, this is the first study assessing the fatigue resistance of laser-sintered Co-Cr alloys, as compared to cast Co-Cr alloy.

Chapter 7 (Manuscript 2):

Title: Determining the retention of removable partial dentures

Authors: Omar Alageel, Ammar A. Alsheghri, Suliman Algezani, Eric Caron, and Faleh Tamimi.

Accepted and published at Journal of Prosthetic Dentistry, 2018.

Contribution: O Alageel designed and performed the experiments of the study, analyzed the data and wrote the manuscript. An Alsheghri helped with analyzing the mechanical tests and reviewed the manuscript. S Algezani and E Caron helped with designing the testing samples. F Tamimi conceived the study, designed the experiments, analyzed and reviewed the data, and co-wrote the manuscript.

Originality: This work is original and innovative. This study established a new approach for predicting and optimizing RPD retention using experimental data of forces produced by food and clasps during mastication. This study presented the first engineering model for predicting RPD retention for each form of edentulism. A working version of the model has been developed and made available online at www.ebhnow.com/apps/0160.

Chapter 8 (Manuscript 3):

Title: Design-driven prediction of removable partial denture retention is associated with patient satisfaction

Authors: Omar Alageel, Nida Ashraf, Marion Bessadet, Emmanuel Nicolas, and Faleh Tamimi.

Submitted to the Journal of Prosthetic Dentistry, 2018.

Contribution: O Alageel designed and performed the study, collect and analyzed the data and wrote the manuscript. N Ashraf helped with collecting the data from McGill University and reviewed the ethical approval and the manuscript. M Bessadet and E Nicolas helped with collecting the data from Estaing University hospital (Clermont-Ferrand, France). F Tamimi designed and supervised the study, also reviewed and edited the manuscript.

Originality: All content of this work is original. This study investigated different factors that influence the patient's satisfaction with their RPDs. In addition, this study is the first one to evaluate and clinically validate the new model for predicting RPD retention that was developed recently. Thus, this study opens the door for further digitalization of the fabrication process and design optimization of RPDs, and consequently, has the potential to enhance the quality of life for millions of patients worldwide.

Chapter 1: Introduction

1.1. Thesis Outline

This thesis is prepared in a manuscript-based format, and it is divided into 11 chapters including one book chapter and three manuscripts. Chapter 1 provides a brief general introduction, research rationale, hypotheses, and objectives. Chapter 2 includes a background and literature review of the types of edentulism and the treatments of each type of the edentulism. Chapter 3 is a book chapter prepared to discuss digital technology in dentistry for the fabrication of dental restorations. Chapter 4 describes the main testing methods performed in this thesis. Chapter 5 contains a list of references for the materials presented in chapters 1 to 4. Chapters 6, 7 and 8 include three manuscripts for the Ph.D. research that prepared by the candidate. The first two manuscripts are published, and the third one is submitted and under review. Chapter 9 draws the general conclusions, Chapter 10 the limitation and future directions, and Chapter 11 is the appendix.

1.2. Thesis Research Rationale

A removable partial denture (RPD) is a simple, cost-effective, and highly functional dental prosthesis that is used to restore missing teeth in partially edentulous patients [1, 2]. This type of denture can be removed and reinserted by the patient, and it is retained and supported by the remaining natural teeth, oral tissue, or dental implants [3]. RPD is an important treatment option that has a significant impact on improving the quality of life for millions of patients worldwide [1, 2, 4]. Even with the great success of dental implant treatments, RPDs remain as the main treatment option used in clinical practice for their economic and functional advantages [1]. In addition, the need for RPDs in the world continues to grow since the number of partially edentulous patients is increasing [1]. However, there are several complications associated with the current RPD

treatments mainly related to the inadequate quality and design of the provided RPDs, and these complications result in treatment failure and patient dissatisfaction with their treatment [5-7]. In fact, it was found that 66% of patients who wear RPDs were dissatisfied with their RPDs according to the National Health and Nutrition Survey [8]. The reason for dissatisfaction was mainly due to the defects in the RPDs, most for these RPDs presented at least one defect [5, 8, 9]. Therefore, there is a need to improve this treatment [1].

The first known teeth replacement for a partially edentulous arch date back to the seventh century and they were done using animal or human teeth to replace the missing teeth [10]. In modern dentistry, the first recorded partial denture was done in 1711 by Heister who use a block of bone carved to fit the mouth [11, 12]. Then, the concept of RPD has significantly improved with the development of the metallic framework of RPDs fabrication using the conventional lost-wax technique [11, 12]. In the 1950s to 1970s, much clinical research was carried out to improve the outcomes of the RPD treatments [12]. However, since then there were no major improvements in the concept of fabrication and designing of the RPDs [1].

RPDs frameworks are traditionally fabricated using the casting (lost-wax) technique that has been used in dentistry for more than a century [13, 14]. This technique is a complex and very laborious manual process that involves many steps and procedures sensitive to the skills of the technicians. This process involves the construction of a wax pattern for the RPD frameworks, investing and forming a model, and then pouring the molten metal into the mold. Thus, producing RPDs by casting technique is not only a time consuming and costly, but also it might generate low precision and ill-fitting frameworks [15, 16].

Digital technology has impacted various industries including dentistry. The evolution of digital technology had a major impact on the fabrication of dental restorations since this

technology can reduce the cost, time, and human errors associated with the treatment of dental restorations. Digital technology and CAD/CAM (computer aided design/computer aided manufacture) technology was developed in the 1960s for industrial applications, and it was first introduced to dentistry for production of dental ceramic crowns by Dr. Duret in 1971 [17].

The CAD/CAM technology in dentistry consists of three systems: data acquisition that can be obtained from the optical scans; CAD (computer-aided design) system or software that to create and manipulate the digital data of a 3D object; and CAM (computer-aided manufacturing) system to manufacture the designed structure in the desired materials [18, 19]. Today, 3D optical scanners either with intraoral or extraoral scanning systems are capable of precise capture and digitalization of patients' jaws and mouths [1]. In addition, special CAD software is now available for designing the for dental prostheses [1]. For the CAM system, the subtractive manufacturing technique (machining and milling) is a good option for fabricating fixed dental prosthesis such as dental crowns, but it cannot fabricate the metallic frameworks of RPD due to the complexity of the RPD structure [20].

In the last decade, new additive manufacturing (AM) processes such as laser sintering and laser melting technologies have been developed for processing 3D metal objects. These methods combine computer-aided design (CAD) of various products and their subsequent fabrication using a high-power laser that fuses metal powder in a layer-by-layer pattern [13, 14, 19, 21, 22]. The laser sintering and laser melting techniques enable the fabrication of complex 3D objects with high precision and at a lower overall cost and faster than the casting technique; thus, making them ideal for digital fabrication of RPD metallic frameworks [19, 21].

The fabrication of RPDs using the laser sintering or laser melting technique instead of casting technique could improve the quality of RPDs, reduce the manufacturing cost and time, and

increase the productivity of RPD fabrication by eliminating many manual steps involved in the traditional fabrication method (i.e. cast duplication, wax-up, investing and casting) [14]. However, there is no data available on the mechanical properties and fatigue resistance of Cobalt-Chromium (Co-Cr) RPD alloys processed by this technology, although the fabrication of Co-Cr RPDs by laser sintering/melting technology can affect the mechanical, physical and biocompatibility properties of the alloys [16, 23, 24]. In addition, the biocompatibility of RPD Co-Cr produced by this method remains unknown [16, 25].

Digital technology has other potential advantages for RPDs by opening the door for digitalizing the RPD design. Currently, RPDs are designed based on the preference and experience of dental professionals (dentists or dental technicians) [26, 27]. This often results in poor designs that result in treatment complications.

Designing RPDs is challenging due to the complexity of the RPD framework and to a large number of partial edentulism forms. In fact, there are 65534 possible presentations of partial edentulism in each jaw, and no suitable guidelines are available to provide the optimal RPD design for each presentation [26, 28, 29]. Although there are some guidelines made to facilitate a proper RPDs design, these guidelines are not clear that covers all the edentulism forms. In fact, the available guidelines for designing RPDs are lacking evidence-based support for its mechanical principles to determine the optimal retention for each specific RPD. Thus, these designs might be poor designs that result in poor RPD treatment causing failure and patient dissatisfaction [1, 5, 9]. Therefore, there is a pressing need to improve this treatment modality by developing a better methodology for designing RPDs, which is crucial for improving the quality of the treatment and preventing any complication associated with RPDs.

1.3. Thesis Hypothesis

Laser sintering/melting technology and advanced algorithms can improve the quality and design of removable partial denture treatments.

1.4. Thesis Objectives

The objectives of this thesis are:

1- To characterize and understand the mechanical properties, physical properties, and biocompatibility of RPD Co-Cr alloys produced by two different laser sintering (laser melting) systems and compare them with those made by the traditional casting method.

2- To develop an algorithm for predicting RPD retention and optimizing RPD design.

3- To validate the developed model for predicting the retention of RPDs experimentally and clinically.

Chapter 2: Background and Literature Review

2.1. Edentulism

Edentulism is defined as the absence of some or all the natural teeth (complete or total edentulism). Persons who have lost some but not all the teeth are called partially edentulous, whereas they are called completely edentulous when they have lost all their teeth. The major reasons for tooth loss are dental caries, periodontal diseases, or trauma [30]. Tooth loss is associated with age, smoking, lifestyle, and socioeconomic status [30-33]. In addition, education level, access to oral health services, and insurance coverage are factors contributing to tooth loss [30-34].

The rate of losing teeth increases every 10 years of adult life by 4% to 10% [31, 32]. Despite the variation among countries, it is estimated that around 20% of the global adult population is complete edentulous [1, 31-33, 35]. Although the rate of complete edentulism in the world is declining every year, the rate of partial edentulism is increasing [36]. In addition, the rate of partial edentulism is expected double by the year 2030 exceeding 200 million patients as a result of population aging [1, 31, 32, 35]. In the USA, it is estimated that prevalence of the partial edentulous is about 44% at the age of 20 years and above, and 71.5% at age of 65 to 75 years [1, 31]. In Canada, the prevalence of partial edentulism is estimated to be between 47% and 58% for people over 65 years old [9, 35].

Tooth loss has a negative effect on both general and oral health as well as patients' quality of life [34, 36]. Losing esthetics, speech and chewing functions are the major concerns for edentulous patients [31]. In addition, tooth loss contributes to bone and residual ridge resorption of the jaws which challenges future treatments [31, 32, 36]. Also, rotation, tipping, and extrusion

of the remaining teeth may occur after tooth loss [36]. Diet-related health issues and psychosocial problems are also associated with tooth loss [32, 36].

2.1.1. Edentulism Classification

The location and length of the edentulous gaps in the dental arch can determine the type of dental prosthesis that can be used to restore the missing teeth. There are many possible forms of edentulism depending on the number and state of the missing teeth [36].

There are several classifications for partial edentulism has been proposed [37-40]. The Kennedy classification, named to Dr. Edward Kennedy, is the most commonly used and accepted classification method in dentistry because it is simple and easy to apply [37-40]. The Kennedy classification divides the partial edentulous arches into four classes (Figure 2.1) [37-40]. Kennedy Class I is defined as bilateral edentulous gaps located posterior to the remaining teeth [39, 40]. Kennedy Class is II defined as unilateral edentulous gaps located posterior to the remaining teeth [39, 40]. Kennedy Class III is defined as a unilateral edentulous gap with tooth bounded both anterior and posterior to the gap [39, 40]. Kennedy Class IV is defined as a single edentulous gap located anterior to the remaining natural teeth and crossing the midline [40]. When more than one class is found in the patient, the classification considered is the class with the lowest number. When multiple edentulous gaps present in one arch, the Kennedy Classes are further classified with modification states. The modification was developed by Dr. Applegate. The modifications can only apply to Kennedy Classes I, II and III, but not with Class IV. One extra edentulous space would be a Modification 1, and 2 extra spaces would be a Modification 2.

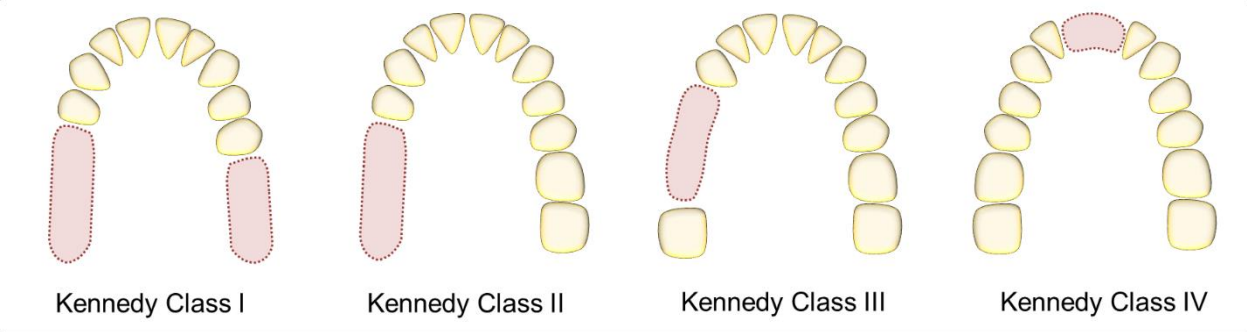


Figure 2.1. A diagram shows the Kennedy classification of the partial edentulism.

2.2. Treatment of Edentulous Patients

The goal of edentulous patient treatment is to restore their masticatory function, esthetics, and speech [39, 41]. There are several options that can be used for treating edentulous patients and selecting the proper options for each case depends on various factors such as the number and the location of the missing teeth, patient's preferences, treatment cost, condition of the oral tissue, and the medical history [39]. The treatment options for the completely edentulous patients and then for the partially edentulous patients will be described underneath.

2.2.1. Treatment of Complete Edentulism

There are several available options to restore missing teeth for complete edentulous patients using removable prostheses [2, 3, 41-43]. The common treatment options are complete dentures and overdentures (Figure 2.2). Complete dentures are considered the standard treatment option for complete edentulism, and they are supported and retained by the residual ridge and oral tissues [38, 44]. Complete dentures provide acceptable esthetics at a reasonable cost; however, poor retention, especially in the lower arch, is the main problem associated with these treatments [3, 38, 44]. The implant overdenture is an excellent treatment option to manage the lack of stability issues with the complete dentures. However, overdentures are more expensive than conventional complete dentures. Overdentures are similar to complete dentures but they are retained on several

dental implants or attachments [3]. Both complete dentures and overdentures consist of a denture base, made of acrylic resin (poly-methyl methacrylate; PMMA), and artificial acrylic resin teeth; while the overdentures might contain metallic frameworks (cobalt-chromium or titanium) attached to the dental implants or supported teeth [3, 38].

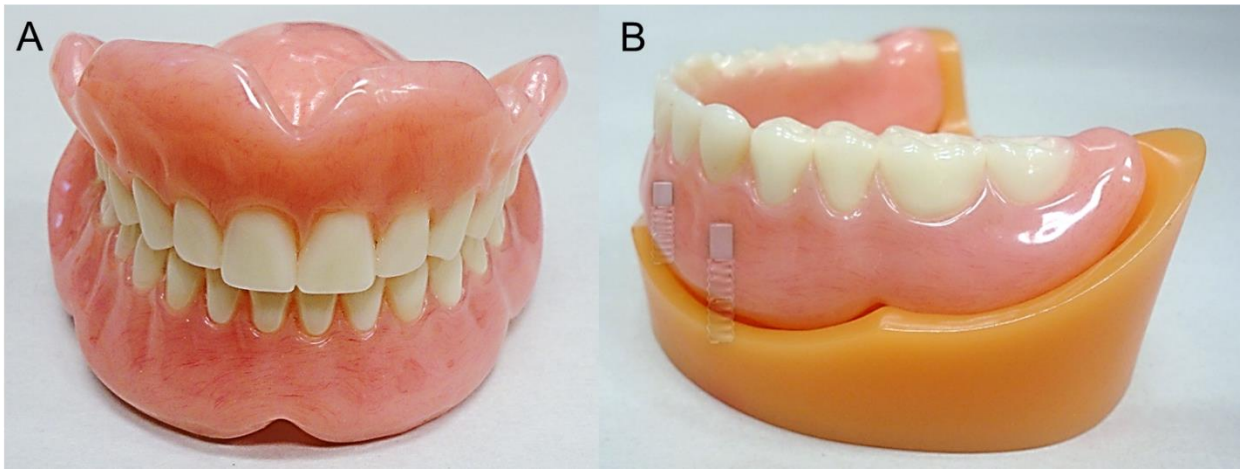


Figure 2.2. Photographs show (A) Complete partial denture for the upper and lower arches, (B) Overdenture supported by dental implants on the lower arch.

2.2.2. Treatment of Partial Edentulism

The common treatment options for partially edentulous arches are fixed partial dentures retained either by natural teeth or dental implants (Figure 2.3), and removable partial dentures (RPDs) retained by natural teeth or dental implants (Figure 2.4) [2, 3, 41-43]. When missing one to three adjacent teeth, fixed partial dentures is the preferred option [3, 41]. A tooth supported bridge is an option for replacing one or two missing teeth by anchoring on the teeth adjacent to the missing teeth [42]. It is effective and affordable; however, it requires trimming the adjacent natural teeth to allow attachment of the bridge [2, 3, 41-43]. On the other hand, a dental crown retained on a dental implant (implant-supported crowns) is considered the preferable option for replacing a few missing teeth [3, 42]. Implant supported crown and bridges have better aesthetic and function

than tooth-supported crowns, but they are expensive and might not be suitable for all patients. Bridges and crowns can be full metal, full ceramic, or a combination of both (metal-ceramic restorations) [45] Fixed partial dentures supported by natural teeth or dental implants are the most suitable types of dental prostheses that can be used for Kennedy Classes III and IV, while fixed partial dentures supported by the dental implants are the most suitable option that can be used for Kennedy Classes I and II. On the other hand, removable partial dentures (RPDs) are an important treatment option for replacing more than three adjacent missing teeth. Underneath, we will discuss the RPD treatment extensively because this thesis will focus mainly on this type of prosthesis.

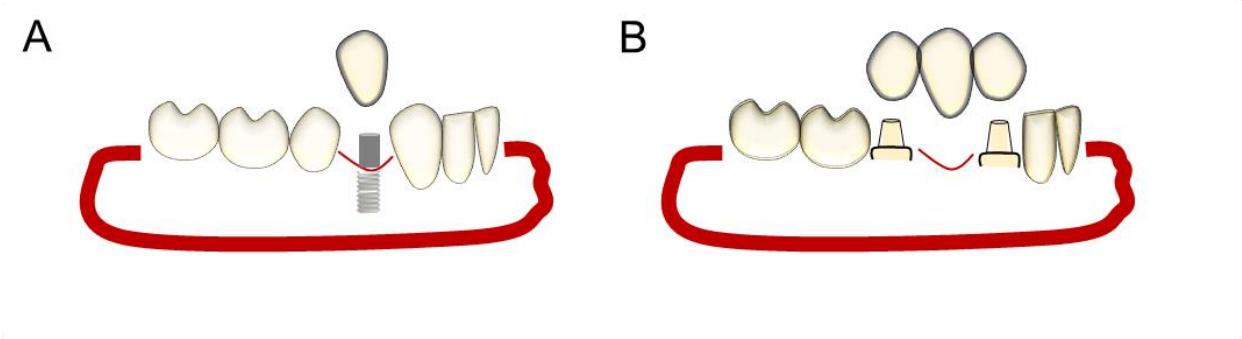


Figure 2.3. A diagram shows an example of (A) dental crown retained by dental implants and (B) dental bridge retained by the natural teeth.

2.3. Removable Partial Denture (RPD)

Removable partial denture (RPD) is a dental prosthesis for replacing the missing teeth for partially edentulous patients (Figure 2.4). RPDs are retained and supported by the remaining natural teeth, oral tissue, and/or dental implants, and are designed to be removed and reinserted by the patient [3]. This type of treatment is very important for millions of patients in the world and is the most common treatment option for partially edentulous patients [36]. It is estimated that around 10-19% of the population in European countries are using RPDs and over 13% of the adult population in North America wear RPDs [4, 8, 36].

RPDs are suitable for patients with partial edentulism who are not able to have fixed prostheses with dental implants because of their clinical, psychological, or economic conditions [38]. In addition, RPDs are cost-effective prostheses as compared to dental implants prostheses, and they provide acceptable function and esthetics [3]. Also, RPDs do not require extracting or trimming the teeth adjacent to the missing teeth as compared to dental bridges. However, RPDs are less comfortable and aesthetic than other fixed dental prostheses. RPDs also may present a risk of potential harm to the remaining teeth and oral tissues such as caries, plaque accumulation, and periodontal disease if oral hygiene and maintenance of denture were not done properly [36, 46, 47].

2.3.1. RPD Type

Removable partial dentures (RPDs) can be classified into two groups: metal-based and acrylic-based RPDs (APDs) [48]. The metal-based RPDs are stronger than the acrylic-based RPDs due to the property's differences of the used materials [1]. In addition, metal-based RPDs provide higher thermal conductivity than acrylic-based RPDs [1]. Therefore, acrylic-based RPDs are usually considered as temporary dentures (interim RPD) and not recommended as long-term prosthesis because of their inadequate mechanical strength of the acrylic resin [3]. On the other hand, acrylic-based RPDs are simple in design, easy to fabricate, and they are at lower cost than metal-based RPDs [1, 48, 49]. Also, acrylic-based RPDs have better esthetic than metal-based RPDs since their clasps (wrought-wire clasps) are less visible because they are flexible [50]. Although acrylic-based RPDs are usually considered as temporary dentures (interim RPD), they have been used as definitive prostheses in many dental practices especially if they are designed and fabricated correctly [1, 48, 51].

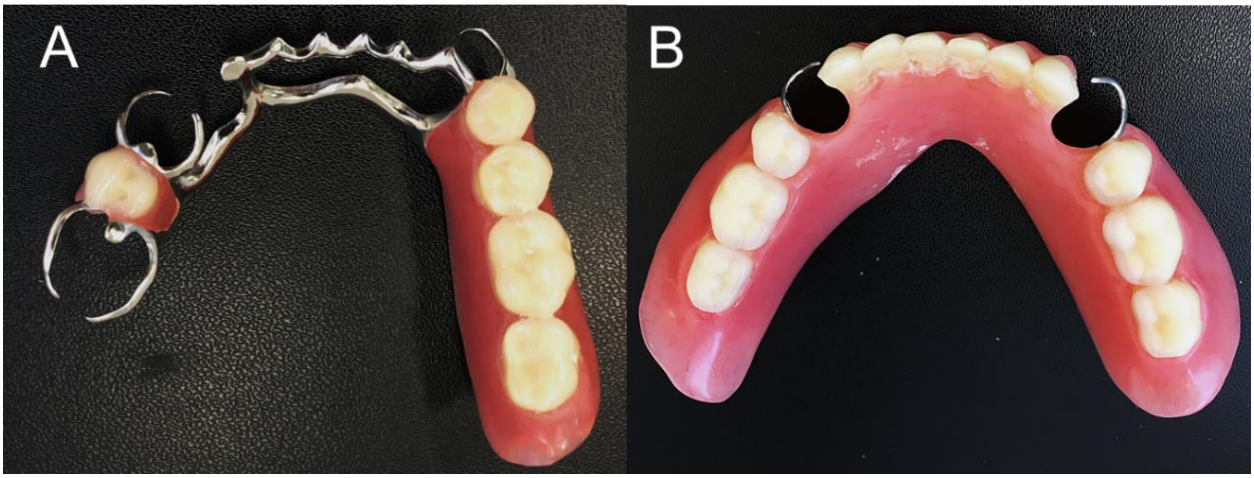


Figure 2.4. Photographs show (A) metal-based removable partial denture (RPD) and (B) acrylic-based RPD with stainless steel wrought wire clasps on the mandibular arch.

2.3.2. RPD Components

The typical removable partial denture, metal-based RPDs, are composed of cast metallic frameworks, acrylic resin base (poly-methyl methacrylate; PMMA), and acrylic teeth (Figure 2.4 A). Whereas, acrylic-based RPDs are composed of wrought-wire clasps, acrylic resin base, and acrylic teeth (Figure 2.4 B). The framework of the metal-based RPD consists of 6 parts: major and minor connectors, direct and indirect retainers, rests, and denture base [3, 38, 39, 52, 53]. Usually, all typical metal-based RPDs have similar components but in different shapes and positions.

Firstly, a major connector is a unit used to connect all the main components of one side in the RPD with the other side [36, 39]. While a minor connector is a component used to connect the major connector with the other part such as direct and indirect retainers and denture base (Figure 2.5) [36, 52]. The major and minor connectors provide rigidity to the RPD by distributing the occlusal forces to the abutment teeth and oral tissues [36, 39, 52]. There are different types and shapes of major and minor connectors available for each edentulous arch [36, 39]. Some examples

of the major connector are the palatal and lingual bars, plates, or straps. Meshwork, lattice work, and bead/ wire/nail are the main design examples of the minor connector.

A direct retainer, such as a clasp, is a unit that engages the tooth to provide retention and to resist movements against dislodging forces (Figure 2.5). The direct retainer can be intracoronal or extracoronal [39, 53, 54]. One common example of the intracoronal direct retainers is the attachment, whereas, clasps are the most common extracoronal direct retainers [36, 39]. Clasps are composed of a rest, a retentive arm (buccal arm), a reciprocal arm (lingual arm) [39, 52, 53]. The rest provides vertical support to the RPD, and it can be positioned on the occlusal, lingual, or incisal tooth surface. Clasps can have different designs including circumferential clasps, or bar clasps [52, 53]. The circumferential clasp is the most commonly used clasp because it is simple, and it provides good retention, but it is less esthetic than the I-bar clasp [36]. The bar clasp is another common type of clasp, and it is mainly used with RPD edentulous arch of Kennedy Class I and Class II [36, 39]. Bar clasps can have different shapes such as the I-clasp, the T-clasp and the Y-clasp [36, 39]. However, the amount of retention provided by clasps can vary according to the length, diameter, cross-sectional form of the clasps arm, and the alloy composition [36, 39]. Longer clasps, thinner clasps, or clasps with round shape cross-sections are more flexible than shorter, thicker clasps or clasps with half-round shape cross-sections, but the flexibility leads to mechanical failure [36, 39].

An indirect retainer is a unit that prevents rotation or movement of the distal extension of the denture base of the RPD, and it only used on RPDs with free-end saddle, edentulous arch Kennedy class I and II. The indirect retainer usually consists of a rest positioned as far as possible from the fulcrum line (imaginary line around which RPD tends to rotate) [36, 39, 55]. Finally, the last part of the RPD framework is a denture base which is the main unit of the RPD covering the

residual ride and support the teeth. The denture base has different designs for maxillary or mandibular arches, and it can be made of metals or acrylic resin.

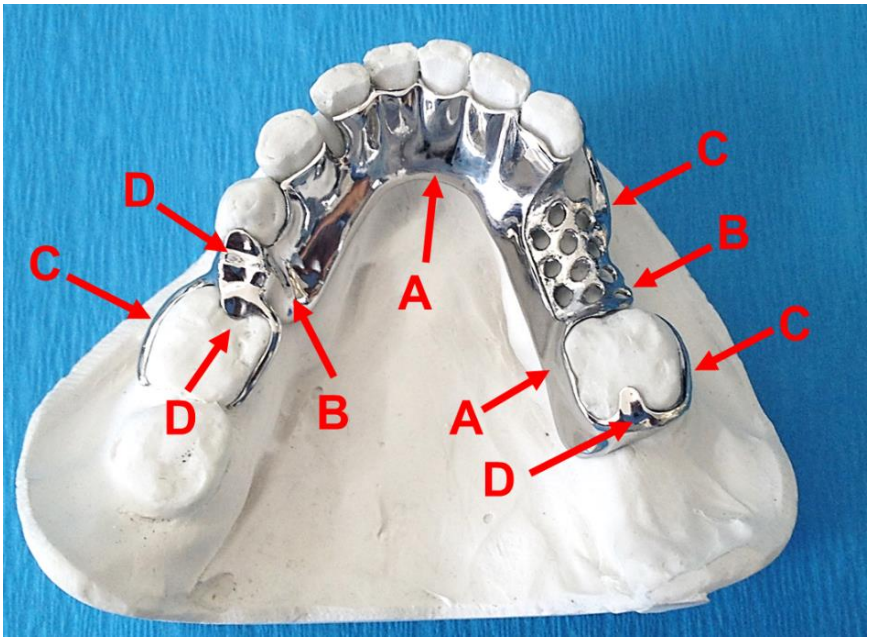


Figure 2.5. A photograph shows the component of the metal framework of RPD: (A) the major connector, (B) the minor connector, (C) the direct retainer, and (D) the rests.

2.3.3. RPD Design

Designing RPDs is a challenging process because there are 65,534 possible combinations of partially edentulous arches in each jaw [36]. Each RPD has its unique design for the edentulous arch, and RPDs should be designed according to clinical and biomechanical principles [36]. The main principles of designing RPDs are considering the rotational movements, levers, displacements, and load transfer of the RPDs [36, 49]. Some of these designing principles depend on the classification of the edentulous arch. For instance, RPD designs on Kennedy class I and II (free-end saddle) arches should have support from both the teeth and oral tissues which the lever principle should be considered. On the other hand, RPDs on Kennedy class III and IV arches should be supported only by the remaining teeth [38]. The rotational movement is the most

common movement happening in the RPD, and it can take place in different planes (horizontal, vertical, or sagittal). This can be prevented by placing the correct RPD components in place. On the other hand, there are some biological and clinical factors that should be considered within designing the RPD. These are factors such as the length of the edentulous span, the type of oral mucosa, quality of the tooth, the occlusion, and salivary flow.

2.3.4. RPD Materials

Cobalt-chromium (Co-Cr) is the main dental alloy used in the metal RPD frameworks while poly-methyl methacrylate (PMMA) is the main material used in acrylic-based RPDs [56-58]. Underneath is an overview of these dental materials.

2.3.4.1. Cobalt-Chromium (Co-Cr) Alloy

Cobalt-chromium (Co-Cr) alloys used in removable partial denture (RPD) are a combination of different metallic elements including cobalt (Co), with a chemical percentage of 60-65%, chromium (Cr), with a percentage of 25-30%, and molybdenum (Mo), with a percentage of 4-6% [56, 57, 59]. Cobalt (Co) provides the strength, and the hardness to the alloy and the presence of chromium (Cr) and other elements improve the mechanical, biological and corrosion properties of the alloy [56, 60, 61].

Co-Cr dental alloys have been extensively used in dental prostheses because they are inexpensive, biocompatible, and provide good mechanical properties. The biocompatibility of Co-Cr alloys is confirmed to the ASTM F75 standard [59]. Generally, the biocompatibility is related to corrosion resistance and ion release in the biological environment [60]. The release of Co and Cr ions in the oral cavity can be toxic, however, Co-Cr dental alloys present good corrosion resistance that reduces the release of these ions into the oral environment [60]. Furthermore, the mechanical properties of Co-Cr dental alloys are suitable for RPD [56, 57, 59]. For instance, Co-

Cr alloys present high strength, modulus of elasticity, and hardness [62]. However, Co-Cr dental alloys show low ductility [62].

Within the crystal structure of Co-Cr-Mo alloys, Co atoms have a hexagonal original close-packed structure ε (hcp), that is transformed to a face-centered cubic structure γ (fcc) at 400 °C during casting procedure [63]. The Cr element also presents a body-center cubic structure (bcc). The γ (fcc) phase improves the strength compared to the ε and σ phases [63, 64]. The γ phase can be converted with high pressures into the ε phase that presents an (hcp) structure [63].

2.3.4.2. Poly(methyl-methacrylate) (PMMA)

Poly(methyl-methacrylate) (PMMA) is an acrylic resin polymer, and it is the main material used in the denture base of RPDs. PMMA was introduced in dentistry in the 1930s and then it became the most commonly used polymer in dental devices [65]. PMMA has been used in dentistry for different applications such as impression custom trays, temporary crowns and bridges, denture bases on complete and partial dentures, and in orthodontic and maxillofacial appliances [56]. PMMA is widely used in dental and orthodontic devices because of its good esthetic, biocompatibility, and mechanical properties as well as due to its lightweight, and high strength and toughness compared to other polymers [66].

PMMA can be classified into three types according to the mechanism of initiating the polymerization, that is chemical (self, auto)-curing, or heat-curing PMMA polymerizations [67]. In the self-cured PMMA, the polymerization process initiates with a chemical initiator at room temperature [56]. The polymerization process of the heat-cured and light-cured PMMA activate with a heat source (e.g. hot water bath, microwave) [56, 68]. The self-cured and heat-cured PMMA are usually supplied the form of a powder and a liquid and are polymerized by mixing the liquid with PMMA powder in a correct ratio [68].

2.3.5. RPD Fabrication

The laboratory procedure for fabrication of metal-based and acrylic-based RPDs is usually similar regarding the processing of the acrylic denture base and teeth, but it is different for the processing of the framework [39]. Acrylic-based RPDs can contain wrought wire clasps that can be formed quickly with premade stainless steed wrought wire [39]. The fabrication of the metal framework of the metal-based RPDs can be done with different techniques [39, 59, 69]. The common and traditional technique is the casting or lost-wax technique [2, 39, 58, 59, 69].

Casting is a manufacturing procedure in which a liquid metal or alloy is poured into a mold containing a hollow cavity that is designed according to the desired shape [70]. After the molten metal solidifies in the mold cavity, the cast metal is removed from the mold [70]. This technique is an old technique that has been used in dentistry for more than a century after it was initially introduced by Dr. Philbrook in 1896 [13].

The fabrication of the metallic framework of metal-based RPDs using the casting technique is a long and time-consuming process that starts with the generation of a stone master cast (a replica of the patients' edentulous arch) from the final impression (the negative imprint of the teeth) [2, 39, 71]. The master cast model is then duplicated with dental stone since the original master cast might be damaged during the fabrication process [2, 39, 71]. Next, the undercuts on the duplicated master cast are blocked out with wax and then relief is added with wax and placed on the master cast [2, 39, 59]. Then, this master cast is duplicated again but with refractory material, which will resist heat temperatures during the casting procedure (Figure 2.6 A) [2, 39, 71, 72]. A wax-up of the RPD framework is then made on the duplicated investment cast according to the provided RPD framework design (Figure 2.6 B). The wax-up can be made with commercial pre-formed plastic patterns available with different sizes and shapes [2, 39, 71]. The wax-up RPD framework is

attached with sprue access premade in wax which later on serves as the path for eliminating the wax from the mold during the wax burnout and for injecting the molten metal into the mold cavity.

After completing the RPD wax framework, the mold is generated with an investment material that can withstand the high casting temperatures (Figure 2.6 C) [2, 39, 71, 72]. The mold is heated in a furnace to burn out the wax from the mold and therefore leaving a cavity in the mold [70]. Next, the mold is placed in the centrifuge casting machine that melts the ingot of metal (such as Co-Cr) and casts the molten metal into the mold cavity [2, 39]. Finally, the metal framework is removed from the casting mold and the framework is finished and polished before returning it to the dentist on the original master cast (Figure 2.6 D) [2, 39, 71, 72]. The framework polishing can be done either manually with handpieces and burs or by an electrochemical polishing process.

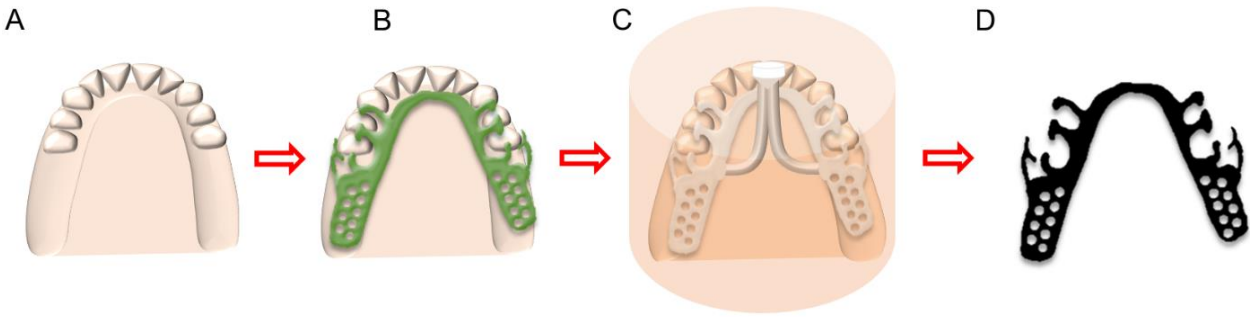


Figure 2.6. The fabrication process of the RPD metallic framework using the casting technique: (A) Dental stone model of the edematous arch, (B) Waxing up the RPD framework on the dental model, (C) Making mold before casting the metals into the mold, (D) The cast RPD metal framework.

After processing the framework for metal-based RPDs or preparing the wrought wire clasps for the acrylic-based RPD, the denture base and the ready-made acrylic teeth can be added (Figure 2.7). First, the ready-made teeth are placed, arranged, and adjusted on an occlusal rim made with wax seated on the metal framework [2, 71]. The teeth adjustment should be done in the

clinic with the patient allowing an accurate teeth arrangement [2]. Then, the RPD denture base is added properly with smooth dental wax. Next, the RPD wax-up is invested in a processing flask and heated with hot water for eliminating the wax and making the mold cavity for the RPD denture base. The acrylic resin (PMMA) is applied and packed on the flask and then heat-cured according to the acrylic resin manufactures [2]. Finally, the surfaces of the RPD denture base is trimmed and polished before sending it to the dental clinic.

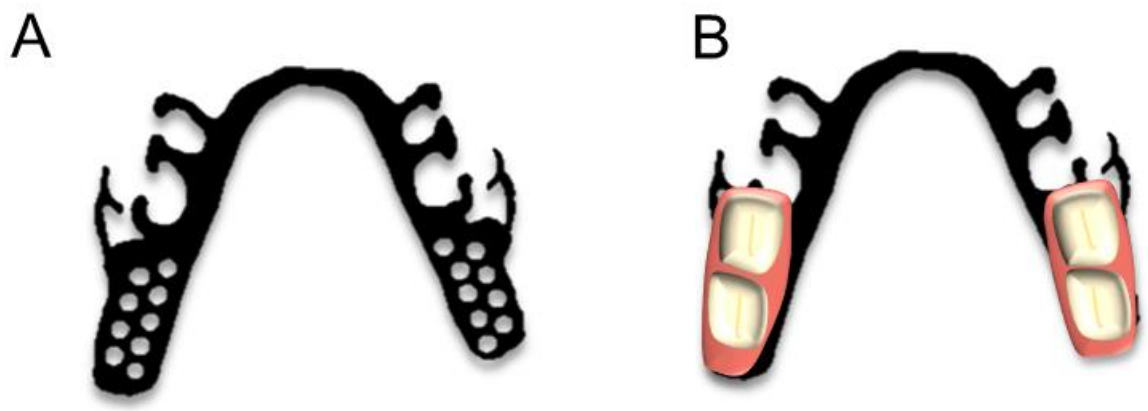


Figure 2.7. The fabrication process of the acrylic part of RPDs: (A) RPD framework, (B) The RPD framework with acrylic denture base and acrylic teeth.

2.4. Patient Satisfaction with RPDs

Patient satisfaction is an important tool for evaluating the success and the effectiveness of treatment [1, 73, 74]. In fact, evaluating the RPDs treatment should not only be evaluated based on clinical estimates but also based on patients' opinions and feedback since patient dissatisfaction with the treatment will lead to underuse and subsequent treatment failure [1, 73, 75]. Generally, a patient's satisfaction with their RPDs is mainly related to comfort, retention, and mastication

ability while esthetics and speech also have significant impacts on patients' satisfaction with RPDs [9, 75-78]. There are many factors contributing to patient satisfaction with an RPD [74]. Table 2.1 summarizes some of the previous literature about the factors that affect patients' satisfaction with RPDs.

Demographic factors can influence patients' satisfaction with an RPD [74]. Patient satisfaction with an RPD can be influenced by patient age, gender, cultural background, socio-economic status, lifestyle, previous dentures experience, and general health [73-75, 79, 80]. For instance, Koyama et al. and Frank et al. found that patient satisfaction is correlated with age [74, 80]. Patients who were older than 60 were more satisfied than those who were younger than age 60 [80]. Patient satisfaction with an RPD can be influenced by the gender depending on the culture and population of patients. For example, Awad et al. found that female patients were more satisfied with their dentures than males, and this study was done on 255 edentulous patients in Canada [73]. Another study done in India on 75 patients found that male patients were more satisfied with their RPDs than female patients [81]. Previous RPD experience also affects patient satisfaction as patient that had a previous RPD tend to be more satisfied [79, 80]. On the other hand, patient satisfaction is lower when there is an opposing denture or when there is poor health condition [80]. Moreover, Zlataric et al. conducted a study on 205 RPD patients on Croatia, and they found that there was no association between age, lifestyle, social and economic status, and general health to patient satisfaction with their RPDs [75].

Factors related to RPD retention such as the number of clasps, number, and location of missing teeth, as well as the number of occlusal rests could also impact patient satisfaction. Only a few studies have studied the relationship between patient satisfaction and some of these factors. For instance, Zlataric et al. found that there was no relationship between patient satisfaction and

the type of edentulism although the number of missing teeth did show a negative impact on patient satisfaction with the RPDs [75]. In addition, Koyama et al. found that patients' satisfaction associated with edentulism location, number of occlusal rests, and number of occluding of teeth but not with the number of missing teeth [74].

Table 2.1. Factors that have been reported to influence RPD patients' satisfaction or usage.

Factor	**Effects**	**References**
Age:		
	Older age increases patient satisfaction.	[80], [74], [82], [83]
	Age has not been associated with patient satisfaction.	[79], [75], [84], [85]
Gender:		
	Female patients are more satisfied.	[73]
	Male patients are more satisfied.	[81]
	Gender has not been associated with patient satisfaction.	[74], [80], [82], [84], [86], [85], [75]
RPD experience:		
	Patient with previous RPD experience are more satisfied.	[80], [79], [87], [88], [83]
	Previous RPD experience has not been associated with patient satisfaction.	[74], [75]
Opposing denture:		
	Patient with opposing denture are less satisfied.	[80]
	Opposing denture has not been associated with patient satisfaction.	[74], [89]
Socio-economic status:		
	Socio-economic status has not been associated with patient satisfaction.	[75]
Life-style:		
	Life-style has not been associated with patient satisfaction.	[75], [85]
Education:		
	Patient with higher education are less satisfied.	[75]
	Education has not been associated with patient satisfaction.	[85]

General health:		
	Patient with poor general health are less satisfied.	[80]
	General health has not been associated with patient satisfaction.	[75], [85]
RPD quality and design:		
Mandibular arch:		
	Patient with RPD in the mandibular arch are less satisfied.	[82], [81]
	RPD in the mandibular arch has not been associated with patient satisfaction.	[74]
Free-end saddle RPD:		
	Patient with free-end saddle RPD are less satisfied.	[74], [90], [84], [87]
	Free-end saddle RPD has not been associated with patient satisfaction.	[75], [80], [82], [91]
Occluding teeth:		
	Higher number of occluding teeth has not been associated with patient satisfaction.	[74]
Occlusal rests:		
	Higher number of occlusal rests has not been associated with patient satisfaction.	[74]
Missing teeth number:		
	Patient with RPD with higher number of missing teeth are less satisfied.	[75]
	RPD with higher number of missing teeth has not been associated with patient satisfaction.	[74], [80]
Pain:		
	Patient with pain while using RPDs are less satisfied.	[74], [92]
RPD condition (teeth colors and shapes):		
	Patient with RPD in unpreferable condition are less satisfied.	[74]
	RPD with unpreferable condition has not been associated with patient satisfaction.	[74]
RPD materials:		
	RPD materials and clasp type has not been associated with patient satisfaction.	[74], [82]

2.5. RPD Challenges

There are many complications associated with RPDs causing failure and might lead to harm of the remaining teeth and oral tissue [1, 78]. These complications might result in patients' dissatisfaction with their RPD which represents a waste of time, money, and psychological energy [80]. Indeed, the acceptance rate of RPDs by the patients is very low. Previous studies reported that around 30-50% of RPD patients were dissatisfied with their dentures [74, 78, 91, 93]. In addition, a study in the USA evaluated 1306 RPDs and found that 65% of the RPDs had defects [8, 9]. Other studies also evaluated RPDs and found that around 60% of RPDs had at least one problem [8, 94].

The most common complication associated with RPDs is the loss of retention [1, 9, 95]. The loss of retention affects patient satisfaction with their dentures, and also it can affect patients' mastication ability [9]. The dentures clasps are the components that get damaged most often mainly due to clasp arm deformation or breakage [96-99]. Indeed, clasp deformation and fatigue fracture caused by repeated loading, are the most common mechanical failures presented in the RPDs [100, 101]. This is because clasps are subjected to movements in response to functional loads during mastication or insertion and removal of the RPDs from the mouth [96-99]. A previous study evaluated RPDs after 8 years of normal use, and it found that the majority of RPD metal clasps were distorted over time and they did not fit the abutment teeth correctly [1, 102]. In fact, RPD retentive clasp arms must resist permanent deformation and fatigue fracture during insertion and removal and mastication, and at the same time, it must also be flexible in order to engage the undercuts of the abutment teeth [103].

Generally, RPD failures could be related to the poor designing and/or the poor manufacturing of the RPD [9]. To minimize this issue, RPDs should be well designed and

fabricated properly. The issues related to the RPD designs and quality will be discussed underneath.

2.5.1. RPD Design

The design of an RPD is an important factor for its success [5]. In fact, a previous study found that around 90–96 % of properly designed RPDs were in function after 5 years of usage [36]. However, designing a retentive RPD by dentists or dental technicians is extremely challenging not only because of the complexity of RPD components and the manufacturing process; but it related the type and form of the edentulism [28, 29, 36]. The dentist has to take into account many factors such as the distribution of the undercuts and tooth angulation of the tooth among others [28, 29, 36]. Therefore, RPDs are often designed subjectively based on the experience of dental professionals and based on the designing guidelines [27]. Due to RPD designing complexity, many dentists delegate design work to dental technicians for their extensive designing experience [104].

There are some guidelines that facilitate designing RPDs; however, many of these guidelines are complex, lack scientific evidence, and do not cover all forms of partial edentulism [2, 29, 39, 105]. In addition, there are some knowledge-based systems available for designing RPDs. This system provides the most relevant RPD design from a database containing RPD designs of previous patients [27, 106]. However, RPD designs in patients' databases are not necessarily optimal since they are also designed subjectively based on operator experience.

Designing a proper RPDs should consider different factors including patient's esthetics, comfort, and the biomechanics of the RPD [107]. Moreover, a properly designed RPD should have adequate retention to resist the dislodging forces caused by food mastication and functional muscle movements [2, 5, 39, 105]. The retention can be achieved by the clasps engaging the undercuts on

abutment teeth, by the interproximal and denture base extensions, and by attachments on teeth or implants [2, 5, 39, 105]. Indeed, clasp retention varies according to type (e.g. circumferential, I-bar, and wrought-wire), number, location, undercut depth, and composition [50, 97, 108, 109].

An important question raised upon designing an RPD is determining the adequate number of clasps to provide enough retention to resists the dislodging forces produced by mastication. RPDs with too many clasps could cause harm and discomfort to the patient as well as increase the level of gingival inflammation [110]. Whereas, RPDs with too few clasps could result in insufficient retention. Currently, there are no guidelines to determine the optimal number of RPD clasps and optimal retention. Also, there is no established optimal amount of retention needed to achieve a retentive RPD treatment. Therefore, determining the optimal retention of any RPD design could be the key to develop better designing guidelines.

2.5.2. RPD Quality

RPD frameworks are conventionally fabricated using the casting technique which has been used in dentistry for more than a century [13, 14, 111, 112]. Casting is a very laborious manual process that is widely influenced by the skill of the dental technicians [13, 15, 16]. Many errors and defects can occur in cast metal especially with base-metal alloys such as Co-Cr alloys. These errors might be because of the operators or the casting procedure [13, 16, 112-116]. Errors of the casting technique are usually a sum of accumulating errors within each fabrication step including errors on the impression, shrinkage of the dental stone, thermal expansion and softening of the waxes, shrinkage of the cast metal, oxidation or thermal expansion of the metal, and errors during finishing or polishing the metal framework [117]. Furthermore, the casting method usually presents internal defects due to gas inclusion during the casting process [14, 118]. However, RPD metal clasps should be fabricated properly with proper shape, diameter, and thickness without any

deformation, defect or bubble in the metal, which might weaken the clasp. Therefore, fabricating RPD frameworks by casting is not only time consuming but may also generate frameworks with low precision and quality [15, 16].

Chapter 3: Fabrication of Dental Restorations Using Digital Technologies

Chapter 3 is prepared as a book chapter, and it was submitted as chapter 4 for a book entitled "Digital Restorative Dentistry" to be published by Springer Nature.

Authors:

Omar Alageel, Berge Wazirian, Balqees Almufleh, Faleh Tamimi.

3.1. Introduction

CAD/CAM (computer aided design/computer aided manufacture) technology was developed in the 1960s for industrial applications, and it was introduced to dentistry by Dr. Francois Duret in 1971 [17]. Dr. Duret introduced the concept of producing a dental crown using an optical impression of the prepared tooth by an intraoral digitizer and a digitally controlled milling machine [17, 119]. In 1983, he performed the first CAD/CAM restoration, and later he developed the Sopha system [17, 119]. Dr. Mörmann invented the first commercial CAD/CAM system, named CEREC (Computer-Assisted Ceramic Reconstruction), in collaboration with Dr. Marco Brandestini, an electrical engineer, who had the idea of using an optical dental scanner [19]. In 1985, the team performed the first chairside dental inlay using an optical scanner and a milling machine system [17, 19, 120]. In addition, Dr. Andersson also developed the Procera system in 1983 for fabrication of dental crowns, and he was the first person to use the CAD/CAM technology for composite dental restorations [17].

Although the CAD/CAM technologies were first introduced in dentistry in the 1970s for a long time, their use was very limited. In the past few decades, CAD/CAM technologies' usage has

grown dramatically, and is expected to grow further in the near future [19, 121]. For instance, the 3D printing market for USA healthcare is expected to grow by 18-22% between 2015 and 2025 exceeding $5 billion dollars by 2020 [122-125]. Digital technology has many benefits in dentistry such as it is faster, more accurate, and more economical than the traditional procedures [126]. Digital technology in dentistry can eliminate the need for some or all of the manual processes such as pouring casts, die fabrication, restoration wax-ups, investment process, metal casting or pressing porcelain [17]. Therefore, digital technology is rapidly spreading into dental laboratories and clinics around the world, and it is transforming dentistry at an unprecedented pace [127]. In dental laboratories, the traditional equipment (such as furnace and casting machine) is being replaced by computers, scanners, and 3D printers or digital machines. Now, dentists do not need to take an impression and wait a few weeks to fabricate appliances or restorations; instead, they only need to scan the teeth and email the digital file to a dental lab for printing the prosthesis, which may take less than an hour.

In dentistry, CAD/CAM technology consists of three systems:

1- Data acquisition that can be obtained from different scanning technologies such as 3D scans [18, 19].

2- Data processing CAD (computer-aided design) system that creates and manipulates the digital data of a 3D object [18, 19].

3- Data manufacturing CAM (computer-aided manufacturing) system that manufactures the designed structure in the desire materials [18, 19].

CAM technologies available in dentistry can be classified as either “subtractive” or “additive” manufacturing methods. With subtractive methods, also known as machining and milling, dental parts are created by subtracting the undesired material from a block with the use of

burs, disks, or lasers [19, 121]. On the other hand, additive methods, such as 3D printing and laser melting technologies, build dental objects layer by layer [19, 121]. The additive manufacturing process is also known as rapid manufacturing, and it is more recent technology than the subtractive manufacturing process [19]. However, subtractive methods are currently more precise and accurate, while additive methods are more versatile [128, 129]. There is a wide range of available machines for both methods [18]. Each technology presents some differences in the process and materials used, and they have different advantages, limitation, and applications [18]. Details about each technology will be explained in this chapter.

3.2. Subtractive Manufacturing

3.2.1. Machining and Milling

3.2.1.1. Overview of Machining and Milling

Machining and milling, also known as subtractive manufacturing, refers to a process in which a block of raw material is cut into a desired final shape by a controlled material removal technique [19, 121]. The cutting process involves power-driven sharp cutting tools such as saws, lathes, drill presses with different sizes designed to remove small chips from the block of material until achieving the final desired shape [19, 121, 129, 130]. The industrial improvements in software and a reduction of size and costs of CAD/CAM machinery have allowed the application of CAD/CAM in dentistry [120].

The CAD/CAM systems for subtractive manufacturing methods can be classified into chairside systems and laboratory systems [121, 129, 131]. For chairside systems, the fabrication of dental restorations by CAD/CAM can be done in the dental clinic without a laboratory procedure [121, 131]. For the laboratory systems, CAM production takes place in the dental laboratory or production centers [121]. The CAD/CAM systems can also be classified into open and closed systems [121, 129]. Open systems allow all the CAD/CAM components, including data acquisition, design by CAD software, and the manufacture by CAM system, to be provided by different companies, while closed systems are restricted to a single supplier [129].

3.2.1.2. Dental Applications of Machining and Milling

Machining and milling have many dental applications in the fields of prosthodontics and restorative dentistry [129]. These include crowns, copings, inlays, onlays, veneers, frameworks for fixed dental prostheses, and implant abutments and bars [129]. In addition, machining and milling can be used as a burnout pattern for casting, pressing, or over-pressing [129]. Moreover, splint and

orthodontic retainers, complete prostheses, verification jigs, diagnostic wax-ups, and digital models can also be fabricated with machining and milling methods [129].

3.2.1.3. Milling and Machining Production Process

The production process starts once the designing step of the final prosthesis is completed using appropriate CAD software (Figure 3.1). The CAD model then translated by the CAM software into a tool path for a computer numerically controlled (CNC) machine. Following this step, the software will run a simulation in order to confirm the capability of the milling unit to process the designed prosthesis (Figure 3.2). Once the software confirms the feasibility of the designed prosthesis, the CNC machine can be initiated. The CNC machine is composed of several machining sequences, and each sequence is a group of calculated machining tool paths, which are automatically calculated with specific machining algorithms [130, 132]. The CNC machines are composed of multiaxis milling units operated in 3-axis, 4-axis, or 5-axis (Figure 3.3).

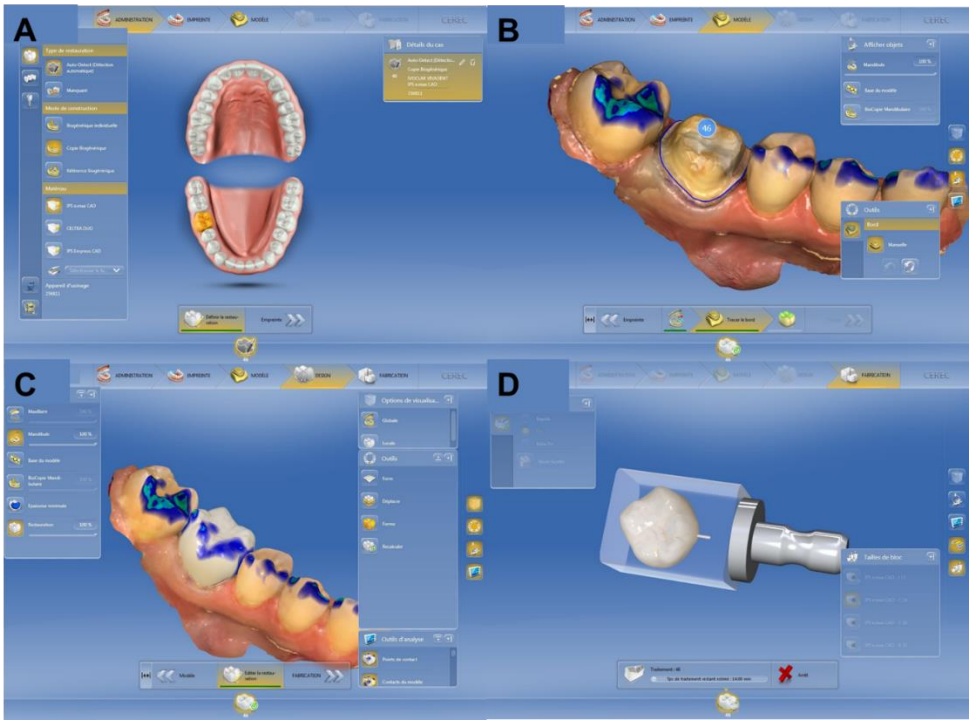


Figure 3.1. Photographs showing the designing a dental crown for milling process through different steps (A) administration, (B) modeling, (C) designing, and (D) fabrication.

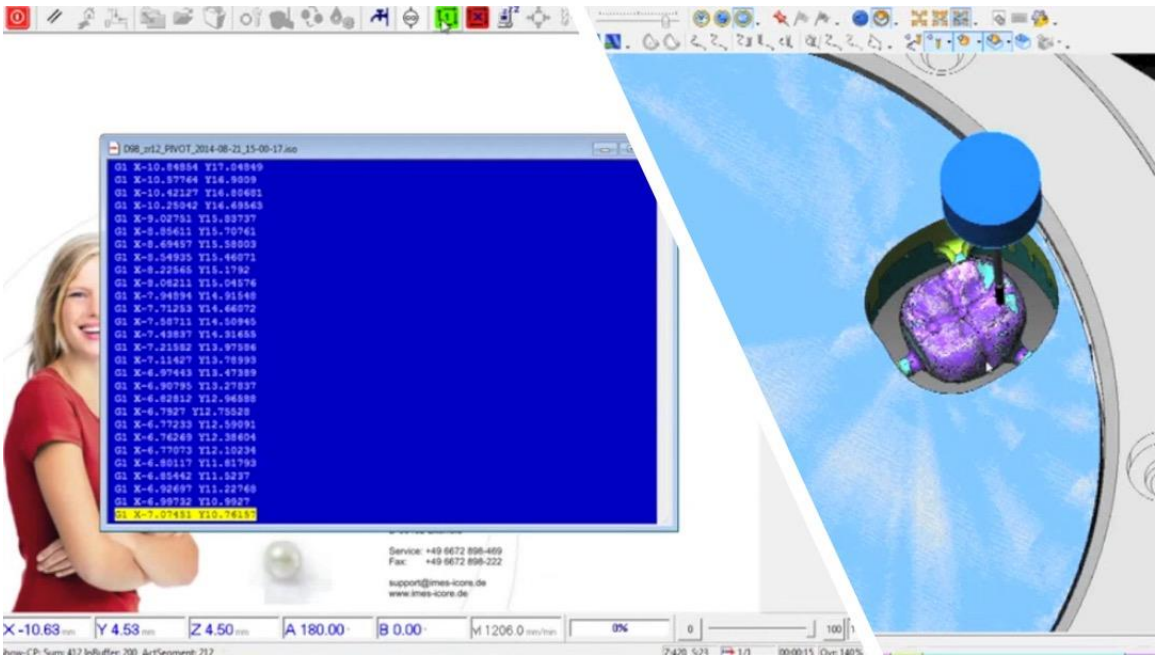


Figure 3.2. Example of calculations and simulation before milling and machining process.

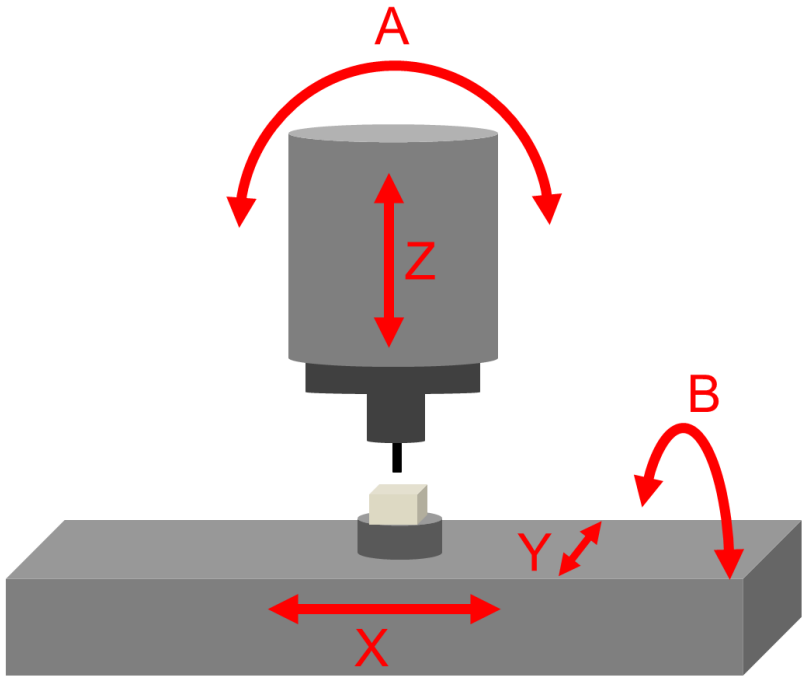


Figure 3.3. Representation of different possible axes: 3-axis includes X, Y, and Z directions; 4-axis includes X, Y, Z, and A directions; 5-axis includes X, Y, Z, A, and B directions.

Three-axis Milling Machine

The 3-axis milling machines are the most commonly used in dentistry (Figure 3.4) [130]. This type of milling machines can move in three spatial directions that are defined by the values X, Y, and Z. The block also can turn 180° during manufacturing to allow the milling of the external and internal surfaces [130]. Thus, 3-axis milling units are faster than other milling units because they have short calculation and cumulative milling time [130]. Also, the simplified control of the 3-axis renders them less expensive compared to machines with more axes [131]. However, 3-axis machines are limited when it comes to producing divergence, convergence, and highly defined features [130].

Four-axis Milling Machine

The 4-axis milling machine involves an additional axis to the three spatial axes, and it can to allow the block to rotate around the X-axis. The fourth axis is defined as tension bridge A. This is useful for milling large blocks for long span frameworks [130]. The tension bridge for the component can also be turned infinitely variable. As a result, it is possible to adjust bridge constructions with a large vertical height displacement into the usual mold dimensions and thus save material and milling time [131]. The 4-axis milling machine can be used for crowns, veneers, Inlays, onlays, copings/frameworks, and fixed partial dentures [129, 130].

Five-axis Milling Machine

The 5-axis milling device contains additional 2 axes that can rotate the block around the X-axis and around the Y-axis (Figure 3.5). The fifth axis is defined as tension bridge B. This enables the milling complex geometries and smooth surfaces with subsections. The 5-axis machines can produce objects with higher accuracy than by 3 or 4 axial milling machines since it

can mill undercuts in all directions [129]. The 5-axis milling can machine produce digital models, implant attachments, denture base, implant abutments, bars, and splints [129, 130].

The milling process can be done in different conditions and forms according to the materials used. It can be done in wet or dry conditions, and also it can be done with soft or hard materials. Dry processing and soft machining are usually applied without cooling liquid, and it is used for machining unsintered zirconium oxide, composite resin, and wax [127]. Wet processing, usually hard marching, uses a spray of cool liquid to protect the milled material and milling burs from overheating; it used with pre-sintered zirconium oxide, metals, and composite resin [127, 129]. Dry processing is less expensive and produces less moisture absorption than wet processing, but it might result in higher shrinkage than wet processing [127, 129].

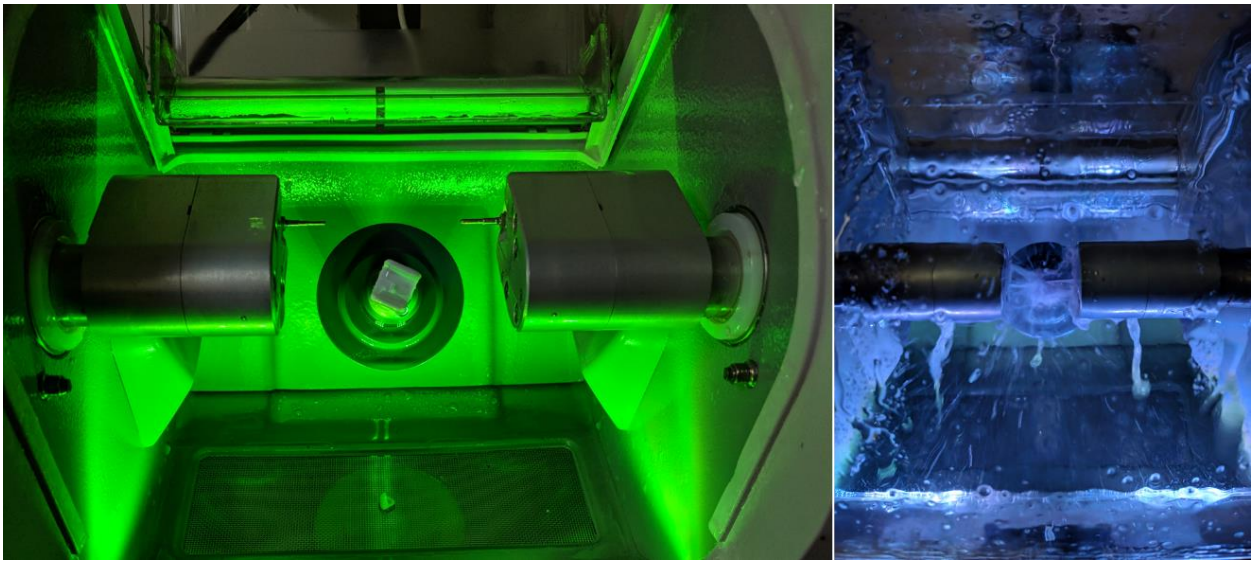


Figure 3.4. Photographs showing the 3-axis milling machine during drilling sequence.

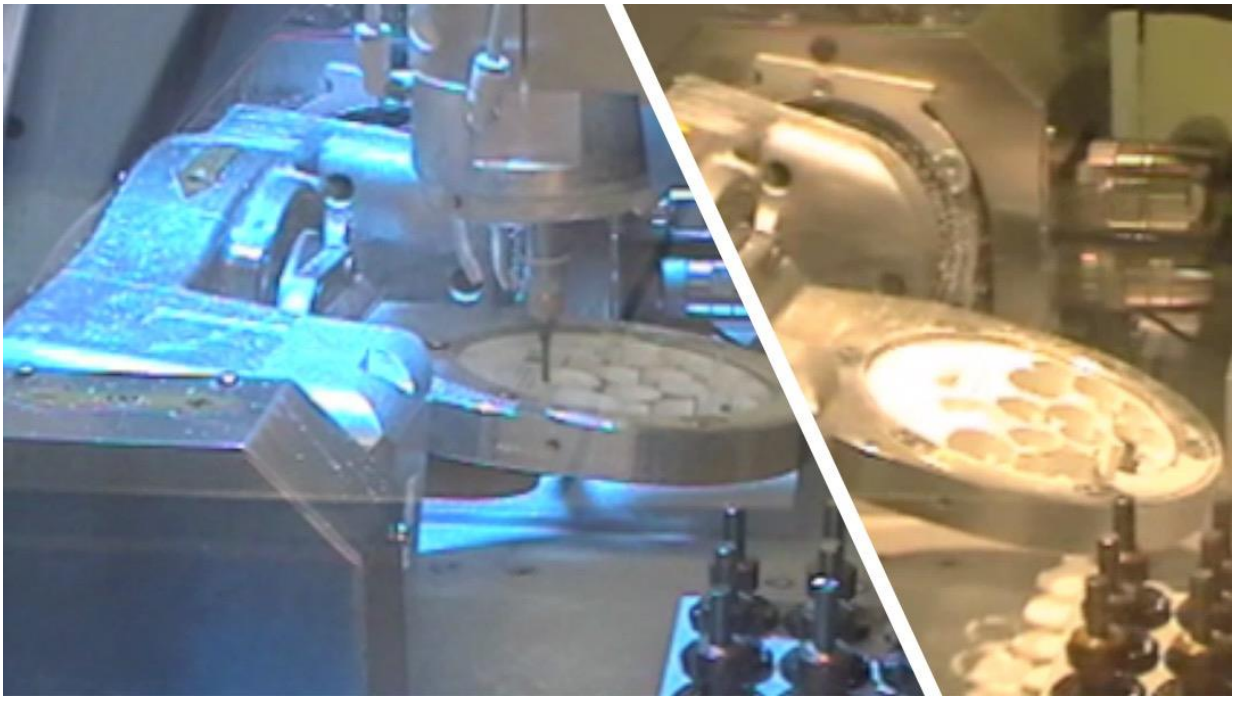


Figure 3.5. Five- axis milling machine during drilling sequence.

3.2.1.4. Advantages of Milling and Machining

There are some advantages to using milling and machining in dental applications in comparison with conventional dental laboratory technologies.

1. High accuracy [131, 133].
2. Standardized manufacturing process [131].
3. Efficient quality control system [131].
4. Increased production capacity [131].
5. Fast production [17].
6. Enable the use of new materials, such as zirconia and titanium [131].
7. Transform laboratories from simple fabrication sites into computerized production centers [131].

3.2.1.5. Disadvantages of Machining and Milling

1. The initial cost of a CAD/CAM system can be higher than the traditional dental equipment [127].
2. Machining and milling is a very wasteful procedure in which more material is removed compared to what is used in the final product [19, 121].
3. The milling procedure accuracy is dictated by the diameter of the smallest bur. Therefore, any surface details less than the diameter of the milling bur will be overmilled [129].
4. The possible uses of CAD/CAM are limited by the capability of the software and the digital scanners available [127].
5. Many current CAD/CAM technologies still require conventional laboratory processing. For example, zirconia frameworks fabricated by CAD/CAM in machining centers require manual veneering with conventional porcelain using by dental technicians [119].

3.2.1.6. Potential and Future Direction of Machining and Milling

Considering the advantages of milling and machining, the application of this technology has become essential in providing appropriate treatment to patients. With the cost of manufacturing units dropping, many laboratories and clinics are acquiring CAD/CAM units for faster fabrication of dental restorations. However, this method of manufacturing is very wasteful as more material is removed compared to what is used in the final product. Around 90% of a block material is removed to create the dental restoration [121]. Accordingly, there has been a major transition from subtractive manufacturing to what is referred to as additive manufacturing. Using additive methods for manufacturing is more advantageous as many problems associated with milling can be readily overcome such as the ability fine detail [19]. However, additive manufacturing is incapable of producing restorations with certain materials such as zirconia, glass ceramic, and composite.

3.2.1.7. Materials

Different materials can be milled by CAD/CAM systems (depending on the system used), and these materials are offered and sold in block form for CAD/CAM systems. Metals including titanium, titanium alloys, and chromium-cobalt alloys can be used with CAD/CAM systems. Noble-metal alloys are not used for cost reasons [131]. Resins can be milled into frames for use in lost-wax casting, and they can also be used to make crowns and long-term provisional fixed partial dentures [131]. Polyurethane is used for the fabrication of digital models [129]. Ceramics blocks are available for the fabrication of inlays, crowns, and veneers, and they can be monochromatic or colored [131]. One such ceramic group is silica-based ceramic such as lithium disilicates that produce natural-looking restorations, thanks to their translucency resembling that of real teeth eliminating the need to add veneering porcelain [131]. Another group of ceramic is the infiltration ceramic blocks such as alumina (Al_2O_3), zirconia (Al_2O_3, ZrO_2), and spinel ($MgAl_2O_4$) [131]. In addition, aluminum oxide (Al_2O_3) and zirconium oxide ($ZrO2_2$ Y-TZP) are milled at the pre-sintered stage and is then sintered; this provides superior mechanical properties such as high strength and high tenacity that are excellent for crowns and fixed partial dentures, as well as implant abutments [131].

3.2.1.8. Equipment

Available systems are designed to mill or grind either in dry or wet conditions as dictated by the type of material used. The selection should also take into consideration the number of axes (3, 4 or 5 axes) and is dictated by the design of the dental restoration [129]. Main dental CAD/CAM systems available are Etkon (Etkon AG), Everest (KaVo electrotechnical work GmbH), Lava (3M ESPE Dental AG), Procera (Nobel Biocare Germany GmbH), Hint ELs DentaCAD system (Hint-ELs GmbH), and CEREC3/inLab (Sirona Dental of system GmbH) [119, 134].

3.2.2. Chairside Solutions

3.2.2.1. Overview of Chairside Solutions

Recent developments of CAD/CAM systems allow the fabrication of dental restorations at the dental chairside without the need for a laboratory procedure [131]. In this context, all CAD/CAM components, such as the scanner, the CAD system, and the CAM system, are allocating in the clinic that saves time and allows the fabrication of the restorations within one appointment [127, 135]. The scanner is used to acquire topographic information of the oral cavity, preparation of the tooth, adjacent teeth, and occlusion. The CAD system is used to design the restorations, while the CAM system is used to convert the information into an actual restoration [135]. Chairside CAD/CAM systems are capable of scanning, designing, and milling within the chairside workflow.

There are two categories of commercially available digital systems for CAD/CAM chairside dentistry: chairside digital impression systems to transfer images to the laboratory, and chairside milling machines for same day restorations [120, 135]. The digital impression systems were developed to replace the traditional impression methods. These digital chairside impression systems include both the hardware for scanning and the software for data analysis. The software captures and stores the digital data from the intraoral scan, and it also records personal information regarding the patient which will allow the replacement of traditionally written laboratory prescriptions by including a comprehensive electronic prescription form. The digital impression can be archived and transmitted to the lab via the Internet. Once the data is transmitted, the restorations can be designed directly from these digital impressions, and then, produced by the CAM system [17, 131, 136].

3.2.2.2. Dental Application of Chairside Solutions

With advances in chairside scanner systems and the ability to image full arches, orthodontic applications of CAD/CAM dentistry have expanded significantly. The chairside CAD/CAM systems allow same day fabrication of inlays, onlays, crowns, veneers, and with improvements in dental material science, they also allow the fabrication of multi-unit restorations, implant abutments/restorations, temporary restorations, and surgical guides [120, 137]. Furthermore, when combined with 3D cone beam computed tomography (CBCT) imaging, the CAD data aids substantially in complex planning surgical treatments [120].

3.2.2.3. Advantages of Chairside Solutions

There are some advantages to using chairside solutions over the laboratory systems.

1. Eliminates the need for a second appointment. Patients appreciate the convenience of having restorations placed in one appointment instead of having to come back for a second delivery appointment [135, 138].
2. Patient information is digitally stored. This saves physical storage space and eliminates the risk of breaking the casts [17].
3. Allows the dentist to have total control of the artistic and creative expression and the manufacturing process without the involvement of the laboratory [135, 138].
4. CAD/CAM systems improve the efficiency and productivity of dental clinics once the initial learning curve period is overcome [135, 138].

3.2.2.4. Disadvantages of Chairside Solutions

1. The high initial and maintenance cost of chairside CAD/CAM systems [138].
2. These chairside systems require special training, and learning curve varies from user to user [138].

3. The possible uses of these systems are limited by the capabilities of the software and milling machines [131].
4. CAD/CAM technology is constantly being upgraded and improved; these alterations must be dealt with as they arise ensuring additional cost in the future [138].
5. Tooth preparations may need to account for limitations of the milling system [139].
6. Closed-data chairside systems in which all components are linked by a unique data format prevent different systems from interacting [139].
7. Chairside CAD/CAM systems have limited materials.

3.2.2.5. Potential and Future Direction of Chairside Solutions

With a newer generation of intraoral scanners, an improvement in the efficiency of scanning provides a better patient experience that treatment results [134]. Most recently, the introduction of portability to intraoral scanning systems has allowed clinicians "plug-and-play" ability. They can use the scanner to obtain data for the designing software that is retained on a central server, using an existing computer and network infrastructure. These scanners eliminate the need for the traditional cart-based system that houses the CPU, viewing monitor, software, and digital intraoral scanner [120].

With this technology becoming readily available, more manufacturers will offer open architecture CAD/CAM systems. Open architecture refers to the format of the data that is acquired during digital scanning as being compatible across multiple, different manufacturers of both software and hardware. An open system allows for the transfer of data across multiple devices for design and final restoration [120]. This will give practitioners the opportunity to combine features from different manufacturers to better meet the needs of their clinical practice.

To provide more sophisticated restorative and prosthetic devices, future prostheses are expected to be designed and fabricated with an improved function related to jaw movements. The analysis of multiple-axis mandible movements in order to reproduce the oral functions of patients has already been widely investigated in prosthodontics. Production of dynamic occlusal morphology during the CAD process is still challenging but must be made practical in the near future to offer restorations that respect the oral function [134]. Additionally, dental CAD/CAM is being used for educations and training purposes to produce explanatory and diagnostic materials for students and patients and for simulations of surgical and reconstructive procedures.

3.2.2.6. Materials

Chairside materials can be categorized as follows:

1- Predominantly Glass Ceramics

The principal features of predominantly glass ceramics are that they contain a glass phase and have excellent translucency and moderate strength. The glass component allows them to be etched with hydrofluoric acid and adhesively bonded to the tooth [140]. Some examples of materials in this category are Vitablocks Mark II and CEREC Blocs [137]. These materials are available in monochromatic or polychromatic multicolored blocks offering the possibility of creating restorations mimicking the transition from dentin to enamel layer. Further customization of either type can be accomplished by shade characterization and glazing [140]. Commonly used for inlays, onlays, and veneers.

2- Leucite-Reinforced Ceramics

These blocks contain a leucite crystal phase which increases their flexural strength without losing their capacity to adhesively bond to the tooth. The percentage of leucite particles varies from 30 to 45% depending on the supplier. Some examples in this category are IPS Empress CAD

from Ivoclar and Paradigm C from 3M ESPE. The IPS Empress CAD blocks are available in different monochromic shades of high translucency (HT), or low translucency (LT) or as polychromatic blocks. Paradigm C is a radiopaque ceramic available in six shades that exhibits a chameleon effect once seated in the tooth due to its enhanced translucency and fluorescence [140]. Customization for both systems can be achieved through staining and glazing. They are commonly used for inlays, onlays, veneers, partial crowns, and crowns.

3- Lithium Disilicate

This ceramic presents 2-3 times the flexural strength of predominantly glass ceramics. Lithium disilicate (IPS Emax) was initially developed as a substructure material that offered greater translucency compared with other ceramic core materials, and it uses as a monolithic restoration for chairside CAD/CAM systems as it has gained popularity due to the enhanced strength [141]. The CAD/CAM block form is available in four translucency levels (high translucency, medium translucency, low translucency, medium opacity) and in different shades for each category [135, 141].

CAD/CAM lithium disilicate is acquired as blue-violet partially crystallized blocks that are easily milled without excessive damage to the material. After milling, the restoration must undergo a firing process in a porcelain oven to complete the crystallization of the lithium disilicate. This process converts the blue shade of the pre-crystallized block to the selected tooth color and increases the flexural strength of the restoration to its final level [141, 142]. This material can be used for inlays, onlays, veneers, partial crowns, single crowns, three-unit fixed dental prostheses in the esthetic zone, and implant superstructures, as well as hybrid abutments and hybrid abutment crowns.

4. Zirconium Oxide and Lithium Silicate

Zirconium oxide and lithium silicate glass ceramics (ZLS) are available in a fully crystallized or pre-crystallized [143]. The fully crystallized ZLS ceramics are more difficult for machining, while, pre-crystallized ZLS ceramics are easy to machine. ZLS ceramics contain 10% of zirconium dioxide and lithium metasilicate and lithium disilicate crystals. ZLS ceramics are more recent, and they are comparable with the lithium disilicate glass ceramics [143].

5. Composite Resin and Polymers

Composite blocks can be used for CAD/CAM fabrication of inlays, onlays, and veneers. A popular block is Paradigm Z100 from 3M ESPE. This material is based on the Z100 composite from the same company. Paradigm Z100 has zirconia-silica filler particles and is 85% filled by weight with an average particle size of 0.6 μm. It is radiopaque and available in six shades, as well as a more translucent enamel color [144]. The advantage of this material over the ceramic blocks is its capacity to be more easily adjusted and polished intraorally.

3.2.2.7. Equipment

Below we discuss some examples of chairside CAD/CAM solutions and digital impression systems. The most popular chairside CAD/CAM systems are CEREC (Dentsply Sirona, York, PA) and Planmeca (Planmeca Oy, Helsinki, Finland). Carestream Dental (Atlanta, GA), Dental Wings (Montreal, Canada), and Zfx (Dachau, Germany) are available as chairside solutions systems [145].

1. CEREC System

The CEREC system was the first commercially available chairside CAD/CAM system and is currently the most popular one [135, 137]. This system was originally developed by Mörmann and Brandestini in 1985, and it was commercially under the name CEREC 1 [17, 19, 120, 146].

The currently available CEREC system includes the CEREC Omnicam scanner and CEREC MC, X, and XL. In 2012, Sirona unveiled Omnicam in which image capture is done via digital streaming and is in full color. The data collected by the scanner is processed by the CEREC software (New CEREC software 4.5). The true highlight of the CEREC Software is the "Biojaw" function. Based on the teeth scanned, the software generates a patient-specific restoration proposal taking into account the existing dental morphology. The software allows modification of morphology, occlusal contact adjustment, marginal detection and has a user interface that can be operated effectively. Once the designing step is completed, the production can be completed using the CEREC MC, MC, X or XL milling units (Figure 3.6). This system was originally developed for wet chairside milling, but the newer units offer the possibility of dry milling zirconia and lithium disilicate restorations chairside and also includes a sintering and glazing unit to finalize the restorations.

2. Planmeca System

The Planmeca system was introduced on the market in 2008 under the name of E4D and has undergone several reiterations. This system offers two intraoral scanning possibilities: the Planmeca Emerald and the Planmeca Planscan. The data collected is in STL open format allowing the possibility of using designing software and manufacturing form other systems. The captured data is then analyzed by the Planmenca PlanCAD which open CAD software is also. The software is easy and fast to use and is ideal for designing prosthetic works from a single crown to bridges. The process is divided into five steps from work description to milling. Once the designing step is completed, manufacturing can be done by the milling unit Planmeca PlanMill 40 S. For certain materials (i.e. Emax), the process of production needs to be completed in a sintering oven which needs to be purchased from a third party.

Figure 3.6. Photograph showing the CEREC system.

3.2.3. Laser Ablation

3.2.3.1. Overview of Laser Ablation

Laser ablation or laser milling is the process of removing material from a solid surface using a laser beam [147, 148]. The laser ablation milling system is similar to the traditional milling systems, but it uses the laser beam to remove the excess materials instead of cutting tools, such as burs. This technology is relatively new in dentistry, and it was introduced to dentistry by Dental Wings Inc in 2015 (Dental Wings Lasermill™). Laser ablation can be used to produce various dental restoration such as crowns, bridges, inlays, onlays, and veneers by milling a block of ceramic, polymer, or composite materials [147, 149].

3.2.3.2. Laser Ablation Process

The process of fabricating a 3D object by the laser ablation milling system starts with designing the 3D model on the scanned model using the computer-aided design (CAD) software [147]. After uploading the CAD file into the system, the laser ablation milling system removes materials from a block using millions of high-intensity laser pulses until the final shape is completed [147, 149]. Each laser pulse removes a small amount of material from the block by vaporizing the excess material. The spot size of the laser pulses is very small making the resolution of this system higher than any other traditional milling system [149]. Finally, the dental restoration is completed, and there is no need for secondary crystallization steps [147].

3.2.3.3. Advantages of Laser Ablation

High Precision and Quality

The laser ablation milling system is extremely precise, and it can mill crowns with high-resolution features. This is because the diameter of the laser beam is smaller than the diameter of the burs in the traditional milling systems at least by the factor of ten [147, 149]. The laser ablation milling system is also integrated with an in-process 3D scanner to achieve high-quality control during the milling process. In addition, this technique reduces some problems associated with the traditional milling systems such as chipping of thin edges [147].

Cost-Effective

The initial cost of the laser ablation milling equipment and materials is high. However, the overall cost of the laser ablation milling system is lower than the traditional milling systems due to low operating costs since the system does not use cutting tools, such as burs, which need to be replaced often due to breakage and wear [147, 149].

High Productivity

The laser ablation milling system is fast and comparable to the traditional milling machines [149]. Dental restorations can be finished on the same day using this milling system. Also, a wide variety of dental restorations materials can be used with this system [149].

3.2.3.4. Potential and Future Direction of the Laser Ablation

Although this technology relatively is new, it has the potential to become the main method for fabrication of dental restorations for its advantages over previous technologies.

3.3. Additive Manufacturing

3.3.1. 3D Printing

3.3.1.1. Overview of the Printing

3D printing or rapid prototyping (RP), which is also known as solid freeform fabrication (SFF), is a type of additive manufacturing that builds up 3D objects in a layer-by-layer pattern by laying down successive layers of material until the final object is formed [150, 151]. 3D printing technologies are growing and developing quickly, and they are used for different applications in various fields such as aerospace, automotive, engineering, jewelry, education, arts, architecture, and medicine [150]. The first 3D printing technology was developed in the 1980s and the first use of the 3D printing technology to treated patients in the late 1990s [19, 151]. However, 3D printing for dental applications is relatively new.

3.3.1.2. Dental Applications of 3D Printing

3D printing can be used for various dental application either directly by printing the final object in resin or metal, or indirectly by printing burn-out resins or waxes for subsequent casting process [151]. Direct applications of 3D printing technology in dentistry include fabrication of custom trays, temporary or definite crowns or bridges, and partial denture frameworks [126, 151-154]. Also, different orthodontic products can be fabricated by 3D printing such as positioning trays, orthodontic models, clear aligners retainers, bite splints, and night guards [151, 152]. Other applications for maxillofacial surgery and dental implants include surgical guides and maxillofacial prostheses [126, 151-153]. The indirect dental applications of the 3D printing include wax or resin castable pattern for crowns or bridges, partial denture frameworks, and complete dentures [151, 155].

3.3.1.3. 3D printing Production Process

The process of 3D printing can vary depending on the technique used, but it always follows similar concepts. A 3D model is created from data generated with a 3D or CT scanners. The object to be printed is designed in a computer-aided design (CAD) software and then using another CAD software, supports are added, and the model is sliced as multi-layers [19, 151, 152]. The 3D object is then printed, and for some 3D printing technologies, post-processing such as supports removal, heat treatment, and washing or polishing apply [151, 152].

3.3.1.4. Types of 3D Printing Technology

There is a large number of 3D printing technologies available for medical and dental applications including stereolithography (SLA), digital light projection (DLP), polyjet or multijet, inkjet printing, fused deposition modeling (FDM), and powder bed fusion (PBF) [19, 125, 150, 156]. The main differences between these techniques are in the materials used and the way the layers are deposited to create the 3D object. 3D printing technologies can be classified into three categories of liquid-based, powder-based, and solid-based depend on the form of the material used [150]. Each technique has its own advantages and drawbacks in term of accuracy, speed, costs, choice and cost of the materials, and color capabilities (Table 3.1). The main types of 3D printing technology are explained underneath including production process, characteristics, materials used, and dental application.

Stereolithography (SLA)

Stereolithography (SLA) is photopolymerization process that builds-up solid parts in multilayers from a liquid-based material using an ultraviolet (UV) light or laser for solidifying the materials [19, 125, 152, 153]. SLA was developed in 1986 by 3D systems, and it is considered to be the first commercially available 3D printing system [150]. SLA systems consist of a bath of

photosensitive liquid polymer monomer (e.g. acrylates and epoxy monomers), an ultraviolet (UV) light or laser, and building platform (Figure 3.7) [125, 153]. Objects are built in a layer-by-layer pattern (50-200 μm); at each layer, the UV light cures and hardens a thin layer of the polymer on specific areas defined by the CAD data, then the platform lowers or raises depending to the technology for the next layer, while the UV light cures the next layer with previous one [19, 153]. The process continues until the completion of the full object [152, 153]. Then, the object is removed from the bath [19, 152]. The post-processing treatment is applied to the final object including support structure removal. The object can be further cured in UV light or laser, and it can also involve surface treatments with primers, paints, or sealants to change surface roughness [152].

Another approach of SLA is digital light projection (DLP) that is similar to SLA, but the object builds upside down with different light source [151, 154]. DLP uses a projector light source that is applied to the entire surface of the photopolymer resin bath. This results in lower running costs and faster processing compared to SLA.

The accuracy of SLA is superior to other 3D printing techniques, and it can print complex geometries with fine details. A resolution of 5 μm in the X/Y axis and 10 μm in the Z-axis can be achieved by SLA [125, 151, 152]. However, this is influenced by many conditions such as the UV light parameters (wavelength, power, and exposure time), layer thickness, and step size [152]. Also, accuracy depends on the position of the object in the build platform that accuracy is high at the center than at the peripheral of the build platform [157]. One limitation is that SLA technology requires support structures to process objects, which increases the production time and consumes additional material [154]. In addition, SLA produces soft objects with limited mechanical strength [151, 152].

Typical dental materials used in SLA technology include acrylic resin, silicone, and epoxies [124, 125]. These materials are available in different colors and present different mechanical and physical properties [125]. These materials include poly (D, L-lactide) (PLLA), polyethylene glycol dimethacrylate (PEG-DMA), poly (propylene fumarate) (PPF), poly (trimethylene carbonate) (PTMC), and poly (methyl methacrylate) (PMMA) [124, 155]. In addition, ceramics can also be used with the SLA such as PLGA/ TCP (polylactic-co-glycolic acid (PLGA)/tricalcium phosphate (TCP), and alumina ceramics [124, 125, 158, 159]. Ceramic in SLA presented some issues with shrinkage but may be useful to be used as a scaffold for tissue regeneration [125].

One ideal dental application for SLA is for fabrication of dental models, surgical guides, and custom trays [19, 121]. Dental models for treatment planning or for educational purposes now can be produced by the SLA technology [19, 160]. The surgical guides that help for the placement of dental implants are commonly produced by the SLA technology [19, 161-166]. In addition, custom trays, temporary crowns and bridges, and prostheses pattern for lost wax casting process are produced by this technology [19, 164, 165, 167-169]. Definitive complete dentures have been fabricated successfully by the SLA technology using poly (methyl methacrylate) (PMMA) with TiO_2 nanoparticles [155]. Orthodontic appliances such as removable orthodontic appliances and occlusion ties were also produced successfully by this technology [170, 171]. Moreover, maxillofacial prostheses and facial replacements have been effectively printed by SLA technology [172, 173]. Scaffold for bone reconstruction using ceramic-based materials such as calcium phosphate hydroxyapatite and PLGA/TCP composite were also fabricated by this technology [124, 158, 159].

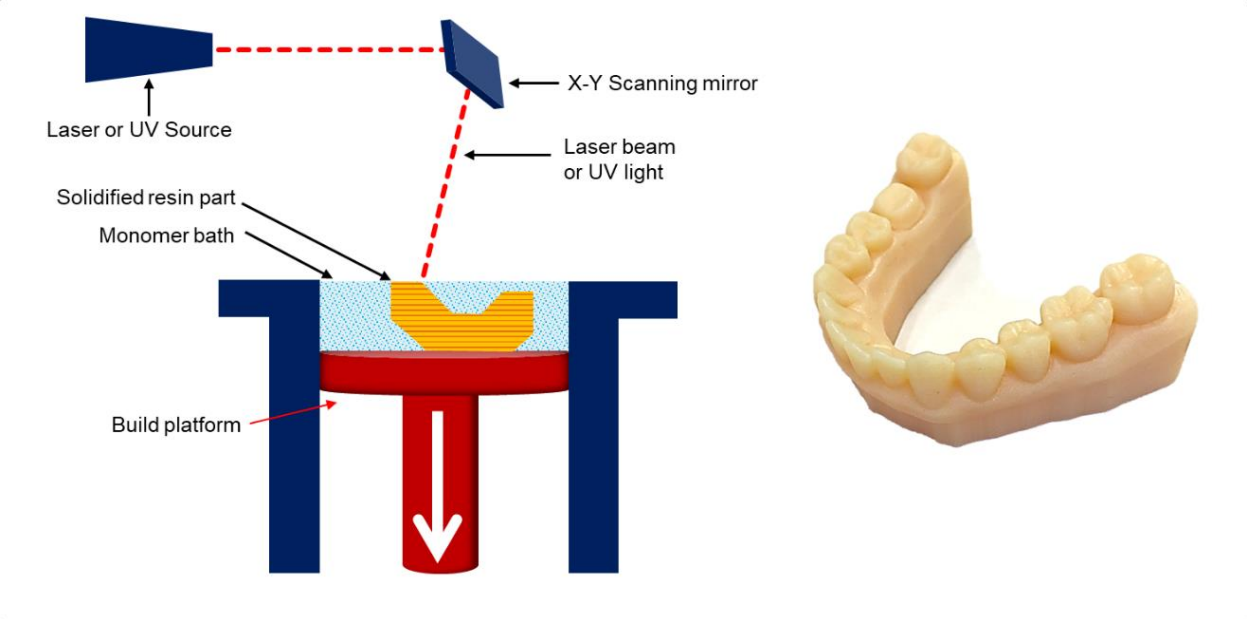


Figure 3.7. Schematic of the SLA production process and a photograph of a dental model printed by the SLA

Polyjet or Multijet

The polyjet or multijet printing (PJP or MJP) is a type 3D printing, which is similar to the 2D inkjet printing, but it builds up the object in multilayers [19, 151, 152]. This technology also can be referred to as photopolymer jetting (PPJ) [151]. With this method, droplets of photopolymer are ejected onto a surface and then cured by UV light (Figure 3.8). In each layer, liquid-based photopolymer materials apply only on the desired area and cured with the previous layers by the UV light [19, 151, 152]. This technique can combine multiple colors and materials in one print [19, 151, 152]. This is an important feature of the technology, for example, it can be used to print a mouth guard with hard and soft parts and with different colors [19]. This technology can print objects with complex geometry since it is possible to print objects with fine details at a resolution of 16 microns [151, 152]. Another advantage of this technology is the ability to use other materials such as wax or gel for the supporting structure for easier removal from the final object [152].

Different materials can be used for printing objects by polyjet or multijet technology include waxes, resins, and silicone [151]. Material jetting technologies are limited in dentistry because of their high cost compared to other less expensive 3D printing technology such as SLA [151]. This 3D printing technology can be used for processing many dental applications such as dental models, custom trays, surgical guides, temporary prosthesis, mouth guards, and orthodontic appliances [19, 151, 152, 174].

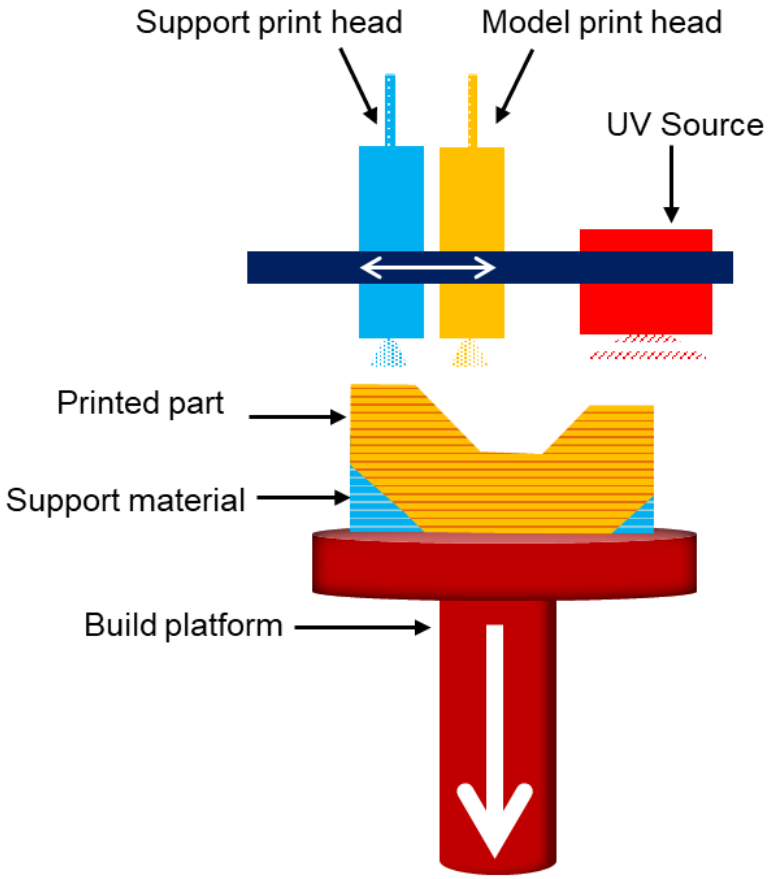


Figure 3.8. Schematic of the Polyjet/Multijet production process.

Inkjet Printing

Inkjet 3D printing or binder jetting process is a 3D printing process which an inkjet is used to eject small ink drops of binding liquid material toward a substrate of powder (plaster, ceramic,

or resins) and build up the object layer-by-layer (Figure 3.9) [153, 156]. The term 3D printing was introduced after the inkjet printing, and then it was subsequently used for all additive manufacturing methods. The process of inkjet printing starts with spreading a thin layer of the substrate powder across the binding platform, and a liquid binding material is applied on top of the powder, this connects together the exposed particles which leaving loose the unexposed particles [152, 153]. This process is repeated with each layer until the final shape is formed [152, 153, 156]. Finally, a heat treatment is applied, and the unbound powders are removed from the building platform [156]. Different colors of the liquid binding material can be used for printing multiple colors objects. The most common materials for this technology is plaster of Paris [151]. Ceramic suspensions were also used in some studies to print zirconia dental restorations [19]. Inkjet printing produces a lower-resolution print with achievable accuracy of ±127 µm, which is not ideal for dental applications, but it can be used for dental models and orthodontic diagnosis models [151-153, 156]. It has also been used experimentally to print bone graft materials [175].

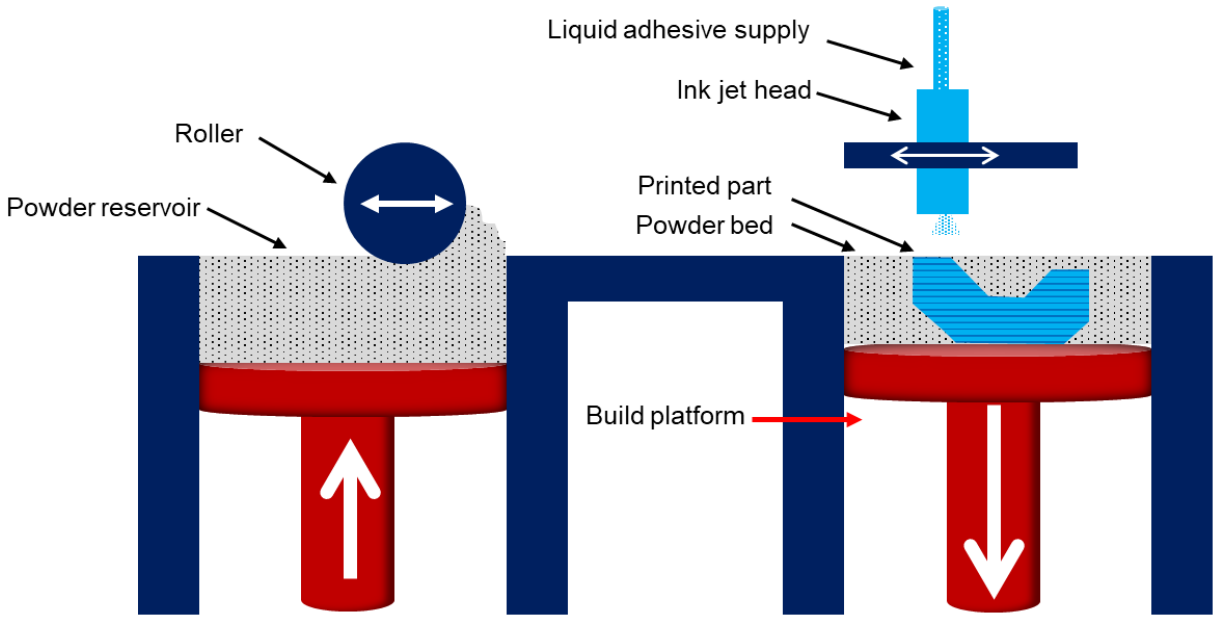


Figure 3.9. Schematic of the 3D Inkjet printing process.

Fused Deposition Modeling (FDM)

Fused deposition modeling (FDM) or fused filament fabrication (FFF) is a technique that builds-up an object by laying down a wire of thermoplastic material onto a building platform through a heated nozzle (Figure 3.10) [19, 151, 152]. This technique was developed in the early 1990s by Stratasys [154]. The 3D object is built from the bottom up, one layer at a time. The nozzle movement is directed by the CAM software and can be moved in both horizontal and vertical directions. The thermoplastic material is partially melted in the nozzle, and upon deposition on the building base, it solidifies immediately within 0.1 seconds [153]. The deposition process continues for the following layers until the final object is completely formed [19, 151, 153]. The layers of the deposited materials can be bonded together by the use of chemical agents or by temperature control [154]. A new approach of FDM such as Bioplotter was recently developed that is the ability to print in multiple materials including ceramic pastes which it can be used to print porous bone scaffolds and body parts [19].

The accuracy of FDM is lower than other 3D printing techniques such as SLA. The average accuracy of FDM is about ± 127 µm [18, 153, 156]. The accuracy of FDM depends on the speed of deposition, the flow of the material, material nozzle thickness, and the size of each layer [151, 154]. One advantage of FDM is no post-processing treatment is required. However, the low-resolution, slow speed, and low surface quality, are the main disadvantages of this technique [150]. The FDM is limited to thermoplastic materials for fabrication complex shapes and geometry. Several thermoplastic materials are available for this technology such as waxes, PLA (polylactic acid), polycarbonates, ABS (acrylonitrile butadiene styrene), PCL (polycaprolactone), and PPSF or PPSU (polyphenylsulfone) [18, 19, 151, 176]. PLA is more suitable to be used in dental application since it is more biocompatible than ABS [18]. In addition, the number of FDM

filaments options are increasing every year [124]. The ideal dental applications for FDM are custom trays, surgical guides, and wax patterns of dental prostheses for subsequent casting or polymerization process [19, 176].

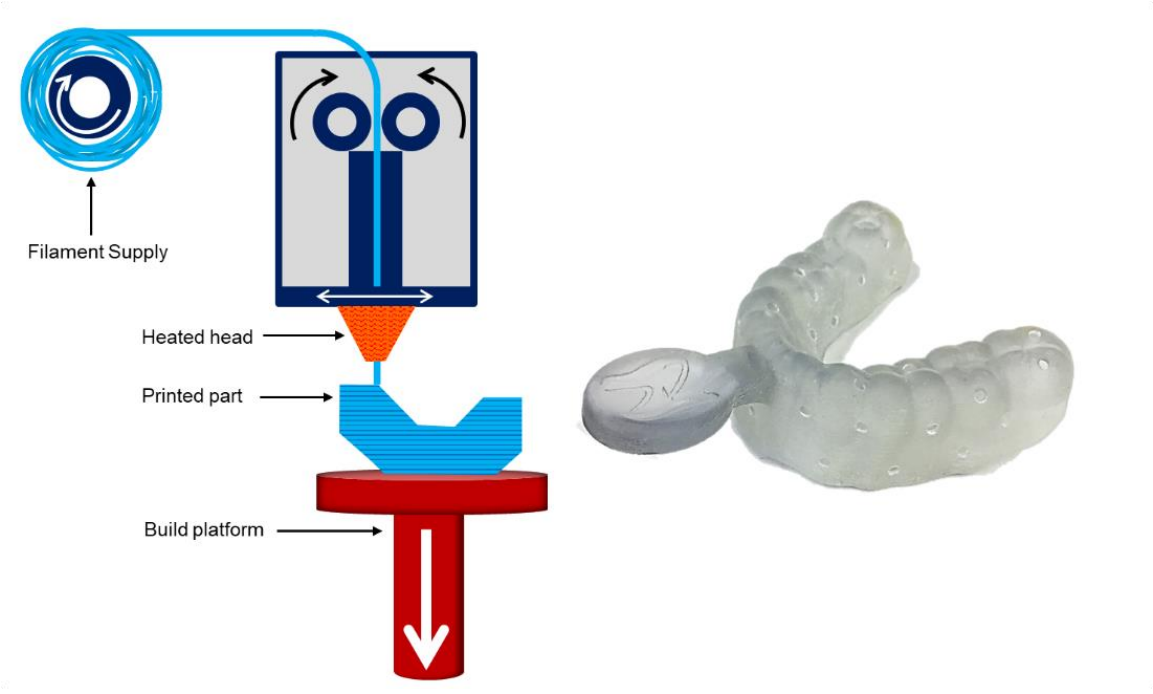


Figure 3.10. Schematic of the FDM/FFF production process and a photograph of a dental custom tray printed by the FDM/FFF.

Powder Bed Fusion (PBF)

Powder bed fusion (PBF) such as laser sintering and laser melting are an additive manufacturing technology used to process 3D objects in a layer-by-layer pattern using a high-power laser that melts or fuses successive layers of compacted powder [14]. To process the first layer of an object by laser melting technology, metal powders are spread onto a production platform by a counter-rotating roller [13, 19, 21]. Then, a laser beam is focused on an area defined by the CAD data file to fuse the powders in that area, while the remaining powders remain unfused [13, 14, 19, 21]. For the subsequent layers, the production platform is lowered for one-layer

thickness, a new layer of powders is applied again on top of the previous one, and the laser fuses the powders with the previous layer [13, 19, 21]. This procedure is repeated until forming the final desired shape. Laser melting are the newest technology in 3D printing, and it will be explained extensively in the next topic of this chapter [13].

Selective electron beam melting (SEBM) is similar to laser sintering and laser melting but the processing occurs in a high vacuum and with an electron beam as the heat source to fully melt the metal powder [19]. Another approach of SEBM is to use an electron beam to melt wire of metal onto a surface to build up an object that similar to the FDM technique but with metal rather than plastics [19]. One main advantage is the ability to produce porous objects by different alloys such as cobalt-chromium and titanium, and this technology can be used for producing customized implants for maxillofacial surgery [19, 177]. The accuracy of laser powder-forming technique such as SEBM can be about ± 20-50 μm [153].

3.3.1.5. Materials

Different materials can be printed by 3D printing technology, and these include polymers, metals, ceramics, and composites [121]. 3D printers in dentistry mainly use polymers as 3D printing material such as polypropylene, polyurethane, ABS (acrylonitrile butadiene styrene), PPSF (poly-phenyl sulfone), nylon, silicon, polystyrene, polylactic acid, polycarbonates, and polycaprolactone [153]. Some techniques allow the use of ceramic materials such as alumina ceramics and zirconia, while other technologies can use metals as the printing materials such as stainless steel, cobalt-chromium, and titanium [19, 151].

3.3.1.6. Equipment

Many manufacturers offer 3D printing for medical and dental application such as 3D Systems, Medical Modeling, EOS, BEGO, Stratasys, Materialise, and Formlab (Figure 3.11)

[153]. For instance, R.Pod® Desktop 3D printer (Arfona, Brooklyn, NY; arfona.com) and Perfactory Vida (EnvisionTEC, Dearborn, MI; envisiontec.com) are 3D printers based of fused deposition modeling (FDM), and they are able to print dental models, custom trays, and temporary prostheses using different materials with different colors such as nylon, PLA, ABS, TPU, and polyethylene. Moreover, Formlabs Form 2 (Formlab, Somerville, MA; formlabs.com), Objet Eden260VS Dental Advantage (Stratasys, Eden Prairie, MN; stratasys.com), and VARSEO 3D printer (Bego Medical, Bremen, Germany; bego.com) are 3D printers based on stereolithography (SLA), and they are effectively able to print dental models, surgical guides, custom trays, orthodontic appliances, and temporary prostheses. In addition, 3D Systems (3D Systems, Rock Hill, SC; 3dsystems.com) have many 3D printers for dental applications based on different technologies such as NextDent™ 5100 based on SLA technology, ProJet MJP 2500 based on poly/multiJet technology, and ProX DMP 100 and 200 Dental based on laser melting technology.

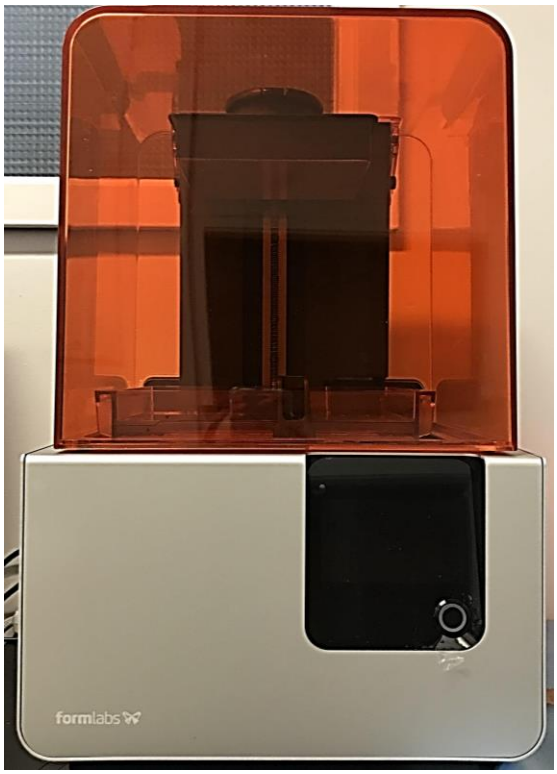


Figure 3.11.Photograph shows a 3D printing machine.

3.3.1.7. Advantages and Limitations of 3D Printing

There are some advantages and disadvantages associated with each 3D printing technique according to their accuracy, cost, strength, speed, availability, and choice of the materials. Generally, 3D printing technology is more economical and faster than traditional methods and milling systems [129]. These advantages and disadvantages are summarised in Table 3.1 [19, 121, 124, 150-153, 155, 158, 159, 174, 176-178]. It is important to know that 3D printing technologies are changing dramatically which can improve their quality and eliminate their limitations.

3.3.1.8. Potential and Future Direction of 3D Printing Technology

The accessibility of 3D printers has grown dramatically in the past decade [152]. Today, there are more than 300 companies selling 3D printers for general use and some 3D printers already cost less than $1000 [152]. Indeed, the 3D printing market has grown more than 33% in the last few years and was valued at $4.1billion in 2014 [152, 179]. In the next few years, the 3D printing market is expected to grow to over $8.9 billion, and the medical and dental application is comprising 21% of the market [126, 152].

Table 3.1. List of main materials, advantages, disadvantages, and dental applications with each dental 3D printing type.

Technique	Materials		Advantages	Disadvantages	Dental Applications
	Form	Type			
SLA	Liquid	Polymers: PLLA, PEG-DMA, PPF, PTMC, PMMA; ceramics, PLGA/ TCP, alumina	High accuracy, smooth surface, high density, low-cost materials	High-cost technology, limited strength, requires support structures, requires post-processing	Dental models, surgical guides, custom trays, temporary crown and bridge, prosthesis pattern, maxillofacial prosthesis, orthodontic prosthesis, and bone
PolyJet/ Multijet	Liquid	Waxes, resins, and silicone	High accuracy, variety of materials and colours, average cost technology.	High-cost materials	Dental models, custom trays, surgical guides, temporary prosthesis, mouth guards, and orthodontic appliances
Inkjet	Powder	Plaster of Paris and ceramic suspension.	Low-cost, and variety of materials and colors.	Low accuracy, low strength, and rough surfaces	Dental models, ceramic dental restoration, bone graft materials
FDM	Filament	Polymers: PLA, PC, ABS, PCL, PPSU, and waxes.	Low-cost, good strength, and variety of materials and colors.	Low accuracy and density, rough surfaces, slow speed, and limited to thermoplastic materials	Custom trays, surgical guides, and prosthesis patterns.
Powder Bed Fusion (PBF)	Powder	Metals: cobalt-chromium and titanium; ceramic; polymers	High accuracy, good strength, high productivity, low-cost materials.	High-cost technology, rough surface, and post-processing required	PRDP framework, crowns and bridge, and PFM coping, customized dental implants

SLA: Stereolithography; FDM: Fused deposition modeling; SEBM: Selective electron beam melting. PLLA: poly(D,L-lactide), PEG-DMA: polyethylene glycol dimethacrylate, PPF: poly(propylene fumarate, PTMC: poly(trimethylene carbonate), PMMA: poly(methyl methacrylate), PLGA/ TCP: poly lactic-co-glycolic acid and tricalcium phosphate; PLA: polylactic acid, PC: polycarbonates, ABS: acrylonitrile butadiene styrene, PCL: polycaprolactone, PPSU: polyphenylsulfone, PRDP: removable partials denture, PFM: porcelain fused to metal

3.3.2. Laser Melting

3.3.2.1. Overview of the Laser Melting Technology

Laser melting is an additive manufacturing technology used to process 3D objects in a layer-by-layer pattern using a high-power laser that melts or fuses successive layers of compacted powder [14]. Laser melting includes different technologies, such as laser melting, selective laser melting (SLM), selective laser sintering (SLS), or direct metal laser-sintering (DMLS) [14]. These technologies are generally referred to as powder-bed fusion (PBF) [180]. All of these technologies rely on the same concept, but they present some differences in the physical process or in the materials used [14]. Selective laser sintering (SLS) involves partial surface melting of the powder particles, and it was initially developed and patented in the mid-1980s for processing thermoplastic polymers [181-183]. The first 3D printed metal object was done in 1990, and this method was patented as selective laser sintering (SLS) [180]. With the development of powerful high-quality lasers, selective laser melting (SLM) and direct metal laser sintering (DMLS) technologies were introduced in 1995 to process metals [23, 181, 183]. The first commercial machine for processing metals by SLM was launched in 1995 by EOS GmbH [180]. SLM involves full melting of the powder particles while DMLS involves both full and partial melting of the powder particles [14, 23, 184]. Electron beam melting (EBM) is another PBF technology developed by Arcam in 2000 [180]. EBM is similar to SLS and SLM, but the processing occurs in a high vacuum, with a hot powder bed, and with an electron beam as the heat source to fully melt the metal powder [19, 185].

SLS techniques often process porous and weak objects, while DMLS and SLM can produce strong and dense objects [18]. SLS is used to process polymers and ceramics while SLM and DMLS are used for processing metal [14, 19, 184]. Nowadays, the systems used to process metal objects are commonly referred to as selective laser melting (SLM) because they rely on full melting

of the metal powder [180]. For this reason, the term laser melting technology will be used in this book chapter to refer to all metal powder-bed processes that use a laser as a heat source.

Laser melting technology involves the melting of powder material with a laser beam [187]. First, the building platform of laser melting machine is heated up to a temperature around ~200 °C and maintained at this temperature during the process [186]. Then, the laser beam is focused onto the powder bed to impart energy to the powder through photons and melt the metallic powder at a temperature between 500 to 1000°C [180]. Various laser parameters such as laser source, laser power, and wavelength can be adjusted to achieve an optimal powder melting [186]. The lasers used are often CO_2 lasers or fiber lasers (Nd: YAG or Yb: YAG) with a power of 200 to 300 Watt [187]. Nd: YAG crystal is a commonly used laser; while, Yb: YAG crystal is a new, and it has a larger absorption bandwidth, a lower thermal loading per unit pump power, and a longer upper-state lifetime than Nd: YAG [187]. Thus, Yb: YAG is expected to replace Nd: YAG [187].

The power of the laser, scanning speed, hatch spacing, and layer thickness are important parameters that can influence the powder melting process [186, 187]. For instance, low laser power, high scanning speed, and large layer thickness can result in insufficient energy to melt the powder [187]. Whereas, high laser and low scanning speed could lead to evaporation of the melted materials. Therefore, a suitable combination of the parameters is essential for successful processing an object by this technology. Also, poor hatch spacing can result in porosity in the processed object because the adjacent melt lines do not fuse together [187]. Therefore, a suitable combination of these parameters is crucial for processing a successful object [186, 187].

3.3.2.2. Dental Applications of the Laser Melting Technology

Laser melting technology in dentistry is currently associated with processing metals since other materials such as polymers, ceramic, and composite are more effectively produced by other CAD/CAM technologies. Thus, this chapter will be focused on metal. Laser melting technology used for different dental applications such as partial denture frameworks, dental crowns and bridges, dental implants, and maxillofacial prostheses [19]. Below we address the main dental applications for the laser melting technology in dentistry:

Removable Dentures

The metallic frameworks of partial removable dental prostheses (PRDPs) can be processed effectively using laser melting technology (Figure 3.12). Cobalt-chromium (Co-Cr) alloys processed by laser melting have shown superior mechanical and physical properties for partial removable dental prostheses (PRDPs) compared with the traditionally cast Co-Cr alloys [59]. Moreover, titanium alloy processed by laser melting technology presented a good quality for PRDP framework [188, 189]. In addition, a randomized controlled clinical trial showed that patients wearing laser melted PRDPs presented better outcomes in terms of patient satisfaction than those treated with conventional PRDPs [69]. Co-Cr and Ti alloys base plates for maxillary complete denture were also fabricated effectively by laser melting technology, and they were suitable for clinical use [190, 191].

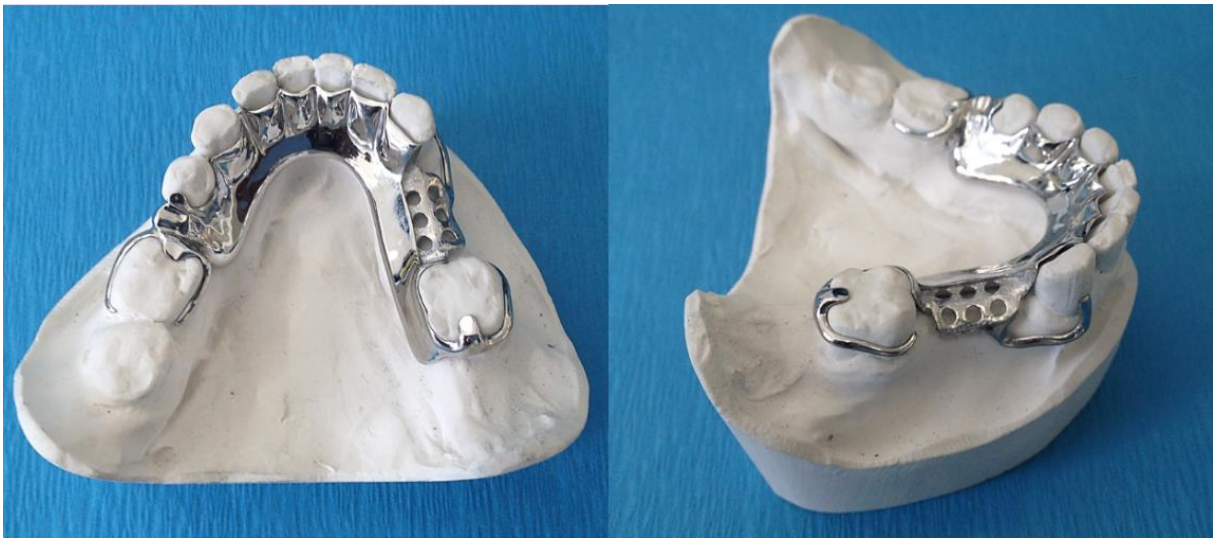


Figure 3.12. Photographs showing the metallic framework of partial removable dental prostheses (PRDPs) processed by the laser melting technology.

Fixed Partials Dentures

The metal copings for dental crowns and bridges can be successfully processed by laser melting technology, and the copings achieved a high internal fit and high marginal accuracy [14, 21, 192, 193]. In addition, the Co-Cr and Ti dental copings manufactured by laser melting technology have presented better mechanical properties and adhesion to ceramic coatings than the conventional cast Co-Cr alloys [194-199]. Clinical studies assessed the effectiveness of a metal-ceramic fixed dental prosthesis by laser melting technique, and they showed a high survival rate and promising results for clinical use [200, 201]. In addition, Co-Cr post-cores were fabricated effectively by laser melting technique [202].

Dental Implants

Dental root implants and implant prosthodontic framework can be produced by laser melting technology. This technology allows creating customized implants or implants with complex geometries opening the door for many promising clinical applications in the future [203-

206]. Moreover, many studies investigated that porous laser melting implants have improved osseointegration [207-210]. Implant prostheses and devices such as frameworks of implant-borne fixed dental prosthesis and bone extension device were successfully fabricated using laser melting technology, and they showed comparable results with conventional one [177, 211].

3.3.2.3. Materials

A large range of materials can be used in laser melting and laser sintering including polymers, ceramics, and metals [19]. Different types of polymer powder can be used in the laser sintering technology such as polyamides, PS (polystyrene), PC (polycarbonate), polypropylene, ABS (poly-acrylonitrile butadiene styrene), HDP (high-density polyethylene), and PEEK (polyether ether ketone) [151, 152]. In addition, ceramic materials such as HA (hydroxyapatite), tricalcium phosphate (TCP), and alumina (Al_2O_3–SiO_2) can be used in laser sintering [124, 212]. However, laser sintering polymers, composite, and ceramic are not yet widely used for dental applications because they can be produced by other 3D printing technologies more effectively and at a lower cost [14, 212].

Metals powders including cobalt-chromium (Co-Cr) alloys, titanium (Ti) alloys, and steel are the main materials used with the laser melting technique [188]. Co-Cr powders are commonly used for fabricating dental crowns and partial removable dental prostheses (PRDPs) frameworks, while titanium (Ti) powder has been used for dental implants and PRDP frameworks [189]. The quality of the powder that is used in the laser melting process determines the quality of the final product, and it is influenced by composition, size, shape, morphology, and amount of internal porosity [180]. Therefore, it is recommended to use a specified metallic powder for each laser melting system as each system is calibrated to suit its alloy. In fact, the chemical composition of the powder can affect the properties of the processed objects. Thus, it is important to measure the

elemental composition of recycled powder and remove any contamination from the powder to use it within their specification [180]. Moreover, smaller powder particles can improve the surface, but they are more costly than large size particles [180]. Therefore, the use of a fine distribution of powder particles can improve the surface finish and reduce the cost [180]. In addition, smooth particle surfaces produce less porosity, while the spherical powder particles tend to improve the apparent density. Table 3.2 shows a list of the main commercially available dental alloys for processing dental prostheses by the laser melting technology [14].

3.3.2.4. Equipment

Different laser melting machines are commercially available for processing metals for dental applications [14, 16, 24]. The main laser melting vendors in the market for medical devices include Phenix Systems (Figure 3.13), 3D Systems Corporation, EOS GmbH, GE, EnvisionTEC GmbH, Stratasys Ltd, Materialise, Renishaw, 3T RPD Ltd., Concept Laser GmbH, Arcam, Bio3D Technologies, Prodways, and Realizer. However, most of the previous studies in the past few years that tested laser-melted metals for dental applications were done by the three commercially available systems: EOSINT M250/M270/M280 (EOS GmbH, Munich, Germany), PM100/PXM (Phenix Systems, Riom, France), and Bego (Bego Medical, Bremen, Germany) [18]. Phenix systems (PM100 dental system) is the first laser melting system use cobalt-chromium powders for dental applications [150]. In the past few years, there were some changes in this industry that 3D Systems bought Phenix Systems, while GM manufacturer bought two systems that are Arcam and SLM Solutions. Table 3.2 shows a list of commercially available equipment that can be used for processing dental prostheses.

Table 3.2. List of commonly available equipment and materials for laser melting that can be used for processing dental prostheses.

Technology			Alloys	
Equipment	**Type**	**Manufacturer**	**Type (brand name: composition)**	**Suppliers**
EOSINT M250	SLM	EOS, Munich, Germany	Co-Cr (SP2: Co 52, Cr 24, Mo 6, W 6, S, Fe, Mn <2; MP1: Co 60-65, Cr 26-30, Mo 5-7, Si, Mn, Fe, C, Ni <2); Ti (TiCP: Pure titanium)	EOS, Munich, Germany
EOSINT M270	SLM	EOS, Munich, Germany	Co-Cr (SP2: Co 52, Cr 24, Mo 6, W 6, S, Fe, Mn <2; MP1: Co 60-65, Cr 26-30, Mo 5-7, Si, Mn, Fe, C, Ni <2); Ti (TiCP: Pure titanium)	EOS, Munich, Germany
PM 100 Dental System	DMLM	Phenix Systems, Clermont-Ferrard, France	Co-Cr (ST2724G: Co balance, Cr 29, Mo 6, Mn, Si, Fe <1)	Sint-Tech, Clermont-Ferrard, France
PM 200 Dental System	DMLM	Phenix Systems, Clermont-Ferrard, France	Co-Cr (ST2724G: Co balance, Cr 29, Mo 6, Mn, Si, Fe <1)	Sint-Tech, Clermont-Ferrard, France
SLM	SLM	Bego Medical, Bremen, Germany	Co-Cr (Wirobond C+: Co 64, Cr 25, W 5, Mo 5, Si 1)	Bego Medical, Bremen, Germany
Laser CUSING	SLM	Concept Laser GmbH, Lichtenfels, Germany	Co-Cr (Remanium Star: Co 60, Cr 28, W 3, Si 2; Mn, N, Nb, Fe < 1)	Dentaurum, Ispringen, Germany
SLM 50	SLM	Realizer GmbH, Borchen, Germany	Co-Cr (Solibond C plus Powder: Co 63, Cr 24, W 8, Mo 3, Nb 1, Si 1)	Yeti Dental; Engen, Germany
SLM 125	SLM	SLM solution GmbH, Lubeck, Germany	Co-Cr; Ti	SLM solution GmbH, Lubeck, Germany
SLM 280	SLM	SLM solution GmbH, Lubeck, Germany	Co-Cr; Ti	SLM solution GmbH, Lubeck, Germany

SLM: selective lase melting; DMLM: direct metal laser melting.

Figure 3.13. Photograph shows a laser melting machine.

3.3.2.5. Laser Melting Production Process

The first step of processing an object by the laser melting technology starts by designing the 3D object on the scanned model using a computer-aided design (CAD) system. Then, special CAD software is used to slice the designed 3D object (STL file) into multiple layers with a defined thickness and to add supports between the model and the production platform [14]. The supports are added to prevent the collapse of the build materials [156]. After uploading the design file into the laser melting system, the production process starts with spreading a thin layer of alloy powder onto a production platform with an accurate thickness of 20-100 μm and powder particle size of 25-45 μm (Figure 3.14) [13, 14, 19, 21]. Then, the directed laser beam fuses or melts the powder only at a specified site defined by the CAD data file, while the remaining powder particles remain

unfused [13, 14, 19, 21]. For the subsequent layer, the production platform moves down a distance of one-layer thickness, and a new layer of powder is applied again on top of the previous one, and the laser fuses or melts the powder with the previous layer [13, 14, 19, 21]. This procedure continues, layer by layer, until object completion (Figure 3.15). It should be noted that it is important to select the proper processing parameter (e.g. scanning rate, laser power, and layer thickness) for each dental material and application since these parameters can change the properties of the processed objects (e.g. accuracy, density, surface roughness, hardness, and strength) [184, 194]. Also, the build orientation can change the mechanical, physical properties of the object which should be considered during the processing [64, 213, 214].

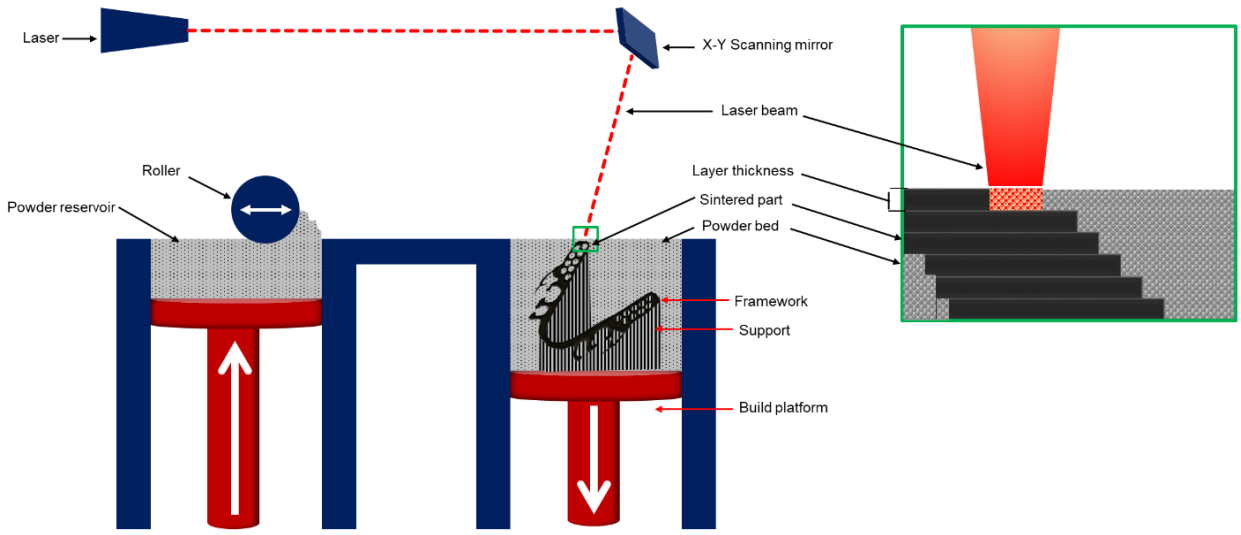


Figure 3.14. Schematic of the laser melting production process for an PRDP framework.

Figure 3.15. Photographs showing the final processed frameworks of removable partials dentures (PRDPs) by laser melting.

3.3.2.6. Post-processing Process

The post-processing process is usually required to improve the properties of the final 3D object, and this involves the following steps [14]. First, the loose powder particles are cleaned from the processed object to remove any unsintered powder sticking to the surface [180]. Next, the support structure that connects the 3D object with the production platform is removed [180]. Then, heat (thermal) treatment is usually applied to the final 3D object according to the manufacturers' instructions for a period of time to enhance the mechanical and microstructural properties [14, 215]. The thermal post-processing is used to relieve residual stress and to improve the mechanical properties of the metals, and it has very important effects on the grain structure of the processed material [180]. The heat treatment for alloys is usually done at a temperature of 800-450ºC for 30-60 minutes in different stages [59]. For examples, post-processing heat treatments for Co-Cr alloy is applied in 3 stages. The object is heated at 450 ºC for 45 minutes, at 750 ºC for 60 minutes, and then cooled down fast. Post-processing heat treatments for Ti alloy is applied in

3 stages, the object is heated at 750 ºC for 2 hours, at 900 ºC for 2 hours, and then cooled down fast [180]. Finally, the surface of the final metallic objects involves different finishing and polishing steps (such as electropolishing) before sending them to the clinic.

3.3.2.7. Advantages of Laser Melting Technology

Laser melting technology is a very suitable technique for processing dental prostheses because it is accurate, fast, and cost-effective, and it can improve the quality of dental prostheses and the productivity of dental laboratories [14, 19, 23]. In addition, a vast variety of dental alloys can be used for dental applications. Underneath are the main advantages of laser melting technology.

High Accuracy and Quality

The accuracy of laser melting is extremely high; this technology is able to fabricate 3D objects with an accuracy of ± 20μm [59, 156, 194, 216]. The minimum feature size that can be printed is 75–100 μm [180]. However, the accuracy depends on the processing parameters, building direction, and the geometry of the objects [59, 194, 216]. The laser melting technology enables producing a complex 3D design and geometries, unlike the subtractive manufacture techniques. Compared to traditional casting technique, one major advantage of laser melting is the ability to produce objects that are more homogenous microstructure (Figure 3.16) [59]. As a result of this, cobalt-chromium (Co-Cr) objects processed by laser melting present better fatigue resistance and physical properties than Co-Cr produced by the traditional casting method [59, 69]. Also, many studies showed that Co-Cr and Ti alloys produced by the laser melting have better or comparable biocompatibility and lower ions releases than with cast alloys [14, 16, 24, 59, 217, 218]. Clinical studies have also shown that the high precision and quality of alloys processed by

this technique might improve the quality of the provided dental prostheses and therefore increase patient satisfaction with their dentures [69].

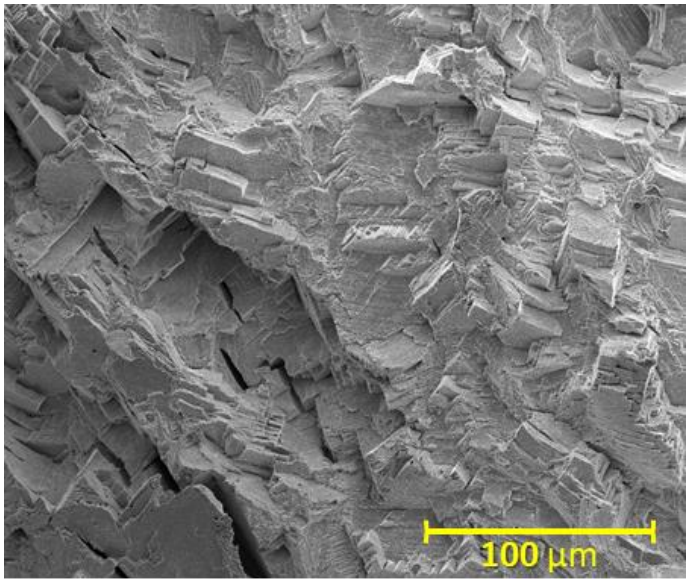


Figure 3.16. Scanning electron microscope (SEM) images showing a homogenous and organized fracture path of the Co-Cr alloys processed by the laser melting technology.

High Productivity

The production speed of laser melting devices is proportional to the size of the objects as well as other processing parameters such as scan speed, scan space, and layer thickness [219]. In the case of the fabrication of dental prostheses, laser melting usually take less than 12 hours which is faster than the time needed to fabricate prostheses by the traditional casting technique, as it reduces the fabrication steps (e.g. waxing up, molding, firing, casting, etc.) into one step [14]. Also, during the manufacturing process, multiple dental prostheses can be processed simultaneously on the same production platform which considerably increases productivity. For instance, one laser melting system can produce around 450 units of dental crowns and bridges within a day [220]. In fact, this technique can speed up the denture delivery, as it enables to finish processing the framework within one day [19].

Cost-Effectiveness

The overall cost of the dental prostheses processed by the laser melting technique is lower than processing by milling or casting techniques [14]. The reduced cost is a result of low labor, time, waste materials, and cost of the materials as well as the ability to recycle the unused materials [14]. It has been estimated that the fabrication of dental prostheses by laser melting technology can reduce manufacturing costs down to less than half the cost of traditional techniques [221].

3.3.2.8. Limitations of the Laser Melting Technology

There are some disadvantages of laser melting technology. The initial cost of laser melting equipment is relatively high [14, 127]. Also, most of the laser melting methods require post-processing treatments for the objects including heat treatment to improve their mechanical properties and support structure removal which may delay the processing time [124, 151]. Other limitations are the staircase effect and surface roughness, which may appear due to the layering nature of the process; however, they can be minimized by reducing the layer thickness of the object [18, 222, 223]. Although laser melting was successful for processing the root parts of dental implants, the accuracy of laser melting is not accurate enough to process dental implants connection parts.

3.3.2.9. Potential and Future Direction of the Laser Melting Technology

Laser melting technology is a very promising technology, and its market is growing rapidly as the manufacturing process improves and the costs keep falling. Manufacturers are expanding rapidly to fulfill the growing demand for this technology for industrial, medical, and dental applications. For instance, in 2016, General Electric (GE) bought two 3D printing groups, Sweden's Arcam and Germany's SLM Solutions, for a total of $1.4 billion, and in 2013, 3D-Systems acquired the French company Phenix [224, 225]. As a result of this competition, the

mechanical properties, precision, and production speed of laser melting technology are expected to be further improved in the future. Moreover, the price of the laser melting machine is expected to decrease drastically by the market competition, especially as the patents of the technology expire in the nearby future. Besides its proven potential for PRDPs, oral and maxillofacial prostheses are also produced by this technology, and the future developments on this technology could render it more competitive over current CAD/CAM subtractive technologies for manufacturing dental crowns, bridges, and implant prosthodontics.

Beyond its impact on dentistry, this technique will also have an impact on society in the next few years. First, the reduced cost of dental prostheses processed by laser melting technology could render the treatment less expensive and more accessible to a larger portion of the population [14]. Large and small dental laboratories both can benefit economically from using this technology through new forms of business models; however, large-scale dental laboratories are at an advantage over smaller laboratories because the initial cost of the equipment can only be amortized across the large-scale production [14]. Instead, small dental laboratories and dental offices can benefit economically if they design the dental prosthesis in CAD file, as only requires a scanner and CAD system, and outsource the fabrication of the prosthesis framework to local processing centers.

3.4. Conclusion

CAD/CAM (computer-aided design/computer-aided manufacture) technology is rapidly growing and changing dentistry at an unprecedented pace. Dental CAD/CAM is now used for an ever-growing number of dental applications such as the fabrication custom trays, surgical guides,

temporary or definite fixed or removable dental prostheses, and orthodontic and maxillofacial appliances.

CAD/CAM technologies available in dentistry can be classified as either “subtractive” or “additive” manufacturing methods. With subtractive methods such as machining and milling and laser ablation technologies, dental parts are manufactured by subtracting the undesired material from a block with the use of burs, disks, or lasers. The CAD/CAM systems for subtractive manufacturing methods can be classified into chairside systems and laboratory systems. Additive methods, such as 3D printing or rapid prototyping, manufacture dental objects in a layer-by-layer pattern by building successive layers of material until the final object is formed. There are many 3D printing technologies available for dental applications such as stereolithography (SLA), digital light projection (DLP), polyjet or multijet, inkjet printing, fused deposition modeling (FDM), selective electron beam melting (SEBM), and laser melting.

Additive manufacturing is a more recent technology and more versatile than subtractive manufacturing, but the subtractive methods are more precise and accurate. Thus, each of these technologies is used for different dental applications according to the accuracy, speed, costs, and materials required.

Chapter 4: Characterization and Testing Methods

This chapter details the main characterization and testing methods that were performed in this thesis in terms of their history, concept, and procedures. According to the objectives of the thesis, we selected these tests in order to characterize the mechanical, physical properties, and biocompatibility assessments. More details about the parameters that were used in this project for the technique will be described in the material and methods section of the manuscripts.

4.1. Mechanical Properties

Mechanical properties of materials are an important characteristic that helps to identify and classify the materials and define their possible applications. The definitions of the mechanical properties assessed in this thesis are as follows:

- Strength: is the ability a material to withstand the mechanical loads applied to it without deformation or fracture [226].
- Yield strength: is the stress at which the plastic deformation of the material begins [226].
- Elasticity: is the capacity of a material to return to its original shape after a mechanical load is removed [226].
- Plasticity: is the ability of a material to withstand permanent deformation without failure [226].
- Young's modulus or elastic modulus is the ability of a material to withstand changes in length when the material is under tension or compression, and it the relationship between stress and strain in a material in the elastic region [226].
- Toughness: is the ability of a material to absorb energy and withstand shock loading or impact before fracture, and it represents a combination of ductility and strength.
- Ductility: is the ability of a material to undergo plastic deformation without breaking.

- Fracture toughness: is the ability of a material to resist fracture and containing crack propagation. The stress-intensity factor (K) is the parameter to determine the fracture toughness of the material.
- Hardness: is the ability of a material to resists scratching and plastic deformation [226, 227].
- Fatigue limit: is the highest stress that a material can resist without failure after a given number of cycles.

4.1.1. Mechanical Tests

Measuring the mechanical properties such as strength, hardness, toughness, and stiffness of a material can be performed through different testing methods [227]. For many mechanical tests, a force is usually applied to a testing specimen of a material resulting in stress and associated deformation (strain) on the specimen shape [226, 228]. The relation between the force applied to the materials and the resulting strain is used to calculate several mechanical properties (e.g. ultimate strength, yield strength) [226, 227]. The force can be applied to the specimen in several different ways, and each one represents a testing method such as tension (tensile), compression, bending, twisting, and shear [226, 227]. Each testing method has its equations to calculate the mechanical properties of the material. Underneath is an overview of mechanical testing methods that were performed in this thesis.

4.1.2. Flexure Test

Bending or flexure test is another testing method that involves applying a vertical load onto a rectangular specimen that is supported on two supports (Figure 4.1). This method is a useful testing method for brittle materials or when it is difficult to obtain a proper specimen for the tensile test. This test can be referred to a three-point bending test when the vertical loads applied at one point, and a four-point bending test when the vertical load is applied at two points [227]. In

addition, the bending test for a pre-cracked specimen is suitable for measuring the fracture toughness of material [226]. From the stress-strain curves obtained from the bending test, different properties can be calculated such as the flexural strength, flexural modulus (Young's modulus), yield strength, and fracture toughness [227].

In this thesis, the three-point bending test is the main test that was used for testing the mechanical strength of the alloys because that this testing method bends the specimen in a similar way to how dental clasp bend upon abutment engagement. Specimens for three-point bending test were small rectangular prisms (20, 4, and 2 mm) and specimens for fracture toughness analysis was rectangular prisms (40, 4, and 2 mm) with a single notch (2, 2, and 2 mm). The geometries of the specimens are showing in Figure 4.2. Three-point bending tests were done at room temperature using a universal testing machine (Instron, 5569, Grove City, PA) (Figure 4.3) to characterize the mechanical properties of the alloys. Each specimen was placed on two supporting pins 18 mm apart of each other (Figure 4.1). Loading was applied through an actuator by moving the loading pin at a constant speed of 1 mm/min on the middle of the specimen until failure.

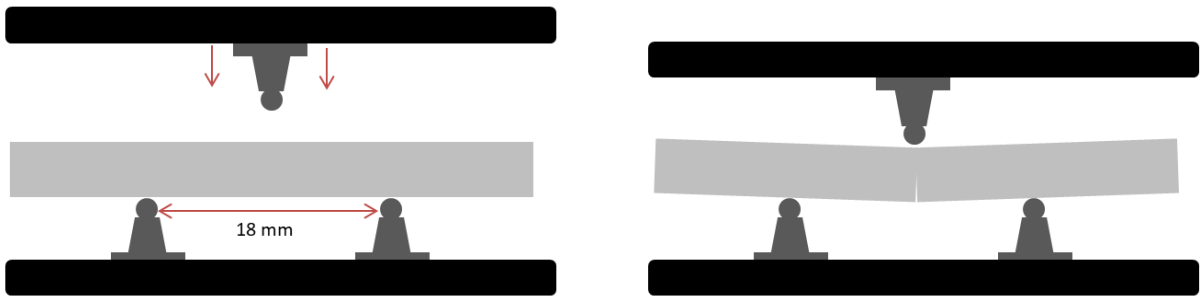


Figure 4.1. Drawing shows the mechanism of the 3-point bending test.

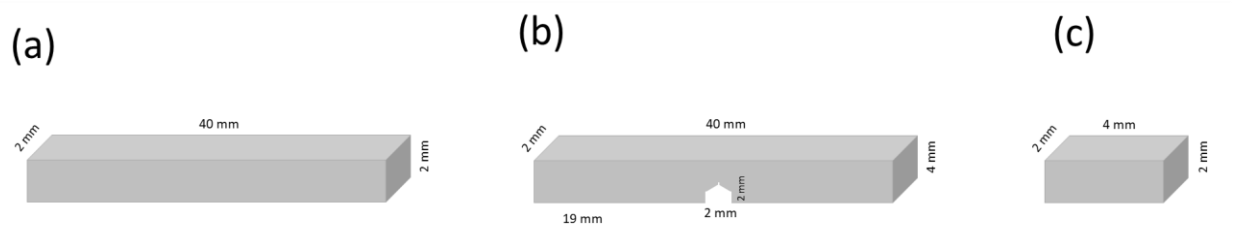


Figure 4.2. Specimen geometries (a) for the 3-point bending test and fatigue test, (b) for the fracture toughness test, and (c) for the *porosity, crystallinity, and biocompatibility tests.*

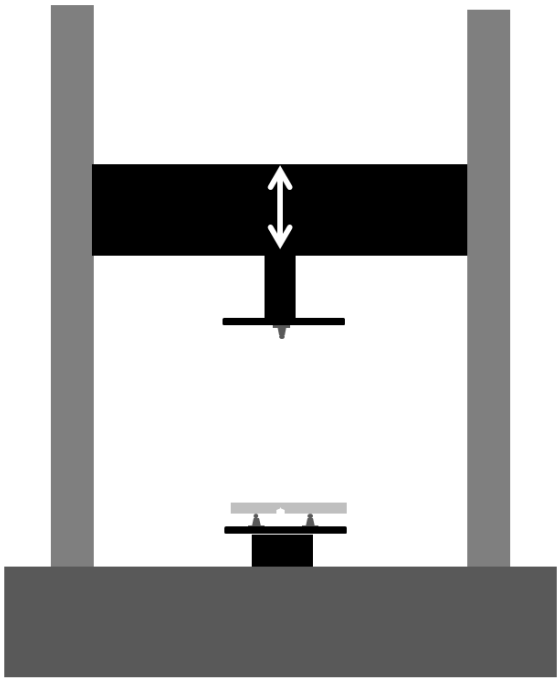


Figure 4.3. Schematic showing the universal testing machine.

4.1.3. Fatigue Tests

Cyclic stresses might change the mechanical properties of materials, this is called fatigue [227]. Fatigue can result from both tension or compression cyclic stresses lower than the yield stress of the material. Thus, a fatigue test usually applies cyclical stresses (e.g. tension and compression) to a specimen until failure [226]. The fatigue test is carried out at various stresses

(S) and a number of cycles (N) [227]. Then, the results are represented in a plot called the S-N curves [227]. A cyclic fatigue test is another approach of fatigue testing that simulates the clinical use of a material, and it calculates the stress change between the minimum and the maximum stress applied on the testing sample [95]. The fatigue test performed in this thesis was done based on this approach. The specimen is exposed to a cyclic of three-point bending of loading and unloading up for many cycles (e.g. 6000 cycles) until reaching a specific displacement (e.g. 0.2 mm). Then, the post-fatigue force is compared to its initial force. The geometries of the specimens and testing setup for the fatigue cyclic test are showing in Figure 4.2 and Figure 4.1.

4.1.4. Hardness Tests

Hardness tests provide information on the strength of a material and of its resistance to scratching and deformation [226, 227]. Hardness measurements can be done at different scales such as macro-hardness, micro-hardness or nano-hardness. Large hardness value indicates higher resistance to plastic deformation or cracking and better wear properties. One common method for testing the hardness is using the hardness test developed by Vickers in 1922 [227]. Vickers hardness test is a microhardness test method that applies a pyramid-shaped diamond indenter onto a material using a static load (range from 10 gm to 1 kg) causing a permanent indentation on the surface of the material. The size of this indentation represents that of the hardness of the materials. The indentations in Vickers hardness tests are small and often known to be measured with a microscope. Then, the hardness value is calculated according to the size of the indenter, the applied load, and the size and depth of indentation. The hardness is calculated as $2P/\pi D(d-\sqrt{D2-d2})$. Where P is the applied force; D is the diameter of the indenter, and d is the diameter of indentation. The Vickers hardness value is indicated by HV [226, 227].

In this thesis, specimens for microhardness (20, 4, and 2 mm) were manually polished to mirror-like in a six-step polishing process. The specimens were grounded and polished with 240-600 grit abrasive grinding papers (Paper C-Wt, AA Abrasives, Philadelphia, PA) and polishing cloths (15-0.02μm and 1-0.02μm; Buchler, Whitby, ON) using colloidal silica (≤ 0.06 μm; MasterMet: Buchler, Whitby, ON).

4.1.5. Testing Machine

A universal testing machine (UTM) is a common instrument used to measure the stress and strain of a material. It is composed of a load frame, load cell, crosshead, and an output device connected to a computer. The system that was used to test the mechanical proprieties of materials in this thesis is the electromechanical system from Instron Inc. (Instron, 5569, Grove City, PA) equipped with a 50 kN load cell (Figure 4.3). The 3-point bending tests were performed on a fixture made for the 3-point bending test attached to the Instron machine.

4.2. Physical Properties

Physical properties of a material are used to observe and describe a state of the physical system of a material. Physical properties of metals include density, porosity, crystallinity, and microstructure. Underneath is an overview of the methods that were used to characterize the physical properties of materials in this thesis.

4.2.1. Density

Density is an important property that affects the mechanical property of the materials. Density (ρ) is the mass (m) of an object divided by its volume (v), and it is often defined as grams per cubic centimeter (g/cm^3) [229]. Density can be bulk, true (real), or apparent. Bulk density is the density of a solid material including the volume of any closed and open pore. True (real) density

is the density of solid material excluding the volume of any closed and open pore. Apparent density is similar to the true (real) density except for the volume of the closed pore is also included.

All density measurements involve the measurement of mass and volume. The bulk density can be calculated simply by dividing the sample's weight by its volume ($\rho=m/v$); the weight can be measured with a balance while the volume can be calculated with an electronic caliper [229]. The true (real) and apparent density require to measure the correct and precise volume, which do not include the pores. This can be measured using a Pycnometry [229].

Pycnometry is a device used for measuring the apparent volume of a solid or powder object. The displacement fluid may be a liquid, or a gas (gas pycnometry). Helium is the commonly used gas because the size of its atoms is small that is easy to penetrate the (open) pores of the sample. Helium Pycnometry is a non-destructive technique, and it measures the gas pressure in a calibrated chamber before and after insertion of the object into the chamber. Thus, the Helium Pycnometry can be used to precisely measure the apparent density excluding any open pore spaces in the object. In addition, the Helium Pycnometry is a useful device to calculate the total porosity (open porosity) of a solid material.

4.2.2. Porosity

Porosity refers to the voids or spaces within a material, and it is presented as the ratio of the volume of voids and spaces over the total volume of the material. It is usually expressed as a percentage between 0% and 100%. The total porosity (open porosity) can be calculated from a measure of both bulk and apparent density on the same materials which the apparent density can be measured using Helium (gas) Pycnometry as described earlier. From the different density and volume, the open porosity percentage can be calculated using the equation of $Porosity = 1 - (\rho\ bulk/\ \rho\ grain$ [229].

4.2.3. Microcomputed Tomography

Microcomputed tomography (micro-CT or μ-CT) is a non-destructive technique that provides high-resolution three-dimensional (3D) images of the internal and external structure of an object without destroying the original object [230, 231]. Micro-CT produces X-ray imaging in 3D similar to the CT scans used in hospitals but on a smaller scale and at a higher resolution. The micro-CT was developed in the early 1980s by Less Feldkamp [230]. The first realization of a micro-CT system was presented in 1982 by Elliott and Dover [232].

The micro-CT equipment is composed of a specimen stand, an x-ray source (e.g. micro-focus x-ray tube), a radiation filter and a collimator, and a detector (e.g. charge-coupled device (CCD) camera) [230, 231]. During the process, the micro-focus x-ray source irradiates the object and pass through the sample, while the sample is rotating in order to capture x-ray images from multiple angles. In the end, the X-ray detector collects the magnified projection images. These images contain information about the density of the material that a higher density material will have greater X-ray absorption. Then, these images are used to generate μ-CT 3D images. In fact, the 3D images produced by the μ-CT are composed of a series of 2D two-dimensional (2D) slices images of the object [230].

The procedures for the micro-CT analyzes can be divided into 3 steps. The first step is scanning the object to obtain the x-ray projection images. The micro-focus x-ray beam penetrates and irradiates the object while the detector collects magnified x-ray projection represented in 2D images. The second step is reconstructing the projection 2D images into 3D images using computer software. The final step is analyzing the 3D image using computer software (e.g. volume and porosity) [231]. Micro-CT scanning parameters can be adjusted according to the equipment,

sample, and what needs to be evaluated [231]. The quality of the µ-CT image depends on the scanning resolution that set according to the area and size of structures within the object [231].

The micro-computed tomography (µ-CT) that was used in the project was the high-resolution Bruker µ-CT (1172; SkyScan; Kontich, Belgium) (Figure 4.4). NRecon v.1.6.9, SkyScan is the computer software for data reconstruction. The 3D modeling and analyzing were performed using the software CTAn v.1.13 (SkyScan Kontich, Belgium) while the 3D imaging was done with the software of CTvol v.2.2.3 (SkyScan Kontich, Belgium). Specimens for the µ-CT test were square prisms (4, 4, and 2 mm) (Figure 4.2).

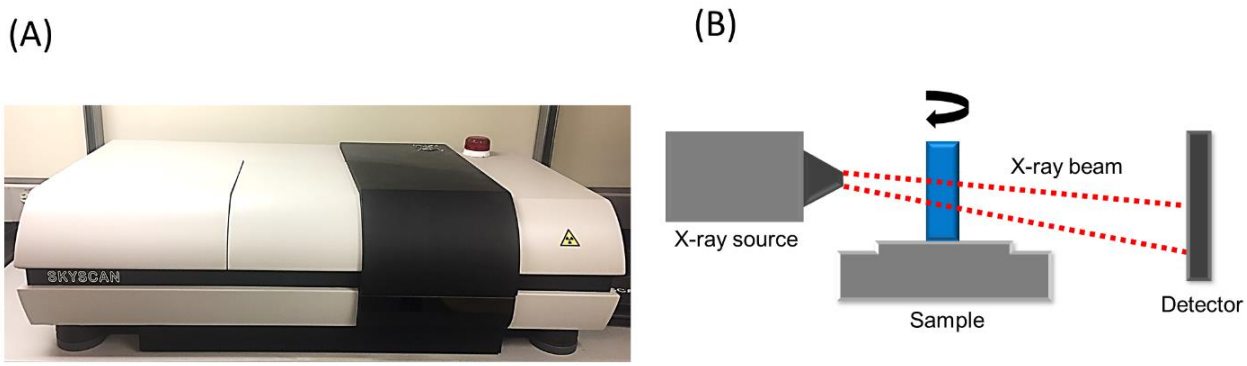


Figure 4.4. (A) the microcomputed tomography (µ-CT) machine, (B) diagram shows the mechanism and the testing sample of the µ-CT test.

4.2.4. X-Ray Diffraction

X-ray diffraction (XRD) or X-ray crystallography is a non-destructive technique used to determine the atomic and molecular structure of crystalline material in the solid state [233, 234]. XRD provides information on crystal structure, phase, crystal orientation, and grain size. Also, it is used for quantitative analysis and fingerprinting of materials due to its sensitivity to chemical changes [234]. The X-ray is electromagnetic radiation of shortwave (0.01 to 10 nm) and high energy (100 eV to 100 KeV) wavelength that is able to pass through a solid material.

X-ray diffraction (XRD) equipment consists of an X-ray tube, a sample stage, and a detector connected to the computer (Figure 4.5). After placing the sample on its position, the X-ray tube projects the X-ray beam on the sample, and the beam passes through the solid sample. The diffracted beam is then recorded by the detector while the detector sends the diffraction data to the computer generating the XRD pattern.

The crystal structure of a material is being built of planes, or layers in specific arrangement and space between them. X-ray with a wavelength similar to the distances between these planes can be reflected and this behavior called diffraction. This behavior was described by Bragg's Law that $2d\sin\theta = n\lambda$ [233, 234]. Where d is the spacing between diffracting planes, θ is the incident angle, n is an integer, and λ is the wavelength of the beam [233, 234]. Using Bragg's Law, the reflection of the diffracted X-ray beams will be recorded by a detector. The positions of these reflections give information about the interlayer spacings of atoms in the crystal structure according to Bragg´s Law. Also, the intensities of the reflections give information about how much X-ray scattering is contributing to that reflection, or how much of a phase is presented in the sample [233, 234].

The X-ray diffraction pattern is produced as a result of the interaction between the incoming electrons and the inner shell electrons of the target element. The peak intensities are related to the distribution of atoms within the lattice while the number of peaks is related to the symmetry of the unit cells [233, 234]. Finally, the XRD pattern is identified and matched with a pattern from a database of ICDD (International Centre for Diffraction Data).

The X-ray diffraction (XRD) equipment used in the project was the D8-Discover/GADDS, Bruker, Karlsruhe, Germany with a cobalt x-ray source of radiation (Figure 4.5). The EVA v.14 software (Bruker AXS, Karlsruhe, Germany) was used for phase identification and crystal size

calculations following Scherrer's formula. Specimens for the XRD test were square prisms (4, 4, and 2 mm) (Figure 4.2). More details about the parameters that were used will be described in the material and methods of chapter 6.

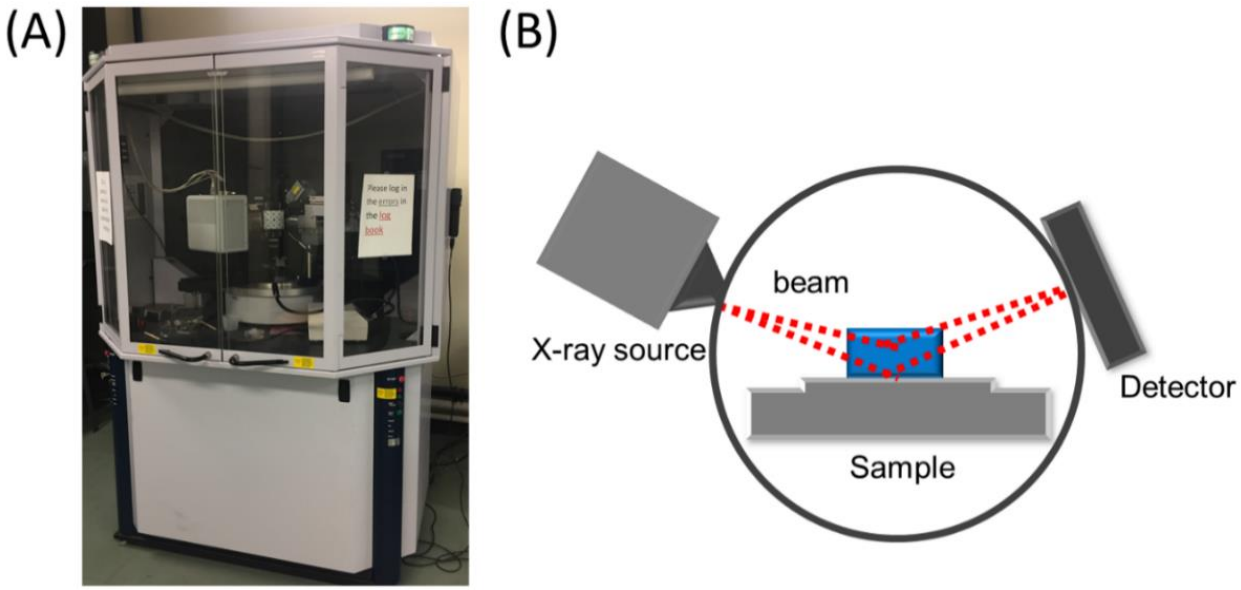


Figure 4.5. (A) photograph of the X-ray diffraction (XRD) equipment, (B) diagram shows the mechanism of the testing sample by XRD test.

4.2.5. Scanning Electron Microscope

Scanning electron microscope (SEM) is a type of electron microscope that generates high-resolution images of solid objects. The images are produced by scanning the surface with a focused beam of electrons that interacts with the atoms in the sample providing information about the surface composition, crystalline structure, and topography. In fact, the scanning signals can include secondary electrons that generate the SEM images, backscattered electrons (BSE), and diffracted backscattered electrons (EBSD) which can be used to analyze the crystal structures and orientations of the materials [235]. SEM is often used to scan an area ranging from 1 cm to 5

microns and can produce images with magnification ranging from 20X to 3000 X, and a resolution of 50 to 100 nm. The first commercial SEM appeared in the 1960s [235].

The SEM consist of an electron gun (energy of 1 to 30keV), electron lenses, sample stage, condenser lenses, detector, and an output device for display (Figure 4.6). The device is also equipped with a vacuum and with a cooling system, and it can include diffracted backscattered electron detectors (EBSD) to analyze the crystal structures and orientations of the materials [235]. Moreover, an SEM can also have an EDX (Energy Dispersive X-ray) system to analyze the elemental composition and concentrations on the surface of the solid object. The scanning electron microscopy (SEM) used in this project was an FE-SEM, FEI, Hillsboro, equipped with SEM back-scattering detector.

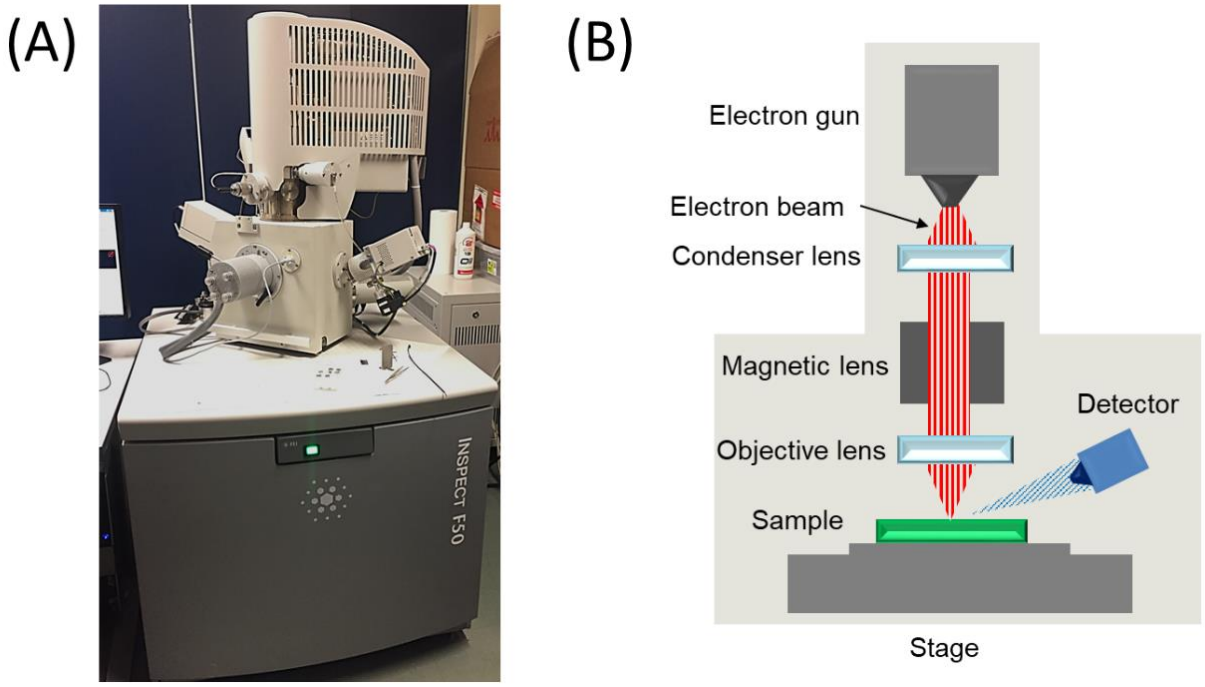


Figure 4.6. (A) photograph of the SEM equipment, (B) diagram shows the mechanism of the testing sample by SEM.

4.3. Biocompatibility

The assessment of any biomaterial should include the biocompatibility tests since the biomaterial must be safe for the patient and biologically compatible with the tissues [236]. The biocompatibility defines as the ability of a material to perform an appropriate biological response in each specific situation [231, 237]. This definition indicates an interaction among 3 factors that are the material, the host, and the application of the material [237]. As a result of this interaction, some biological responses occur after placing a material into the living tissue [237]. The biocompatibility of a material is a dynamic and continues process in which the interaction between the material and host might occur over time [237]. Determining the physical, chemical, biological interactions between a material and the living tissues of the body is the key to understand biomaterial biocompatibility [231, 237]. Several methods can be used to evaluate the biocompatibility of biomaterials, which include in vitro tests, animal test, and human clinical trials [231, 236, 237]. The in-vitro tests can be corrosion test, elemental releases test, and in-vitro cell culture assays [231, 237].

The experimental part of this project involved performing in-vitro tests including the assessment of the releases of elements, and cell culture assays. It should be noted that the in-vitro data does not completely present in-vivo clinical performance since several factors found in the oral cavity, such as salivary enzymes and acids presented in the foods and beverages, are difficult to simulate in laboratory setups [238]. However, in-vitro assessment is the first step to assess the biocompatibility of biomaterials, and it can help screen out potentially harmful materials.

4.3.1. Element Release

Dental metals and alloys might release ions into the human body, and some of these ions can have negative biological effects [236, 238]. Dental alloy release elements depending on

different factors, such as environmental conditions, elemental composition, and the crystal structure of the alloys [238, 239]. For instance, some elements have a higher tendency than other elements to be released from alloys [239]. Moreover, most alloys are more biocompatible when they are polished than when they are unpolished or sandblasted [238]. Also, elemental release and corrosion rate can vary between in-vitro test and in clinical condition (in-vivo test) as the challenge oral environment presents exposure to acidity and heat fluctuation according to the different types of foods and beverages [238].

The releases of metal ions in this project were measured using inductively coupled plasma atomic emission spectroscopy (ICP-AES; Perkin-Elmer, Wellesley, Mass). The releases of metal ions were measure from a solution of PBS (phosphate buffered saline) simulating artificial saliva containing the subjected metals and incubated for few days at 37^{o}C [217]. Then, the extracted solution was digested with concentrated nitric acid, and the standard solutions containing the measured elements were prepared for calibration at different concentrations. In this thesis, 3 different concentrations were used that are 2.0, 1.0, and 0.5 ppm (parts per million). Specimens for elemental releases test were square prisms (4, 4, and 2 mm) (Figure 4.2).

4.3.2. In-vitro Biocompatibility Assays

In-vitro biocompatibility assays are often performed in cell culture dishes, or in test tubes [231, 237]. There are many in-vitro techniques available to test biocompatibility, typically by placing cells or bacteria in contact with the tested material [236, 237]. The contact between the material and the cells can be either direct, with material and cells, or indirect, with interposition of an agar or gel layer which prevent the material from damaging the cells [236]. In these tests, the biocompatibility is usually determined by measuring the growth rate, and the metabolic function

of the cells interacting with that material [231, 237]. The in-vitro biocompatibility assays have some advantages when compared to animal experiments or human clinical trials, they are fast, inexpensive, repeatable, and do not require ethical or legal approval [231, 237]. However, these tests are very sensitive to the biological environment might provide misleading results about the ultimate biologic response [237]. The in-vitro biocompatibility assays are generally based on using fluorescent dyes or colorimetric as indicators for cell metabolism (for assessing the membrane integrity) and cell viability (for assessing the activity of living cells) [231]. Then, the cell can be examined using the fluorescent microscope.

4.3.2.1. Assays for Assessing Membrane Integrity

These assays assess cell viability by determining the number of healthy cells placed in contact with a sample of the tested material [231, 236]. This process can be done using fluorescent or colorimetric reagents that assess membrane integrity either indirectly by assessing the incubation media or directly by staining the cells [231, 236]. Some example of stains that can be used to measure the fluorescence or absorbance changes in culture media include the LDH (lactate dehydrogenase) release assay, the tetrazolium compounds-based assays, and the resazurin-based assays [231, 240]. Furthermore, some example of assays used to measure the cellular fluorescence changes by a microscope includes the trypan blue exclusion assay, the fluorescein diacetate or calcein-AM assay (calcein acetoxymethyl ester), and the propidium iodide or ethidium bromide assay [231].

LDH (lactate dehydrogenase) release assay is used to detect the cytoplasmic enzymes released from dead cells when there is no interaction with the cell membranes [231, 240]. Lactate dehydrogenase (LDH) is a stable cytoplasmic enzyme found inside living cells, and it is released rapidly into the cell culture when the plasma membrane is damaged [240]. These assays use the

tetrazolium salt that can be reduced to a red formazan [231, 240]. PI (propidium iodide) is one of the most commonly used reagents to stain the nucleic acids of dead cells with red fluorescent dyes; this stain visualizes the dead cell and that be seen using a fluorescent microscope [231]. On the other hand, there are other reagents, such as fluorescein diacetate (FDA), calcein acetoxymethyl ester (calcein-AM) that stain the living cells directly [231].

4.3.2.2. Assays for Assessing Cell Metabolic Activity

These assays assess cell viability by measuring their metabolic activity [231]. This process is done by incubating a number of living cells with a reagent that changes color on fluorescence according to cellular interactions [231]. These changes in fluorescence or color can be detected with the plate reader to assess the number of viable cells [231]. There are few assays used to evaluate cell metabolic activity, these include the resazurin reduction assays and the tetrazolium reduction assay [231].

Resazurin reduction assays are more simple, safer, sensitive and cost-effective compared to the tetrazolium reduction assays [231]. In this type of test, Resazurin is reduced according to the number of viable cells into resorufin that can be identified by measuring the changes in absorbance or fluorescence [231]. AlamarBlue®, Resazurin sodium salt, and CellTiter-Blue® Cell Viability Assay are the main commercial products based on resazurin reduction [231].

In this thesis, 2 independent test kits for cell viability (Alamar Blue, Life Technologies, Ontario, Canada) and cytotoxicity (CytotoxOne, PROMEGA, Wisconsin, USA) were used to analyze material biocompatibility with human gingival epithelial cells culture (HGEs) (Cedarlane Laboratories, Ontario, Canada). The LDH release from damaged cells and the cell metabolism were analyzed by a microplate reader (Spectramax M2E, Molecular Devices, CA, USA). Whereas, the live/dead staining assay for assessing the cytocompatibility of the metals was performed using

the fluorescein diacetate (FDA) (Sigma, Steinheim, Germany) and propidium iodide (PI) (Sigma, Steinheim, Germany, 2 mg in 1 ml PBS). Cells were analyzed and captured under a Zeiss AX10 fluorescence microscope (Carl Zeiss, Göttingen, Germany). Specimens for the biocompatibility test were square prisms (4, 4, and 2 mm) (Figure 4.2).

4.4. Patients Satisfaction Questionnaire

Dental treatments can be judged and evaluated differently by dentists or patients [76]. Successful dental treatment from the dentists' viewpoint is considered when they meet certain technical and clinical standards, but from the patients' viewpoint is considered when they meet their personal satisfaction [75]. Therefore, patient satisfaction is an important tool for evaluating the success and efficiency of treatment [1, 73, 74]. In fact, evaluating dentures should not only be evaluated based on clinical estimates but also based on patients' opinions and feedback since patient dissatisfaction with the treatment will lead to underuse and subsequent treatment failure [1, 73, 75]. Even though patients' satisfaction with their dentures is subjective and could vary among patients, it is mainly related to some denture related aspects such as hygiene, appearance, speech, comfort, retention, and mastication [69, 73-75, 241-243].

One successful way of evaluating Removable Partial Dentures (RPDs) is using a self-administered instrument such as the visual analogue scale (VAS) questionnaire [73, 241]. The VAS questionnaire was described for the first time in 1921, but it was used seriously after 1969 in many research [244]. Due to its simplicity, the VAS questionnaire has been used efficiently in many clinical studies to evaluate patient satisfaction [69, 73-75, 241-244]. A VAS questionnaire is a scale that consists of a line and anchors words at each end (e.g. extremely satisfied and extremely dissatisfied, or good and bad) [244].

The McGill Denture Satisfaction Instrument was used for evaluating dental prostheses based on patients' satisfaction [73, 243, 245]. Dr. Feine and Dr. Awad developed and proposed the McGill denture satisfaction instrument in 1998, and it was used to evaluate the patient's satisfaction with their mandibular implant-supported dental prosthesis [73, 243, 245]. This questionnaire is assessing patient's satisfaction relating to some aspects of the dental prosthesis such as general satisfaction, comfort, ability to speak, ease of cleaning, aesthetics, stability, chewing ability, chewing efficiency, and oral condition. The McGill Denture Satisfaction Instrument is a VAS questionnaire that consists of a scale of 100 mm anchored at the end of the scales by the words "not at all satisfied" and "extremely satisfied" [73, 243, 245]. In this thesis, McGill Denture Satisfaction Instrument was used to evaluate the removable partial denture (RPD).

Chapter 5: List of References (Chapters 1-4)

1. Campbell SD, Cooper L, Craddock H, Hyde TP, Nattress B, Pavitt SH, et al. Removable partial dentures: The clinical need for innovation. The Journal of Prosthetic Dentistry. 2017.
2. Carr AB, Brown DT. McCracken's removable partial prosthodontics Saint Louis, Mo.: Elsevier Mosby; 2011.
3. Christensen GJ. A consumer's guide to dentistry St. Louis, Mo.: Mosby; 2002.
4. Preshaw P, Walls A, Jakubovics N, Moynihan P, Jepson N, Loewy Z. Association of removable partial denture use with oral and systemic health. Journal of Dentistry. 2011; 39(11):711-9.
5. Saito M, Notani K, Miura Y, Kawasaki T. Complications and failures in removable partial dentures: a clinical evaluation. Journal of Oral Rehabilitation. 2002; 29(7):627-33.
6. Bural C, Buzbas B, Ozatik S, Bayraktar G, Emes Y. Distal extension mandibular removable partial denture with implant support. European Journal of Dentistry. 2016; 10(4):566.
7. Chaiyabutr Y, Brudvik JS. Removable partial denture design using milled abutment surfaces and minimal soft tissue coverage for periodontally compromised teeth: a clinical report. The Journal of Prosthetic Dentistry. 2008; 99(4):263-6.
8. Hummel SK, Wilson MA, Marker VA, Nunn ME. Quality of removable partial dentures worn by the adult U.S. population. The Journal of Prosthetic Dentistry. 2002; 88(1):37-43.
9. Bilhan H, Erdogan O, Ergin S, Celik M, Ates G, Geckili O. Complication rates and patient satisfaction with removable dentures. The Journal of Advanced Prosthodontics. 2012; 4(2):109-15.
10. Donaldson J. The use of gold in dentistry: an historical overview. Journal of the History of Dentistry. 2012; 60(3):134-47.

11. Girardot RL. History and development of partial denture design. The Journal of the American Dental Association. 1941; 28(9):1399-408.

12. Becker CM, Kaiser DA, Goldfogel MH. Evolution of removable partial denture design. Journal of Prosthodontics. 1994;3(3):158-66.

13. Venkatesh KV, Nandini VV. Direct Metal Laser Sintering: A digitized metal casting technology. The Journal of Indian Prosthodontic Society. 2013; 13(4):389-92.

14. Koutsoukis T, Zinelis S, Eliades G, Al-Wazzan K, Rifaiy MA, Al Jabbari YS. Selective laser melting technique of co-cr dental alloys: a review of structure and properties and comparative analysis with other available techniques. Journal of Prosthodontics. 2015; 24(4):303-12.

15. Vallittu PK, Miettinen T. Duration of induction melting of cobalt-chromium alloy and its effect on resistance to deflection fatigue of cast denture clasps. The Journal of Prosthetic Dentistry. 1996; 75(3):332-6.

16. Puskar T, Jevremovic D, Williams RJ, Eggbeer D, Vukelic D, Budak I. A comparative analysis of the corrosive effect of artificial saliva of variable Ph on DMLS and cast Co-Cr-Mo dental alloy. Journal of Materials. 2014; 7(9):6486-501.

17. Davidowitz G, Kotick PG. The use of CAD/CAM in dentistry. Dental Clinics of North America. 2011;55(3):559-70.

18. Barazanchi A, Li KC, Al-Amleh B, Lyons K, Waddell JN. Additive technology: update on current materials and applications in dentistry. journal of prosthodontics. 2017; 26(2):156-63.

19. Van Noort R. The future of dental devices is digital. Dental Materials. 2012; 28(1):3-12.

20. Lima JMC, Anami LC, Araujo RM, Pavanelli CA. Removable partial dentures: use of rapid prototyping. Journal of Prosthodontics. 2014; 23(7):588-91.

21. Akova T, Ucar Y, Tukay A, Balkaya MC, Brantley WA. Comparison of the bond strength of laser-sintered and cast base metal dental alloys to porcelain. Dental Materials. 2008; 24(10):1400-4.

22. Kruth J-P, Mercelis P, Van Vaerenbergh J, Froyen L, Rombouts M. Binding mechanisms in selective laser sintering and selective laser melting. Rapid Prototyping Journal. 2005; 11(1):26-36.

23. Hollander DA, von Walter M, Wirtz T, Sellei R, Schmidt-Rohlfing B, Paar O, et al. Structural, mechanical and in vitro characterization of individually structured Ti–6Al–4V produced by direct laser forming. Biomaterials. 2006; 27(7):955-63.

24. Vandenbroucke B, Kruth J-P. Selective laser melting of biocompatible metals for rapid manufacturing of medical parts. Rapid Prototyping Journal. 2007; 13(4):196-203.

25. Hedberg YS, Qian B, Shen Z, Virtanen S, Wallinder IO. In vitro biocompatibility of CoCrMo dental alloys fabricated by selective laser melting. Dental Materials. 2014; 30(5):525-34.

26. Omar Alageel, Ammar Alsheghri, Suliman Algezani, Tamimi F. Determining the retention of removable partial dentures. The Journal of Prosthetic Dentistry. 2018.

27. Chen Q, Wu J, Li S, Lyu P, Wang Y, Li M. An ontology-driven, case-based clinical decision support model for removable partial denture design. Scientific Reports. 2016; 6.

28. John Beaumont A. Microcomputer-aided removable partial denture design: The next evolution. The Journal of Prosthetic Dentistry. 1989;62(5):551-6.

29. Bonnet G, Lance C, Bessadet M, Tamini F, Veyrune J-L, Francois O, et al. Teaching removable partial denture design: 'METACIEL', a novel digital procedure. International Journal of Medical Education. 2018; 9:24.

30. Cote SE, Singh H. Chapter 45 - Dental diseases and disorders. In: Walker PF, Barnett ED, editors. Immigrant Medicine. Edinburgh: W.B. Saunders; 2007. p. 597-610.

31. Carr AB, Brown DT. Chapter 1 - Partially edentulous epidemiology, Physiology, and Terminology. McCracken's Removable Partial Prosthodontics (Twelfth Edition). Saint Louis: Mosby; 2011. p. 2-7.

32. Misch CE. Chapter 1 - Rationale for dental implants. Dental Implant Prosthetics (Second Edition). St. Louis: Mosby; 2015. p. 1-25.

33. Misch CE. Dental Implant Prosthetics (Second Edition). St. Louis: Mosby; 2015.

34. Emami E, de Souza RF, Kabawat M, Feine JS. The impact of edentulism on oral and general health. International Journal of Dentistry. 2013.

35. Smith B. Implant overdentures: The standard of care for edentulous patients. British Dental Journal. 2005;199(3):180.

36. Sakar O. Removable Partial Dentures: a Practitioners' Manual. 2016.

37. Carr AB, Brown DT. Chapter 2 - Considerations for managing partial tooth loss: tooth replacements from the patient perspective. McCracken's Removable Partial Prosthodontics (Twelfth Edition). Saint Louis: Mosby; 2011. p. 8-15.

38. Freedman GA. Contemporary esthetic dentistry St. Louis, Mo.; London: Mosby; 2012.

39. Phoenix RD, Cagna DR, DeFreest CF, Stewart KL. Stewart's clinical removable partial prosthodontics Chicago: Quintessence Pub.; 2008.

40. Carr AB, Brown DT. Chapter 3 - Classification of partially edentulous arches. McCracken's Removable Partial Prosthodontics (Twelfth Edition). Saint Louis: Mosby; 2011. p. 16-20.

41. Craig RG, Powers JM, Sakaguchi RL. Craig's restorative dental materials. St. Louis, Mo.: Mosby Elsevier; 2006.

42. Wiskott HWA. Fixed prosthodontics: principles and clinics. London; Chicago: Quintessence Publishing; 2011.

43. Gill DSNFB. Orthodontics principles and practice Chichester, West Sussex: Wiley-Blackwell; 2011.

44. Massad J. Chapter 24 - Complete denture esthetics. Contemporary Esthetic Dentistry. Saint Louis: Mosby; 2012. p. 560-74.

45. Smith ED, Howe LC. Planning and making crowns and bridges. CRC Press; 2006.

46. Yeung A, Lo E, Chow T, Clark R. Oral health status of patients 5–6 years after placement of cobalt–chromium removable partial dentures. Journal of Oral Rehabilitation. 2000; 27(3):183-9.

47. Bergman B, Hugoson A, Olsson C-O. Caries, periodontal and prosthetic findings in patients with removable partial dentures: a ten-year longitudinal study. The Journal of Prosthetic Dentistry. 1982; 48(5):506-14.

48. Wilson V. Acrylic partial dentures-interim or permanent prostheses. Journal of South African Dental Association. 2009;64(10):430-p6.

49. Hohmann A, Hielscher W. Principles of design and fabrication in prosthodontics. 2016.

50. Khan S, Geerts G. Aesthetic clasp design for removable partial dentures: a literature review. 2005.

51. Walmsley AD. Acrylic partial dentures. Dental Update. 2003; 30(8):424-9.

52. Carr AB, Brown DT. Chapter 5 - Major and Minor Connectors. McCracken's Removable Partial Prosthodontics (Twelfth Edition). Saint Louis: Mosby; 2011. p. 29-55.

53. Carr AB, Brown DT. Chapter 7 - Direct retainers. McCracken's Removable Partial Prosthodontics (Twelfth Edition). Saint Louis: Mosby; 2011. p. 67-95.

54. Carr AB, Brown DT. Chapter 8 - Indirect retainers. McCracken's Removable Partial Prosthodontics (Twelfth Edition). Saint Louis: Mosby; 2011. p. 96-102.

55. Carr AB, Brown DT. Chapter 6 - Rests and Rest Seats. McCracken's Removable Partial Prosthodontics (Twelfth Edition). Saint Louis: Mosby; 2011. p. 56-66.

56. Sakaguchi RL, Powers JM. Craig's restorative dental materials St. Louis, Mo.: Elsevier/Mosby; 2012.

57. O'Brien WJ. Dental materials and their selection Hanover Park, IL: Quintessence Pub. Co.; 2008.

58. Carr AB, Brown DT. Chapter 9 - Denture base considerations. McCracken's Removable Partial Prosthodontics (Twelfth Edition). Saint Louis: Mosby; 2011. p. 103-14.

59. Alageel O, Abdallah MN, Alsheghri A, Song J, Caron E, Tamimi F. Removable partial denture alloys processed by laser-sintering technique. Journal of Biomedical Materials Research Part B: Applied Biomaterials. 2017.

60. Lucchetti MC, Fratto G, Valeriani F, De Vittori E, Giampaoli S, Papetti P, et al. Cobalt-chromium alloys in dentistry: An evaluation of metal ion release. Journal of Prosthetic Dentistry. 2015; 114(4):602-8.

61. Matković T, Matković P, Malina J. Effects of Ni and Mo on the microstructure and some other properties of Co–Cr dental alloys. Journal of Alloys and Compounds. 2004; 366(1):293-7.

62. Carek A, Živko-Babić J, Schauperl Z, Badel T. Mechanical properties of Co-Cr alloys for metal base framework. International Journal of Prosthodontics and Restorative Dentistry. 2011; 1(1):13-9.

63. Giacchi J, Morando C, Fornaro O, Palacio H. Microstructural characterization of as-cast biocompatible Co–Cr–Mo alloys. Materials Characterization. 2011; 62(1):53-61.

64. Takaichi A, Suyalatu, Nakamoto T, Joko N, Nomura N, Tsutsumi Y, et al. Microstructures and mechanical properties of Co–29Cr–6Mo alloy fabricated by selective laser melting process for dental applications. Journal of the Mechanical Behavior of Biomedical Materials. 2013; 21:67-76.

65. Vuorinen AM, Dyer SR, Lassila LVJ, Vallittu PK. Bonding of BisGMA–TEGDMA-Resin to Bulk Poly (Paraphenylene) Based Rigid Rod Polymer. Composite Interfaces. 2011; 18(5):387-98.

66. Alageel O, Abdallah M-N, Luo ZY, Del-Rio-Highsmith J, Cerruti M, Tamimi F. Bonding metals to poly (methyl methacrylate) using aryldiazonium salts. Dental Materials. 2015; 31(2):105-14.

67. Rupali Gautam RDS, Vinod P. Sharma, Ramashanker Siddhartha, Pooran Chand RK. Biocompatibility of polymethylmethacrylate resins used in dentistry. Wiley Periodicals. 2012.

68. Dental Materials and Their Selection. 4 ed. William J. O'Brien PF, editor. Hanover Park, IL: Quintessence Publishing Co, Inc.; 2008.

69. Almufleh B, Emami E, Alageel O, de Melo F, Seng F, Caron E, et al. Patient satisfaction with laser-sintered removable partial dentures: A crossover pilot clinical trial. The Journal of Prosthetic Dentistry. 2017.

70. DeGarmo EP, Black JT, Kohser RA, Klamecki BE. Materials and process in manufacturing: Prentice Hall; 1997.

71. Loney RW. Removable partial denture manual. Dalhousie University. 2011:43-56.

72. Sakar O. Removable Partial Dentures A Practitioners' Manual. 2016.

73. Awad MA, Feine JS. Measuring patient satisfaction with mandibular prostheses. Community Dentistry and Oral Epidemiology. 1998; 26(6):400-5.

74. Koyama S, Sasaki K, Yokoyama M, Sasaki T, Hanawa S. Evaluation of factors affecting the continuing use and patient satisfaction with removable partial dentures over 5 years. Journal of Prosthodontic Research. 2010; 54(2):97-101.

75. Zlatarić DK, Čelebić A, Valentić-Peruzović M, Jerolimov V, Pandurić J. A survey of treatment outcomes with removable partial dentures. Journal of Oral Rehabilitation. 2003; 30(8):847-54.

76. Čelebić A, Knezović-Zlatarić D. A comparison of patient's satisfaction between complete and partial removable denture wearers. Journal Dentistry. 2003; 31(7):445-51.

77. Zlataric DK, Celebic A. Factors related to patients' general satisfaction with removable partial dentures: a stepwise multiple regression analysis. International Journal of Prosthodontics. 2008; 21(1):86.

78. Benso B, Kovalik AC, Jorge JH, Campanha NH. Failures in the rehabilitation treatment with removable partial dentures. Acta Odontologica Scandinavica. 2013; 71(6):1351-5.

79. Weinstein M, Schuchman J, Lieberman J, Rosen P. Age and denture experience as determinants in patient denture satisfaction. The Journal of Prosthetic Dentistry. 1988; 59(3):327-9.

80. Frank RP, Milgrom P, Leroux BG, Hawkins NR. Treatment outcomes with mandibular removable partial dentures: a population-based study of patient satisfaction. The Journal of Prosthetic Dentistry. 1998; 80(1):36-45.

81. Sharma A, Siwach A, Nayak L, Tiwary VK. Intricate assessment and evaluation of levels of satisfaction and maintenance among patients wearing removable partial dentures; a survey-based study. Journal of Advanced Medical and Dental Sciences Research. 2017;5(8):37.

82. Shams A, Tavanafar S, Dastjerdi MR, Chaijan KA. Patient satisfaction and complication rates after delivery of removable partial dentures: a 4-year retrospective study. Journal of Research in Dental Sciences. 2015; 6(4):225.

83. Kok IJ, Cooper LF, Guckes AD, McGraw K, Wright RF, Barrero CJ, et al. Factors influencing removable partial denture patient-reported outcomes of quality of life and satisfaction: a systematic review. Journal of Prosthodontics. 2017; 26(1):5-18.

84. Roberts B. A survey of chrome-cobalt partial dentures. The New Zealand Dental Journal. 1978; 74(338):203.

85. Kamber-Cesir A, Dzonlagic A, Ajanovic M, Delalic A. Assessment of patient's satisfaction with the partial removable denture therapy. Pesquisa Brasileira em Odontopediatria e Clínica Integrada. 2011;11(2).

86. Abouelkomsan AM, Butt AM, Dall AQ. Removable partial dentures: Patient satisfaction with associated demographic and biomechanical factors. Pakistan Oral and Dental Journal. 2012; 32(3).

87. Germundsson B, Hellman M, Odman P. Effects of rehabilitation with conventional removable partial dentures on oral health--a cross-sectional study. Swedish Dental Journal. 1984; 8(4):171-82.

88. Wong M, Ettinger R, Barsby M. An evaluation of removable partial dentures: a retrospective study. The Iowa Dental Journal. 1995; 81(3):13-6.

89. Jepson N, Thomason J, Steele J. The influence of denture design on patient acceptance of partial dentures. British Dental Journal. 1995; 178(8):296.

90. Wetherell JD, Smales RJ. Partial denture failures: A long-term clinical survey. Journal of Dentistry. 1980; 8(4):333-40.

91. Yeung A, Lo E, Clark R, Chow T. Usage and status of cobalt–chromium removable partial dentures 5–6 years after placement. Journal of Oral Rehabilitation. 2002; 29(2):127-32.

92. Koyama S, Sasaki K, Kawata T, Atsumi T, Watanabe M. Multivariate analysis of patient satisfaction factors affecting the usage of removable partial dentures. International Journal of Prosthodontics. 2008; 21(6).

93. Mazurat NM, Mazurat RD. Discuss before fabricating: communicating the realities of partial denture therapy. part II: clinical outcomes. Journal of the Canadian Dental Association. 2003; 69(2):96-100.

94. Redford M, Drury T, Kingman A, Brown L. Denture use and the technical quality of dental prostheses among persons 18–74 years of age: United States, 1988–1991. Journal of Dental Research. 1996; 75(2_suppl):714-25.

95. Cheng H, Xu M, Zhang H, Wu W, Zheng M, Li X. Cyclic fatigue properties of cobalt-chromium alloy clasps for partial removable dental prostheses. The Journal of Prosthetic Dentistry. 2010; 104(6):389-96.

96. Sato Y. Clinical methods for adjusting retention force of cast clasps. The Journal of Prosthetic Dentistry. 1999; 82(5):557-61.

97. Ahmad I, Sherriff M, Waters N. The effect of reducing the number of clasps on removable partial denture retention. The Journal of Prosthetic Dentistry. 1992; 68(6):928-33.

98. Brockhurst P. A new design for partial denture circumferential clasp arms. Australian Dental Journal. 1996; 41(5):317-23.

99. Sandu L, Faur N, Bortun C. Finite element stress analysis and fatigue behavior of cast circumferential clasps. The Journal of Prosthetic Dentistry. 2007; 97(1):39-44.

100. Mahmoud AAA, Wakabayashi N, Takahashi H. Prediction of permanent deformation in cast clasps for denture prostheses using a validated nonlinear finite element model. Dental Materials. 2007; 23(3):317-24.

101. Vallittu PK, Kokkonen M. Deflection fatigue of cobalt-chromium, titanium, and gold alloy cast denture clasp. The Journal of Prosthetic Dentistry. 1995; 74(4):412-9.

102. Keltjens H, Mulder J, Käyser A, Creugers N. Fit of direct retainers in removable partial dentures after 8 years of use. Journal of Oral Rehabilitation. 1997; 24(2):138-42.

103. Kim D, Park C, Yi Y, Cho L. Comparison of cast Ti-Ni alloy clasp retention with conventional removable partial denture clasps. The Journal of Prosthetic Dentistry. 2004;91(4):374-82.

104. Hammond P, Davenport J, Potts A. Knowledge-based design of removable partial dentures using direct manipulation and critiquing. Journal of Oral Rehabilitation. 1993;20(2):115-23.

105. Weinstein GM, Zientz MT. The Dental Reference Manual. 2017.

106. Mendonça EA. Clinical decision support systems: perspectives in dentistry. Journal of Dental Education. 2004; 68(6):589-97.

107. Budtz-Jorgensen E, Bochet G. Alternate framework designs for removable partial dentures. The Journal of Prosthetic Dentistry. 1998; 80(1):58-66.

108. Arda T, Arikan A. An in vitro comparison of retentive force and deformation of acetal resin and cobalt-chromium clasps. The Journal of Prosthetic Dentistry. 2005;94(3):267-74.

109. Tannous F, Steiner M, Shahin R, Kern M. Retentive forces and fatigue resistance of thermoplastic resin clasps. Dental Materials. 2012;28(3):273-8.

110. Dula LJ, Ahmedi EF, Lila-Krasniqi ZD, Shala KS. Clinical evaluation of removable partial dentures on the periodontal health of abutment teeth: a retrospective study. The Open Dentistry Journal. 2015; 9:132.

111. Taggart WH. A new and accurate method of making gold inlays: Philadelphia: SS White Dental Manufacturing Company; 1969.

112. Ucar Y, Akova T, Akyil MS, Brantley WA. Internal fit evaluation of crowns prepared using a new dental crown fabrication technique: laser-sintered Co-Cr crowns. Journal of Prosthetic Dentistry. 2009; 102(4):253-9.

113. Kohorst P, Brinkmann H, Li J, Borchers L, Stiesch M. Marginal accuracy of four-unit zirconia fixed dental prostheses fabricated using different computer-aided design/computer-aided manufacturing systems. European Journal of Oral Sciences. 2009; 117(3):319-25.

114. Bae E-J, Jeong I-D, Kim W-C, Kim J-H. A comparative study of additive and subtractive manufacturing for dental restorations. Journal of Prosthetic Dentistry. 2017; 118(2):187-93.

115. Averyanova M, Bertrand P, Verquin B. Manufacture of Co-Cr dental crowns and bridges by selective laser Melting technology. Virtual and physical prototyping. 2011;6(3):179-85.

116. carr ab, stewart rb. full-arch implant framework casting accuracy: preliminary in vitro observation for in vivo testing. Journal of Prosthodontics. 1993; 2(1):2-8.

117. Ito M, Yamagishi T, Oshida Y, Munoz CA. Effect of selected physical properties of waxes on investments and casting shrinkage. The Journal of Prosthetic Dentistry. 1996; 75(2):211-6.

118. Gapido CG, Kobayashi H, Miyakawa O, Kohno S. Fatigue resistance of cast occlusal rests using Co-Cr and Ag-Pd-Cu-Au alloys. The Journal of Prosthetic Dentistry. 2003; 90(3):261-9.

119. Miyazaki T, Hotta Y. CAD/CAM systems available for the fabrication of crown and bridge restorations. Australian Dental Journal. 2011; 56(s1):97-106.

120. Patel N. Contemporary dental CAD/CAM: modern chairside/lab applications and the future of computerized dentistry. Compendium of Continuing Education in Dentistry. 2014; 35(10):739-46.

121. Strub JR, Rekow ED, Witkowski S. Computer-aided design and fabrication of dental restorations: current systems and future possibilities. The Journal of the American Dental Association. 2006; 137(9):1289-96.

122. Healthcare 3D Printing Market worth over $2.2 Billion by 2024: Global Market Insights, Inc. Global Market Insights. 2017.

123. Additive Manufacturing Market to Develop Rapidly by 2016 – 2026. Persistence Market Research. 2017.

124. Schubert C, van Langeveld MC, Donoso LA. Innovations in 3D printing: a 3D overview from optics to organs. The British Journal of Ophthalmology. 2014; 98(2):159-61.

125. Barazanchi A, Li KC, Al-Amleh B, Lyons K, Waddell JN. Additive technology: update on current materials and applications in dentistry. Journal of Prosthodontics. 2017; 26(2):156-63.

126. Ventola CL. Medical applications for 3D printing: current and projected uses. Pharmacy and Therapeutics. 2014; 39(10):704.

127. Beuer F, Schweiger J, Edelhoff D. Digital dentistry: an overview of recent developments for CAD/CAM generated restorations. British Dental Journal. 2008; 204(9):505.

128. Braian M, Jonsson D, Kevci M, Wennerberg A. Geometrical accuracy of metallic objects produced with additive or subtractive manufacturing: A comparative in vitro study. Dental Materials. 2018.

129. Alghazzawi TF. Advancements in CAD/CAM technology: Options for practical implementation. Journal of Prosthodontic Research. 2016;60(2):72-84.

130. Abduo J, Lyons K, Bennamoun M. Trends in computer-aided manufacturing in prosthodontics: a review of the available streams. International Journal of Dentistry. 2014.

131. Beuer F, al. Digital dentistry: an overview of recent developments for CAD/CAM generated restorations. British Dental Journal. 2008; 204(9):505-11.

132. Tapie L, Lebon N, Mawussi B, Fron HC, Duret F, Attal J. Understanding dental CAD/CAM for restorations--the digital workflow from a mechanical engineering viewpoint. International journal of computerized dentistry. 2015; 18(1):21-44.

133. Alghazzawi TF. Advancements in CAD/CAM technology: options for practical implementation. Journal of Prosthodontic Research. 2016; 60(2):72-84.

134. Miyazaki T, Hotta Y, Kunii J, Kuriyama S, Tamaki Y. A review of dental CAD/CAM: current status and future perspectives from 20 years of experience. Dental Materials Journal. 2009; 28(1):44-56.

135. Baroudi K, Ibraheem SN. Assessment of chair-side computer-aided design and computer-aided manufacturing restorations: a review of the literature. Journal of International Oral Health. 2015; 7(4):96-104.

136. Brochu M. Focalisation sur les scanneurs dentaires numériques: la science a l'appui. Journal Canadien de Dentisterie Restauratrice et de Prosthodontie. 2009 :45-8.

137. Giordano R. Materials for chairside CAD/CAM–produced restorations. The Journal of the American Dental Association. 2006; 137:14S-21S.

138. Christensen GJ. Is now the time to purchase an in-office CAD/CAM device? The Journal of the American Dental Association. 2006; 137(2):235-8.

139. Cheng LKL, Kooistra CS. Current chairside CAD/CAM systems and materials for dental restorations. Clinical Update. 2014; 36(4).

140. Fasbinder DJ, Dennison JB, Heys D, Neiva G. A clinical evaluation of chairside lithium disilicate CAD/CAM crowns. The Journal of the American Dental Association. 2010; 141:10.

141. Fasbinder DJ. Chairside CAD/CAM: an overview of restorative material options. Compendium of Continuing Education in Dentistry. 2012; 33(1):50, 2-8.

142. Fasbinder DJ, Neiva GF. Surface evaluation of polishing techniques for new resilient CAD/CAM restorative materials. Journal of Esthetic and Restorative Dentistry. 2016; 28(1):56-66.

143. Lambert H, Durand J-C, Jacquot B, Fages M. Dental biomaterials for chairside CAD/CAM: State of the art. The Journal of Advanced Prosthodontics. 2017; 9(6):486-95.

144. Skramstad M, Fasbinder D. Full-contour zirconia fixed partial dentures as chairside applications: a case report. Compendium of Continuing Education in Dentistry. 2016; 37(9):648-54.

145. Blatz MB, Conejo J. The current state of chairside digital dentistry and materials. Dental Clinics. 2019; 63(2):175-97.

146. Sannino G, Germano F, Arcuri L, Bigelli E, Arcuri C, Barlattani A. CEREC CAD/CAM chairside system. Oral Implantol. 2014; 7(3):57-70.

147. Andreev A, Kosenko M, Petrovskiy V. The use of laser milling for prototyping of dental structures. Bulletin of the Lebedev Physics Institute. 2015; 42(9):255-9.

148. Joda T, Ferrari M, Gallucci GO, Wittneben JG, Bragger U. Digital technology in fixed implant prosthodontics. Periodontology 2000. 2017; 73(1):178-92.

149. Rynerson M. Introduction of our revolutionary laser milling technology dentalwings.com: Dentalwings.com; 2015.

150. Touri M, Kabirian F, Saadati M, Ramakrishna S, Mozafari M. Additive manufacturing of biomaterials− the evolution of rapid prototyping. Advanced Engineering Materials. 2019: 1800511.

151. Dawood A, Marti BM, Sauret-Jackson V, Darwood A. 3D printing in dentistry. British Dental Journal. 2015; 219(11):521.

152. Stansbury JW, Idacavage MJ. 3D printing with polymers: Challenges among expanding options and opportunities. Dental Materials. 2016; 32(1):54-64.

153. Azari A, Nikzad S. The evolution of rapid prototyping in dentistry: a review. Rapid Prototyping Journal. 2009; 15(3):216-25.

154. Revilla-León M, Özcan M. Additive manufacturing technologies used for processing polymers: Current status and potential application in prosthetic dentistry. Journal of Prosthodontics. 2019; 28(2):146-58.

155. Totu EE, Nechifor AC, Nechifor G, Aboul-Enein HY, Cristache CM. Poly(methyl methacrylate) with TiO2 nanoparticles inclusion for stereolithographic complete denture manufacturing - the future in dental care for elderly edentulous patients?. Journal of Dentistry. 2017; 59:68-77.

156. Liu Q, Leu MC, Schmitt SM. Rapid prototyping in dentistry: technology and application. The International Journal of Advanced Manufacturing Technology. 2006; 29(3-4):317-35.

157. Osman RB, Alharbi N, Wismeijer D. Build angle: does it influence the accuracy of 3d-printed dental restorations using digital light-processing technology? International Journal of Prosthodontic. 2017; 30(2):182-8.

158. Li SY, Zhou M, Lai YX, Geng YM, Cao SS, Chen XM. Biological evaluation of three-dimensional printed co-poly lactic acid/glycolic acid/tri-calcium phosphate scaffold for bone reconstruction. Chinese Journal of Stomatology. 2016; 51(11):661-6.

159. Dehurtevent M, Robberecht L, Hornez JC, Thuault A, Deveaux E, Behin P. Stereolithography: A new method for processing dental ceramics by additive computer-aided manufacturing. Dental Materials. 2017; 33(5):477-85.

160. Abbasi AJ, Azari A, Mohebbi SZ, Javani A. Mandibular rami implant: a new approach in mandibular reconstruction. journal of oral and maxillofacial surgery. Official Journal of The American Association of Oral and Maxillofacial Surgeons. 2017; 75(12):2550-8.

161. Ahn SY, Kim NH, Kim S, Karabucak B, Kim E. Computer-aided design/computer-aided manufacturing-guided endodontic surgery: guided osteotomy and apex localization in a mandibular molar with a thick buccal bone plate. Journal of Endodontics. 2018; 44(4):665-70.

162. Wang L, Chen ZY, Liu R, Zeng H. Clinical application of individualized three-dimensional printing implant template in multi-tooth dental implantation. Shanghai Journal of Stomatology. 2017; 26(4):453-7.

163. Rathi N, Scherer MD, McGlumphy E. Stabilization of a computer-aided implant surgical guide using existing dental implants with conversion of an overdenture to a fixed prosthesis. Journal of Prosthodontic. 2014; 23(8):634-8.

164. Lee WS, Lee DH, Lee KB. Evaluation of internal fit of interim crown fabricated with CAD/CAM milling and 3D printing system. Journal of Advanced Prosthodontic. 2017; 9(4):265-70.

165. Lin WS, Harris BT, Morton D. Use of CBCT imaging, open-source modeling software, and desktop stereolithography 3D printing to duplicate a removable dental prosthesis-a proof of concept. Compendium of Continuing Education in Dentistry. 2017; 38(8):e5-e8.

166. Reyes A, Turkyilmaz I, Prihoda TJ. Accuracy of surgical guides made from conventional and a combination of digital scanning and rapid prototyping techniques. Journal of Prosthetic Dentistry. 2015; 113(4):295-303.

167. Sun YC, Jin EL, Zhao T, Wang Y, Ye HQ, Zhou YS. Progress in research and application of the edentulous custom trays. Chinese Journal of Stomatology. 2016; 51(11):698-701.

168. Ishida Y, Miyasaka T. Dimensional accuracy of dental casting patterns created by 3D printers. Dental Materials Journal. 2016; 35(2):250-6.

169. Alharbi N, Osman R, Wismeijer D. Effects of build direction on the mechanical properties of 3D-printed complete coverage interim dental restorations. Journal of Prosthetic Dentistry. 2016; 115(6):760-7.

170. Johnson AW. Dental occlusion ties: A rapid, safe, and non-invasive maxillo-mandibular fixation technology. Laryngoscope Investigative Otolaryngology. 2017; 2(4):178-83.

171. Martorelli M, Gerbino S, Giudice M, Ausiello P. A comparison between customized clear and removable orthodontic appliances manufactured using RP and CNC techniques. Dental Materials. 2013; 29(2):e1-e10.

172. Batstone MD. Reconstruction of major defects of the jaws. Australian Dental Journal. 2018; 63 Suppl 1:S108-S13.

173. Unkovskiy A, Spintzyk S, Brom J, Huettig F, Keutel C. Direct 3D printing of silicone facial prostheses: A preliminary experience in digital workflow. Journal of Prosthetic Dentistry. 2018.

174. Shaheen E, Sun Y, Jacobs R, Politis C. Three-dimensional printed final occlusal splint for orthognathic surgery: design and validation. International Journal of Oral and Maxillofacial Surgery. 2017; 46(1):67-71.

175. Tamimi F, Torres J, Al-Abedalla K, Lopez-Cabarcos E, Alkhraisat MH, Bassett DC, et al. Osseointegration of dental implants in 3D-printed synthetic onlay grafts customized according to bone metabolic activity in recipient site. Biomaterials. 2014; 35(21):5436-45.

176. Deng KH, Wang Y, Chen H, Zhao YJ, Zhou YS, Sun YC. Quantitative evaluation of printing accuracy and tissue surface adaptation of mandibular complete denture polylactic acid pattern fabricated by fused deposition modeling technology. Chinese Journal of Stomatology. 2017; 52(6):342-5.

177. Revilla Leon M, Klemm IM, Garcia-Arranz J, Ozcan M. 3D Metal Printing - Additive manufacturing technologies for frameworks of implant-borne fixed dental prosthesis. The European Journal of Prosthodontics and Restorative Dentistry. 2017; 25(3):143-7.

178. Rengier F, Mehndiratta A, von Tengg-Kobligk H, Zechmann CM, Unterhinninghofen R, Kauczor H-U, et al. 3D printing based on imaging data: review of medical applications. International Journal of Computer Assisted Radiology and Surgery. 2010; 5(4):335-41.

179. Weller C, Kleer R, Piller FT. Economic implications of 3D printing: Market structure models in light of additive manufacturing revisited. International Journal of Production Economics. 2015; 164:43-56.

180. Sames WJ, List F, Pannala S, Dehoff RR, Babu SS. The metallurgy and processing science of metal additive manufacturing. International Materials Reviews. 2016; 61(5):315-60.

181. Mazzoli A. Selective laser sintering in biomedical engineering. Medical and Biological Engineering and Computing. 2013; 51(3):245-56.

182. Wang X, Laoui T, Bonse J, Kruth J-P, Lauwers B, Froyen L. Direct selective laser sintering of hard metal powders: experimental study and simulation. The International Journal of Advanced Manufacturing Technology. 2002; 19(5):351-7.

183. Shellabear M, Nyrhilä O. DMLS-Development history and state of the art. Laser Assisted Netshape Engineering 4, Proceedings of the 4th Lane. 2004:21-4.

184. Oyar P. Laser sintering technology and balling phenomenon. Photomedicine and Laser Surgery. 2018;36(2):72-7.

185. Gokuldoss PK, Kolla S, Eckert J. Additive manufacturing processes: selective laser melting, electron beam melting and binder jetting-selection guidelines. Materials. 2017;10(6).

186. Tan XP, Tan YJ, Chow CSL, Tor SB, Yeong WY. Metallic powder-bed based 3D printing of cellular scaffolds for orthopedic implants: A state-of-the-art review on manufacturing, topological design, mechanical properties and biocompatibility. Materials Science and Engineering: C. 2017;7 6:1328-43.

187. Yap CY, Chua CK, Dong ZL, Liu ZH, Zhang DQ, Loh LE, et al. Review of selective laser melting: Materials and applications. Applied Physics Reviews. 2015; 2(4):041101.

188. Liu YF, Yu H, Wang WN, Gao B. A preliminary study on the forming quality of titanium alloy removable partial denture frameworks fabricated by selective laser melting. Chinese Journal of Stomatology. 2017; 52(6):351-4.

189. Ohkubo C, Sato Y, Nishiyama Y, Suzuki Y. Titanium removable denture based on a one-metal rehabilitation concept. Dental Materials Journal. 2017; 36(5):517-23.

190. Ye Y, Jiao T, Zhu J, Sun J. Adaptation and micro-structure of Co-Cr alloy maxillary complete denture base plates fabricated by selective laser melting technique. Lasers in Medical Science. 2018.

191. Kanazawa M, Iwaki M, Minakuchi S, Nomura N. Fabrication of titanium alloy frameworks for complete dentures by selective laser melting. Journal of Prosthetic Dentistry. 2014; 112(6):1441-7.

192. Kim M-J, Choi Y-J, Kim S-K, Heo S-J, Koak J-Y. Marginal accuracy and internal fit of 3-d printing laser-sintered co-cr alloy copings. Materials. 2017; 10(1):93.

193. Hama Suleiman S, Vult von Steyern P. Fracture strength of porcelain fused to metal crowns made of cast, milled or laser-sintered cobalt-chromium. Acta Odontologica Scandinavica. 2013; 71(5):1280-9.

194. Dzhendov D, Dikova T. Application of selective laser melting in manufacturing of fixed dental prostheses. Journal of IMAB–Annual Proceeding Scientific Papers. 2016; 22(4):1414-7.

195. Kul E, Aladag LI, Duymus ZY. Comparison of the metal-ceramic bond after recasting and after laser sintering. Journal of Prosthetic Dentistry. 2015; 114(1):109-13.

196. V H, Ali SAM, N J, Ifthikar M, Senthil S, Basak D, et al. Evaluation of internal and marginal fit of two metal ceramic system - in vitro study. Journal of Clinical and Diagnostic Research. 2014; 8(12):Zc53-6.

197. Sundar MK, Chikmagalur SB, Pasha F. Marginal fit and microleakage of cast and metal laser sintered copings--an in vitro study. Journal of Prosthodontic Research. 2014; 58(4):252-8.

198. Tamac E, Toksavul S, Toman M. Clinical marginal and internal adaptation of CAD/CAM milling, laser sintering, and cast metal ceramic crowns. Journal of Prosthetic Dentistry. 2014; 112(4):909-13.

199. Iseri U, Ozkurt Z, Kazazoglu E. Shear bond strengths of veneering porcelain to cast, machined and laser-sintered titanium. Dental Materials Journal. 2011; 30(3):274-80.

200. Prabhu R, Prabhu G, Baskaran E, Arumugam EM. Clinical acceptability of metal-ceramic fixed partial dental prosthesis fabricated with direct metal laser sintering technique-5-year follow-up. Journal of Indian Prosthodontic Society. 2016; 16(2):193-7.

201. Abou Tara M, Eschbach S, Bohlsen F, Kern M. Clinical outcome of metal-ceramic crowns fabricated with laser-sintering technology. The International Journal of Prosthodontics. 2011; 24(1):46-8.

202. Bilgin MS, Erdem A, Dilber E, Ersoy İ. Comparison of fracture resistance between cast, CAD/CAM milling, and direct metal laser sintering metal post systems. Journal of Prosthodontic Research. 2016; 60(1):23-8.

203. Chen J, Zhang Z, Chen X, Zhang C, Zhang G, Xu Z. Design and manufacture of customized dental implants by using reverse engineering and selective laser melting technology. The Journal of Prosthetic Dentistry. 2014; 112(5):1088-95.e1.

204. Mangano F, Chambrone L, Van Noort R, Miller C, Hatton P, Mangano C. Direct metal laser sintering titanium dental implants: a review of the current literature. International Journal of Biomaterials. 2014.

205. Traini T, Mangano C, Sammons RL, Mangano F, Macchi A, Piattelli A. Direct laser metal sintering as a new approach to fabrication of an isoelastic functionally graded material for manufacture of porous titanium dental implants. Dental Materials. 2008 ;24(11):1525-33.

206. Leuders S, Thöne M, Riemer A, Niendorf T, Tröster T, Richard H, et al. On the mechanical behaviour of titanium alloy TiAl6V4 manufactured by selective laser melting: Fatigue resistance and crack growth performance. International Journal of Fatigue. 2013; 48:300-7.

207. Tan N, Liu X, Cai Y, Zhang S, Jian B, Zhou Y, et al. The influence of direct laser metal sintering implants on the early stages of osseointegration in diabetic mini-pigs. International Journal of Nanomedicine. 2017; 12:5433-42.

208. Mangano F, Mangano C, Piattelli A, Iezzi G. Histological evidence of the osseointegration of fractured direct metal laser sintering implants retrieved after 5 years of function. BioMed Research International. 2017; 9732136.

209. Garcia-Gareta E, Hua J, Orera A, Kohli N, Knowles JC, Blunn GW. Biomimetic surface functionalization of clinically relevant metals used as orthopedic and dental implants. Biomedical Materials. 2017; 13(1):015008.

210. Mumith A, Coathup M, Chimutengwende-Gordon M, Aston W, Briggs T, Blunn G. Augmenting the osseointegration of endoprostheses using laser-sintered porous collars: an in vivo study. The Bone and Joint Journal. 2017; 99-b(2):276-82.

211. Sumida T, Otawa N, Kamata YU, Kamakura S, Mtsushita T, Kitagaki H, et al. Custom-made titanium devices as membranes for bone augmentation in implant treatment: Clinical application and the comparison with conventional titanium mesh. Journal of Cranio-Maxillofacial Surgery. 2015; 43(10):2183-8.

212. Gahler A, Heinrich JG, Guenster J. Direct laser sintering of AL2O3–SiO2 dental ceramic components by layer-wise slurry deposition. Journal of the American Ceramic Society. 2006; 89(10):3076-80.

213. Mengucci P, Gatto A, Bassoli E, Denti L, Fiori F, Girardin E, et al. Effects of build orientation and element partitioning on microstructure and mechanical properties of biomedical Ti-6Al-4V alloy produced by laser sintering. Journal of the Mechanical Behavior of Biomedical Materials. 2017; 71:1-9.

214. Sun S-H, Koizumi Y, Kurosu S, Li Y-P, Matsumoto H, Chiba A. Build direction dependence of microstructure and high-temperature tensile property of Co–Cr–Mo alloy fabricated by electron beam melting. Acta Materialia. 2014; 64:154-68.

215. Benedetti M, Torresani E, Leoni M, Fontanari V, Bandini M, Pederzolli C, et al. The effect of post-sintering treatments on the fatigue and biological behavior of Ti-6Al-4V ELI parts made by selective laser melting. Journal of the Mechanical Behavior of Biomedical Materials. 2017; 71:295-306.

216. Alifui-Segbaya F, Evans J, Eggbeer D, George R. Clinical relevance of laser-sintered co-cr alloys for prosthodontic treatments: a review. 2014.

217. Xin XZ, Xiang N, Chen J, Wei B. In vitro biocompatibility of Co–Cr alloy fabricated by selective laser melting or traditional casting techniques. Materials Letters. 2012; 88(0):101-3.

218. Xin X, Chen J, Xiang N, Gong Y, Wei B. Surface characteristics and corrosion properties of selective laser melted Co–Cr dental alloy after porcelain firing. Dental Materials. 2014; 30(3):263-70.

219. Kruth J-P, Vandenbroucke B, Vaerenbergh vJ, Mercelis P. Benchmarking of different SLS/SLM processes as rapid manufacturing techniques. 2005.

220. Laser sintering high-quality dental components Cologne, Germany: Industrial lasers solution; 2011.

221. 3D Printing in Dentistry: Laser melting is the future. TCT Accelerating 3D Technology. 2015.

222. Bacchewar P, Singhal S, Pandey P. Statistical modelling and optimization of surface roughness in the selective laser sintering process. Proceedings of the Institution of Mechanical Engineers, Part B: Journal of Engineering Manufacture. 2007; 221(1):35-52.

223. Nakata T, Shimpo H, Ohkubo C. Clasp fabrication using one-process molding by repeated laser sintering and high-speed milling. Journal of Prosthodontic Research. 2017; 61(3):276-82.

224. CNBC.com. GE to buy SLM, Arcam for $1.4 billion in 3D printing push cnbc.com: cnbc.com; 2016.

225. HILL R. 3D systems to acquire phenix systems 3Dsystem.com: 3Dsystem.com; 2013.

226. Slotwinski J, Cooke A, Moylan S. Mechanical properties testing for metal parts made via additive manufacturing: a review of the state of the art of mechanical property testing. National Institute of Standards and Technology. 2012.

227. Kalpakjian S, Schmid S, Musa H. Mechanical behavior, testing, and manufacturing properties of materials. Manufacturing, Engineering & Technology, 6th ed, Pearson Education Inc, Upper Saddle River, NJ. 2008:56-87.

228. Roylance D. Mechanical properties of materials. Massachusetts Institute of Technology. 2008:51-78.

229. Kiefer WS, Macke RJ, Britt DT, Irving AJ, Consolmagno GJ. The density and porosity of lunar rocks. Geophysical Research Letters. 2012; 39(7).

230. Boerckel JD, Mason DE, McDermott AM, Alsberg E. Microcomputed tomography: approaches and applications in bioengineering. Stem Cell Research and Therapy. 2014; 5(6):144.

231. Abdallah MN. Percutaneous medical devices. 2017.

232. Stock SR. Microcomputed tomography: methodology and applications: CRC press; 2008.

233. Jenkins R. X-ray diffraction. X-Ray fluorescence spectrometry, Volume 152, Second Edition. 2012:37-51.

234. Gilmore CJ. X-Ray Diffraction. Solid State Characterization of Pharmaceuticals. 2011:35-70.

235. Khursheed A. Scanning electron microscope optics and spectrometers. 2011.

236. Pizzoferrato A, Ciapetti G, Stea S, Cenni E, Arciola CR, Granchi D, et al. Cell culture methods for testing biocompatibility. Clinical Materials. 1994;15(3):173-90.

237. Wataha JC. Principles of biocompatibility for dental practitioners. Journal of Prosthetic Dentistry. 2001; 86(2):203-9.

238. Geurtsen W. Biocompatibility of dental casting alloys. Critical Reviews in Oral Biology and Medicine. 2002; 13(1):71-84.

239. Wataha JC. Biocompatibility of dental casting alloys: a review. Journal of Prosthetic Dentistry. 2000; 83(2):223-34.

240. Weyermann J, Lochmann D, Zimmer A. A practical note on the use of cytotoxicity assays. International Journal of Pharmaceutics. 2005; 288(2):369-76.

241. Sato Y, Hamada S, Akagawa Y, Tsuga K. A method for quantifying overall satisfaction of complete denture patients. Journal of Oral Rehabilitation. 2000; 27(11):952-7.

242. Feine JS, Lund JP. Measuring chewing ability in randomized controlled trials with edentulous populations wearing implant prostheses. Journal of Oral Rehabilitation. 2006; 33(4):301-8.

243. Michaud P-L, de Grandmont P, Feine JS, Emami E. Measuring patient-based outcomes: Is treatment satisfaction associated with oral health-related quality of life? Journal of Dentistry. 2012; 40(8):624-31.

244. Reips U-D, Funke F. Interval-level measurement with visual analogue scales in Internet-based research: VAS Generator. Behavior Research Methods. 2008; 40(3):699-704.

245. De Grandmont P, Feine J, Tache R, Boudrias P, Donohue W, Tanguay R, et al. Within-subject comparisons of implant-supported mandibular prostheses: psychometric evaluation. Journal of Dental Research. 1994; 73(5):1096-104.

Chapter 6: Manuscript 1

The poor quality of the RPD is one of the main reasons for the complications and failures of removable partial denture treatment. To overcome this issue, the fabrication of RPDs using the laser sintering (laser melting) technique instead of casting technique could improve the quality of RPDs. This chapter is a published manuscript to characterize and understand the property of the RPD alloys that processed by the laser sintering (laser melting) technology in comparison with cast alloy. In this chapter, the term laser sintering was used, and it is referred to the laser melting technique. This manuscript including the references formatting and references list is identical to the published version.

Removable Partial Denture Alloys Processed by Laser-sintering Technique

Author names and affiliations:

Omar Alageel [1,2], Mohamed-Nur Abdallah [1], Ammar Alsheghri [3], Jun Song [3], Eric Caron [4], Faleh Tamimi [1,*]

[1] Faculty of Dentistry, McGill University, Montreal, QC, Canada

[2] College of Applied Medical Sciences, King Saud University, Riyadh, Saudi Arabia

[3] Department of Mining and Materials Engineering, McGill University, Montreal, QC, Canada

[4] 3DRPD Inc., Montreal, QC, Canada

This manuscript has been published at Journal of Biomedical Materials Research Part B: Applied Biomaterials, 106B: 1174–1185, 2018.

6.1. ABSTRACT

Removable partial dentures (RPDs) are traditionally made using a casting technique. New additive manufacturing processes based on laser-sintering has been developed for quick fabrication of RPDs metal frameworks at low cost. The objective of this study was to characterize the mechanical, physical, and biocompatibility properties of RPD cobalt-chromium (Co-Cr) alloys produced by two laser-sintering systems and compare them to those prepared using traditional casting methods. The laser-sintered Co-Cr alloys were processed by the selective laser-sintering method (SLS) and the direct metal laser-sintering (DMLS) method using the Phenix system (L-1) and EOS system (L-2), respectively. L-1 and L-2 techniques were 8 and 3.5 times more precise

than the casting (CC) technique ($p<0.05$). Co-Cr alloys processed by L-1 and L-2 showed higher ($p<0.05$) hardness (14-19%), yield strength (10-13%), and fatigue resistance (71-72%) compared to CC alloys. This was probably due to their smaller grain size and higher microstructural homogeneity. All Co-Cr alloys exhibited low porosity (2.1-3.3%); however, pore distribution was more homogenous in L-1 and L-2 alloys when compared to CC alloys. Both laser-sintered and cast alloys were biocompatible. In conclusion, laser-sintered alloys are more precise and present better mechanical and fatigue properties than cast alloys for removable partial dentures.

Keywords

Laser-sintering; Cobalt-chromium (Co-Cr); Removable partial dentures (RPDs); Fatigue resistance; Biocompatibility.

Running Heads:

Laser-sintered removable partial dentures (RPDs).

6.2. INTRODUCTION

Removable partial dentures (RPDs) are simple and cost-effective prostheses that can restore missing teeth in partially edentulous patients, and thus improving their quality of life [1, 2]. This type of treatment has an important impact on the life of millions of patients in the world; indeed, over 13% of the adult population in North America and Europe wear RPDs [1, 3]. RPD frameworks are commonly made of cobalt-chromium (Co-Cr) alloys because of their suitable cost and mechanical properties, as well as their excellent corrosion resistance and biocompatibility [4].

RPD frameworks are traditionally fabricated using the casting (lost-wax) technique that has been used in dentistry for more than a century [5, 6]. The casting technique is a very laborious manual process that involves making a wax replica of the object, making a mold of the object, and

then cast the melted metal into the mold. Due to its complexity, this technique is strongly influenced by the skill of the dental technician [5, 7]. Moreover, producing RPDs by casting technique is not only time consuming and costly but may also generate low precision and ill-fitting frameworks [7, 8].

Different methods were introduced in the last few decades for fabricating RPD frameworks without using casting techniques [6, 9, 10]. A new additive manufacturing (AM) process based on laser-sintering has been developed for processing 3D metal objects. The laser-sintering technique combines computer-aided design (CAD) of any products and their subsequent fabrication using a high-power laser that fuses metal powder in a layer-by-layer pattern [5, 6, 10-12]. The laser-sintering technique enables the fabrication of complex 3D objects quickly with high precision (20 µm) and at low cost (Figure 6.1) [10-15].

Laser-sintering technology can be described using different terminologies, such as selective laser melting (SLM), selective laser-sintering (SLS), or direct metal laser-sintering (DMLS) [6, 9, 12, 13]. SLM involves full melting of the metal powder; while, both SLS and DMLS involve partial melting of some the metal powder, particularly melting at the surface of the particle [12-14]. The main difference between SLS and DMLS is that SLS powder can be metal or other materials (e.g. ceramic or polymer), and the powder only partially melts during the process [12-16]. Whereas, DMLS uses a mixture of metal powders with different melting temperatures (high or low) [12-14, 16-18]. During the DMLS process, the powder with the low melting temperature fully melts while the powder with high melting temperature partially melts [12, 13, 16]. In this study, we used two systems that are commercially available for dental applications; the Phenix system (Phenix, Riom, France) that is based on the SLS method, and the EOS system (EOS, Krailling, Germany) that is based on the DMLS approach [12-16].

Fabricating RPDs using the laser-sintering technique, instead of casting technique, could increase the quality of RPDs and render the treatment less expensive and more accessible to a larger portion of the population [6]. However, the fabrication of Co-Cr RPDs by laser-sintering technology can affect the mechanical, physical and biocompatibility properties of the alloys and subsequently affect the clinical performance of RPDs [8, 19, 20]. The properties of laser-sintered alloys can be influenced by differences in the fabrication process, such as laser beam power, scanning speed, metals powder size, and layer thickness [8, 19-21].

The mechanical property, such as elastic modulus and bending yield strength, is crucial for RPD because it prevents clasps, the retentive element of RPD, from catastrophic failure during the repetitive cycles of insertion and removal of the dentures from the mouth [22, 23]. However, there is no data currently available on fatigue resistance of laser-sintered RPD alloys. Previous studies evaluated the physical properties including microstructure, corrosion resistance and solubility of laser-sintered Co-Cr alloys for other applications [8, 20, 21, 24, 25]. These studies showed that laser-sintered alloys had better physical properties than cast Co-Cr alloys. In addition, the biocompatibility of cast Co-Cr alloy has been previously investigated [26]. Although the Co-Cr alloys produced by DMLS (EOS) system is certified by the ISO 9693 and ISO 10993, the biocompatibility of RPD Co-Cr produced by SLS (Phenix) system remains unknown [8, 27]. Therefore, the objective of this study was to characterize and understand the mechanical properties, physical properties, and biocompatibility of RPD Co-Cr alloys produced by two different laser-sintering systems and compare them to those made by the traditional casting method.

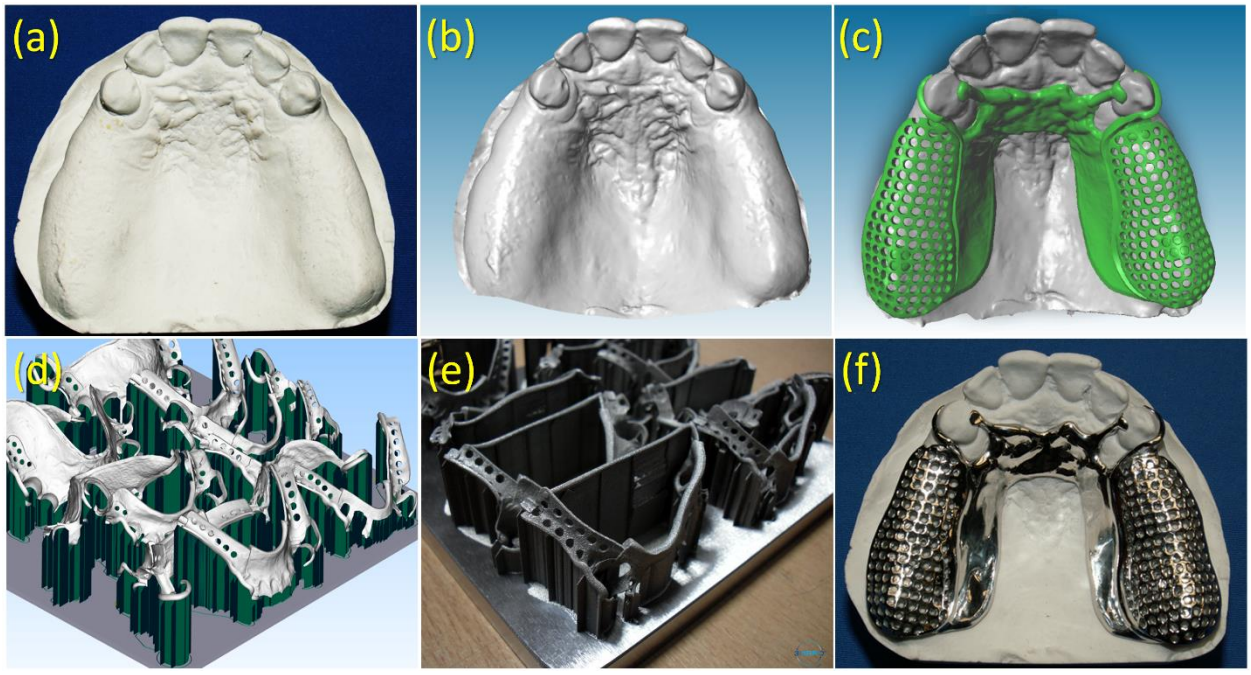


Figure 6.1. Photographs illustrating the process of designing and fabricating removable partial dentures (RPDs) framework using laser-sintering technique; (a) master cast of partially edentulous arch, (b) 3D scan of the model, (c) designing of RPDs framework, (d) placing RPDs frameworks in a digital platform, (e) processed RPDs frameworks in the producing platform, (f) the final RPD framework.

6.3. MATERIALS AND METHODS

6.3.1. Sample preparation

All experiments were performed using dental Co-Cr alloys. The chemical composition of the Co-Cr ingot and powder as provided by the manufacturers is listed in Table 6.1. Co-Cr samples were fabricated by conventional casting (CC group), selective laser-sintering (SLS) method (L-1 group), and direct metal laser-sintering (DMLS) method (L-2 group) at the prototyping center (3DRPD Inc., Montreal, QC, Canada). Samples were prepared in different geometries according to the property under investigation. All samples were designed using a CAD system. The cast alloys (CC group) were fabricated following similar steps used for fabricating traditional RPD.

The wax-ups of the CC samples were printed in a plastic form using a 3D printer (UtraHD, EnvisionTEC, Dearborn, MI), invested in metal casting rings, and cast using an automatic vacuum-pressure casting machine with induction heating (Nautilus CC, BEGO, Bremen, Germany) using an ingot form of Co-Cr alloy (NobilStar Ultra; Nobilium, Albany, NY).

Two laser-sintering systems equipped with their specified Co-Cr powders were used to fabricate the L-1 and L-2samples. L-1 samples were processed by the selective laser-sintering (SLS) technology using the PXM system (Phenix Systems, Riom, France) with 300W of Fibre laser power and wavelength of 1070 nm and equipped with a roller to compact the powder. The particle size of the Co-Cr powder in the L-1 group (ST2724G-A, Sint-Tech; Clermont-Ferrand, France) as observed by the scanning electron microscope (SEM) was 6-22 µm. On the other hand, L-2 samples were processed using the EOSINT M270 system (EOS, Krailling, Germany), which is based on the direct metal laser-sintering (DMLS) approach, with 200W of Fibre laser power and 1064 nm wavelength. The average particle size of the Co-Cr powder (CobaltChrome SP2, EOS; Krailling, Germany) in the L-2 group was 20 µm [28]. For both L-1 and L-2 samples, the layer thickness, the laser scan speed and building direction were 30 µm, 5-7 m/s, and 90º respectively. Post-processing heat treatments were applied to the L-1 and L-2 samples according to the manufacturers' instructions (L-1: 800 ºC for 30 minutes; L-2: 450 ºC for 45 minutes, 750 ºC for 60 minutes and then cooled down). Alloy samples for microhardness and crystallography were manually polished to produce a mirror-like surface using a six-step polishing process [29].

Table 6.1. The manufactures chemical composition of the cast (CC) and laser-sintered (L-1 and L-2) Co-Cr alloys.

Mass %	Cobalt (Co)	Chromium (Cr)	Molybdenum (Mo)	Silicon (Si)	Manganese (Mn)	Iron (Fe)	Tungsten (W)
CC	64	28.5	5.3	< 1	< 1	< 1	-
L-1	63	29	5.5	< 1	< 1	< 1	-
L-2	64	25	5.1	1	< 1	< 1	5

6.3.2. Precision error calculation

The dimension of the samples used for toughness analysis were measured with an electronic caliper (Fower, Newton, MA) in order to calculate the precision error of CC, L-1 and L-2 techniques. For each sample, the dimension of each sample was measured for all the 3 side surfaces (length, width, thickness). Then, the dimensional changes between the processed and the CAD designed samples were calculated.

6.3.3. Mechanical characterization

Three-point bending tests, which are mimicking the fracture of RPDs clasps, were done at room temperature using a universal testing machine (Instron, 5569, Grove City, PA) to characterize the mechanical properties of CC, L-1, and L-2 alloys. Each sample (n=9) was placed on two supporting pins 18 mm apart of each other. Loading was applied through an actuator by moving the loading pin at a constant speed of 1 mm/min on the middle of the specimen until failure. The testing machine then provided a force/deflection curve for each sample through the Bluehill v.2 software (Instron, Grove City, PA). The elastic modulus (E), bending yield strength (σ_y), flexural strength (σ_F), and fracture toughness (K_{1C}) values were calculated using the following equations (1-5) [30-32].

$$E = \frac{F_{max} L^3}{4\delta b d^3} \quad (1)$$

$$\sigma_y = \frac{3F_y L}{2bd^2} \quad (2)$$

$$\sigma_F = \frac{3F_{max} L}{2bd^2} \quad (3)$$

$$K_{IC} = \frac{3F_{max} L}{2bd^2\, Y\left(\frac{a}{b}\right)\sqrt{a}} \quad (4)$$

where $$Y\left(\frac{a}{b}\right) = 1.93 - 3.07\left(\frac{a}{b}\right) + 14.5\left(\frac{a}{b}\right)^2 + 25.1\left(\frac{a}{b}\right)^3 + 25.8\left(\frac{a}{b}\right)^4 \quad (5)$$

where F_y is the yield force, F_{max} is the maximum force applied, L is the distance between the supports, δ is the deflection of the tested specimen, b is the width of the tested specimen, d is the height of the tested specimen, and a is the notch depth.

In order to determine the fatigue resistance of the CC, L-1 and L-2 alloys, 6 rectangular specimens from each group were exposed to a cyclic of three-point bending loading and unloading up to 6000 cycles. At each cycle, the load was applied at a constant speed of 15 mm/min until reaching a constant deflection of 0.2 and 0.1 mm, since these deflections are similar to the depth of undercuts on the abutment tooth surface where the RPD clasps engage with [2]. Then, the loading was repeated at a frequency of 5 Hz, and the force change (N) was recorded at each cycle. Finally, the post-fatigue force was compared to their initial force in order to evaluate the fatigue resistance.

A Vickers microhardness indenter (Clark CM100 AT, HT-CM-95605, Shawnee Mission, KS) was employed on the polished surfaces of the Co-Cr samples. Tooth enamel sections fixed in resin blocks were also analyzed for comparison [33, 34]. Nine measurements were obtained per specimen (n=3) under indentation load of 500 g for 10 seconds of dwell time. Computer software (Vision PE 3.5, Clemex Technologies Inc., Shawnee Mission, KS) was used to measure the microhardness value at the site of indentation from images captured by a built-in camera.

6.3.4. Physical characterization

Density and porosity of the CC, L-1 and L-2 alloys were analyzed using five samples per group. The bulk density, which includes the volume of pore spaces in the alloys, was calculated by dividing the sample's weight by its volume. While, the real volume and real density (grain density), which do not include the pore spaces in the alloys, were measured using helium pycnometry (Accupyc 1330; Micromeretics; Bedfordshire, UK). Helium pycnometry measures the

gas pressure in a calibrated chamber before and after insertion of the specimen into the chamber. Porosity percentage was calculated using the equation below (6).

$$Porosity\ percentage\ = \frac{(bulk\ volume\ -\ real\ volume)}{bulk\ volume} \times 100 \quad (6)$$

In order to further analyze the porosity of the CC, L-1 and L-2 alloys, the specimens were scanned using a high-resolution micro-computed tomography (µ-CT). The µ-CT (SkyScan 1172; SkyScan; Kontich, Belgium) was set at a resolution of 11.56 µm, a voltage of 100 kV, a current of 100 µA, and an aluminium (Al+Cu) filter of 0.5 mm. The total rotation angle of the sample was 360° with a rotation step size angle of 0.4°. Data were reconstructed using standardized cone-beam reconstruction software (NRecon v.1.6.9, SkyScan). The 3D modeling and analysis involving porosity percentage, number and volume of pores, and degree of anisotropy were performed using CTAn v.1.13 (SkyScan Kontich, Belgium). The 3D images were performed with the software CTvol v.2.2.3 (SkyScan Kontich, Belgium).

X-ray diffraction (XRD) was used to analyze the crystallography of the alloys and the L-1 powders. The experiment was performed using X-ray diffraction (D8-Discover/GADDS, Bruker, Karlsruhe, Germany) with a cobalt source radiation set at 40 kV and 40 mA, a 10-60^{o} scanning angle, 0.02^{o} step size, 1 second scan step time, and an integration time of 120 seconds. The EVA v.14 software (Bruker AXS, Karlsruhe, Germany) was used for phase identification and crystal size calculations following Scherrer's formula.

Fractured surface of CC, L-1, and L2- specimens were observed with scanning electron microscopy (SEM) (FE-SEM, FEI, Hillsboro, OR). SEM micrographs of the samples fractured by three-point bending were taken at x2500 magnification. The SEM was operated at 5-10 kV accelerating voltage, a spot size of 2-3 µm, and a working distance of 9.2-10.1 mm. To perform SEM back-scattered electron imaging, the polished samples were etched for 30 seconds in order

to reveal both the macrostructure and microstructure of the welds. The chemical etch was composed of a solution of 80% of hydrochloric acid (HCL) and 20% hydrogen peroxide (H_2O_2) (v/v) (Sigma Aldrich, St Louis, MO) [35]. The SEM back-scattering images were obtained with a SEM operated at 20 kV, spot size of 3 μm, and a working distance of 9.5-9.3 mm. The micrographs were taken at x200 magnification.

6.3.5. Biocompatibility assays

The releases of toxic metal ions from CC and L-1 alloys were measured using inductively coupled plasma atomic emission spectroscopy (ICP-AES; Perkin-Elmer, Wellesley, Mass). Each sample (n=6) was immersed in 5 ml of PBS (phosphate buffered saline) simulating artificial saliva and incubated for 7 days at 37°C [25]. The extracted solution was digested with 2 ml of concentrated nitric acid and then diluted in 7 ml of deionized water. Standard solutions containing Co, Cr and Mo elements were prepared for calibration at a concentration of 2.0, 1.0, and 0.5 ppm (parts per million). Triplicate absorbance readings for all the above elements were recorded from each sample to determine the concentration of the elements released from the alloys in parts per million and these measurements were converted to units of μg per cm^2.

The cell biological response was studied in vitro using equally sized CC and L-1 specimens (n=9) according to the international standard ISO 10993-5. Human gingival epithelial cells (HGEs) (Cedarlane Laboratories, Ontario, Canada) were cultured for 1, 3, and 7 days in serum free CnT-Prime medium (Cedarlane Laboratories, Ontario, Canada) in a 20% O_2 and 5% CO_2 humidified atmosphere at 37°C. HGEs were seeded at a density of $3*10^4$ cells/cm^2 on the bottom of 24 well plates (Transwell, Costar, Corning, NY) while the Co-Cr metals were hanged on the middle of the well plates.

Two independent test kits for cell viability (Alamar Blue, Life technologies, Ontario, Canada) and cytotoxicity (CytotoxOne, PROMEGA, Wisconsin, USA) were combined to analyze the two different parameters from one single sample. After 24 hours of cell culture, triplicates of 100 µl of the supernatant was transferred into a Microfluor 96-well fluorescence plate with clear bottom (Thermo Fisher Scientific, Waltham, MA) and mixed with 100 µl of cytotoxicity reagent. After an incubation of 10 min, the lactate-dehydrogenase (LDH) release from damaged cells was analyzed by Spectra Max M2E (Molecular Device, Sunnyvale, CA). Then, 400 µl of cell culture media containing 10% of Alamar blue was added to each well and incubated for 4 hours. The supernatant was then transferred into a 96-well plate with a clear bottom to detect the cell metabolism using a microplate reader (Spectramax M2E, Molecular Devices, CA, USA). For viability and cytotoxicity, the excitation and emission were 560 and 590 nm, respectively. Cells seeded without any metal exposure were considered as positive control. Metals without cells were incubated in parallel to serve as controls to remove the background fluorescence. Cell lysed with lysis solution provided with the cytotoxicity kit (2 µl/100 µl) were used to determine the maximum LDH release. The viability and cytotoxicity assays were repeated as described above; at 3 and 7 days in triplicates.

The live/dead staining assay for assessing the cytocompatibility of the metals was performed only at 24 hours after seeding. The assay consisted of fluorescein diacetate (FDA) (Sigma, Steinheim, Germany, 5 mg in 1 ml of acetone) and propidium iodide (PI) (Sigma, Steinheim, Germany, 2 mg in 1 ml PBS). After removing the culture media from the well plates, a freshly prepared staining solution was added to each well and incubated for 5 min. Cells were analyzed and captured under the Zeiss AX10 fluorescence microscope (Carl Zeiss, Göttingen, Germany). Green fluorescence indicates viable cells and red fluorescence dead cells.

6.3.6. Statistical analysis

Statistical analysis for the mechanical and biocompatibility data was performed with the software Origin 8.0 (Origin lab, Northampton, MA). Mean and standard deviation (SD) values were calculated for all measurements. A one-way analysis of variance (ANOVA) and Tukey HSD multiple comparison test was used to test for statistical differences of mechanical and physical properties between the CC, L-1, and L-2 groups. The statistical differences for the biocompatibility test between CC and L-1 groups were tested using Student's t-test. The significance level was set at $p < 0.05$.

6.4. RESULTS

Table 6.2 provides an overview of all the results obtained from the mechanical and physical characterization techniques. The precision error that was calculated based on the dimensional comparison between the CAD designed specimen and the fabricated one indicated that both laser-sintering techniques (L-1 and L-2) were up to 8 times more precise than the conventional casting technique (CC) ($p<0.05$). The precision error of CC samples was 9.3±6.5%, while the precision errors of L-1 and L-2 were 1.2±1.2% and 2.9±2.5%, respectively. The results of the three-point bending test indicated that the elastic modulus of the L-1 alloys were significantly lower (202±16 GPa) than the CC alloys (229±7 GPa) and the L-2 alloys (225±10 GPa). The bending yield strength of the L-1 and L-2 alloys were significantly higher (1626±118 and 1685±109 MPa, respectively) than the CC alloys (1462±142 MPa). The flexural strength and fracture toughness of the L-1 alloys (2837±97 MPa and 61.2±2.1 1 $MPa*m^{1/2}$) were significantly higher than the CC (2647 ±208 MPa and 57.1±4.5 $MPa*m^{1/2}$) and L-2 alloys (2602±106 MPa and 56.1±2.3 1 $MPa*m^{1/2}$).

Both L-1 and L-2 samples presented higher ($p<0.05$) resistance to fatigue than the CC samples after 6000 stress cycles simulating the insertion and removal of the dentures from the mouth for 5 years (Figure 6.2 and Table 6.2). The fatigue resistance tests of the L-1 and L-2 groups showed that they maintained 91.1% and 89.6% of their original stress, respectively, at 0.2 mm deflection. Whereas, the CC group maintained only 25.4% of the original stress (Figure 6.2 and Table 6.2). Similar fatigue resistance behaviour was recorded at 0.1 mm deflection for all groups.

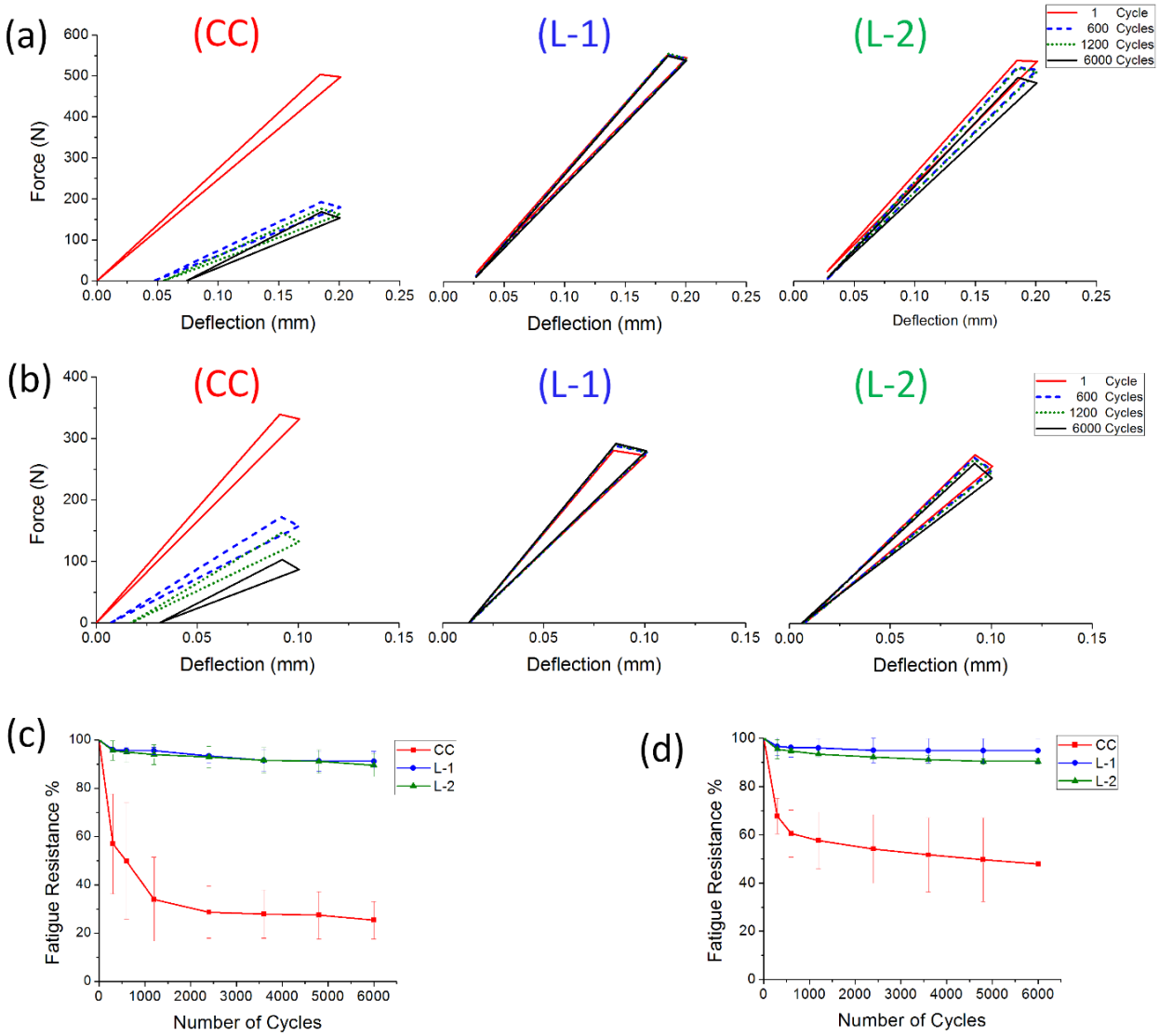


Figure 6.2. Load/deflection diagrams showing cycles of loading and unloading the different alloys for a deflection of 0.2 mm (a) and 0.1 mm (b). Percentage of the fatigue resistance comparing the post-fatigue force with the initial force at a deflection 0.2 mm (c), and 0.1 mm (d).

Table 6.2. Results (mean ± SD) of mechanical tests for the cast (CC), laser-sintered (L-1 and L-2) cobalt-chrome alloys.

		CC	L-1	L-2
Technique precision error	(%)	9.5 (±6.5)	1.2 (±1.2) *	2.9 (±2.5) *
Elastic modulus	(GPa)	229 (±7)	202 (±16) * †	225 (±10)
Bending yield strength	(MPa)	1462 (±142)	1626 (±118) *	1686 (±109) *
Flexural strength	(MPa)	2647 (±208)	2837 (±97) * †	2602 (±106)
Fracture toughness	K_{1C} (MPa*$m^{1/2}$)	57.1 (±4.5)	61.2 (±2.1) * †	56.1 (±2.3)
Fatigue resistance	0.2 mm deflection (%)	25.4 (±7.7)	91.1 (±4.2) *	89.6 (±4.7) *
	0.1 mm deflection (%)	47.9 (±18.3)	94.9 (±5.0) *	90.4 (±1.0) *
Hardness	(HV)	390 (±11) ‡	453 (±9) * † ‡	483 (±24) * ‡
Density	Bulk (g/mm^3)	8.2 (±0.1)	8.0 (±0.0) * †	8.4 (±0.1)
	Real (g/mm^3)	8.5 (±0.0)	8.3 (±0.1) * †	8.6 (±0.0)
Porosity	(%)	2.2 (±0.7)	4.1 (±1.0) *	3.8 (±1.2) *

* indicates a significant difference to cast (CC) group ($p<0.05$). † indicates a significant difference between L-1 and L-2 groups ($p<0.05$). ‡ indicates a significant difference to hardness of teeth enamel ($p<0.05$).

The microhardness values of the L-1 and L-2 alloys (453±9 and 477±14 HV, respectively) were higher ($p<0.05$) than CC alloys (390±11 HV), and the L-2 group had higher microhardness than L-1 group ($p<0.05$) (Table 6.2). However, all alloys had higher microhardness than tooth enamel (353±40 HV). CC and L-2 alloys had similar bulk density and real density; however, the density of the L-1 alloys was lower than CC and L-2 alloys ($p<0.05$). In addition, the L-1 alloys had a higher total porosity than the CC alloys ($p<0.05$), but not the L-2 alloys (Table 6.2). The porosity of L-1 and L-2 alloys was mainly closed porosity, and more isotropic ($p<0.05$) than in the CC alloys (Table 6.3 and Figure 6.3).

Table 6.3. µ-CT scan analysis results for the cast (CC), laser-sintered (L-1and L-2) Co-Cr alloys. Values presented as mean ± SD

		CC	L-1	L-2
Number of pores		4610 (±693)	5507 (±573) *	5522 (±644) *
Closed porosity percentage	(%)	1.63 (±0.56)	2.63 (±0.72) *	1.96 (±0.44) *
Open porosity percentage	(%)	0.50 (±0.44)	0.69 (±0.39)	0.64 (±0.46)
Total volume of pores	(mm^3)	0.02 (±0.01)	0.03 (±0.01) * †	0.02 (±0.01)
Total porosity percentage	(%)	2.13 (±0.85)	3.32 (±1.06) *	2.60 (±0.80)
Degree of porosity anisotropy		0.71 (±0.04)	0.67 (±0.03) * †	0.47 (±0.11) *

* indicates a significant difference to cast (CC) group ($p < 0.05$). † indicates a significant difference between L-1 and L-2 groups ($p < 0.05$).

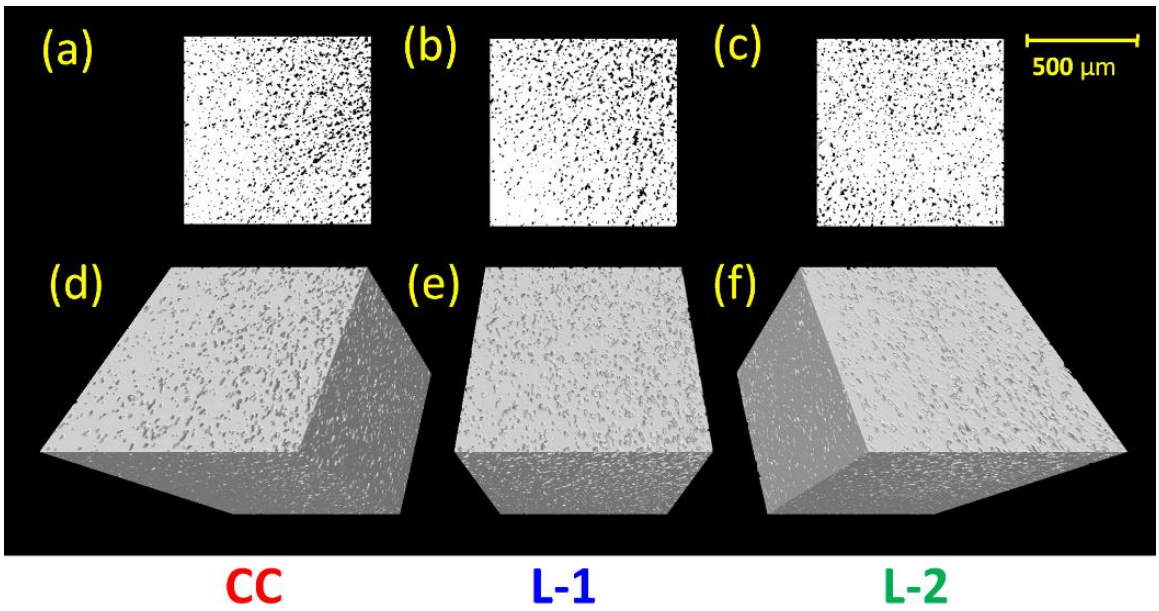


Figure 6.3. 2D images (a-c) and 3D images (d-f) by µ-CT showing the porosity of CC, L-1 and L-2 Co-Cr alloys.

The XRD crystallographic analysis (Figure 6.4) of the structure of CC, L-1, L-2 alloys, and L-1 powders revealed that the face-centred cubic (fcc) phase, which is characteristic of Co-Cr, was present in all groups as evident in Figure 6.4. However, the L-1 alloys showed an additional hexagonal close-packed (hcp) phase of Co-Mo that was not present in the other groups. The XRD spectra showed that the crystal size of CC (16.3±2.2 nm) and L-2 (16.2±2.1 nm) alloys was similar,

but larger than the crystal size of both the L-1 alloys (14.6±1.1 nm) and the L-1 powder (14.3±1.8 nm) ($p<0.05$).

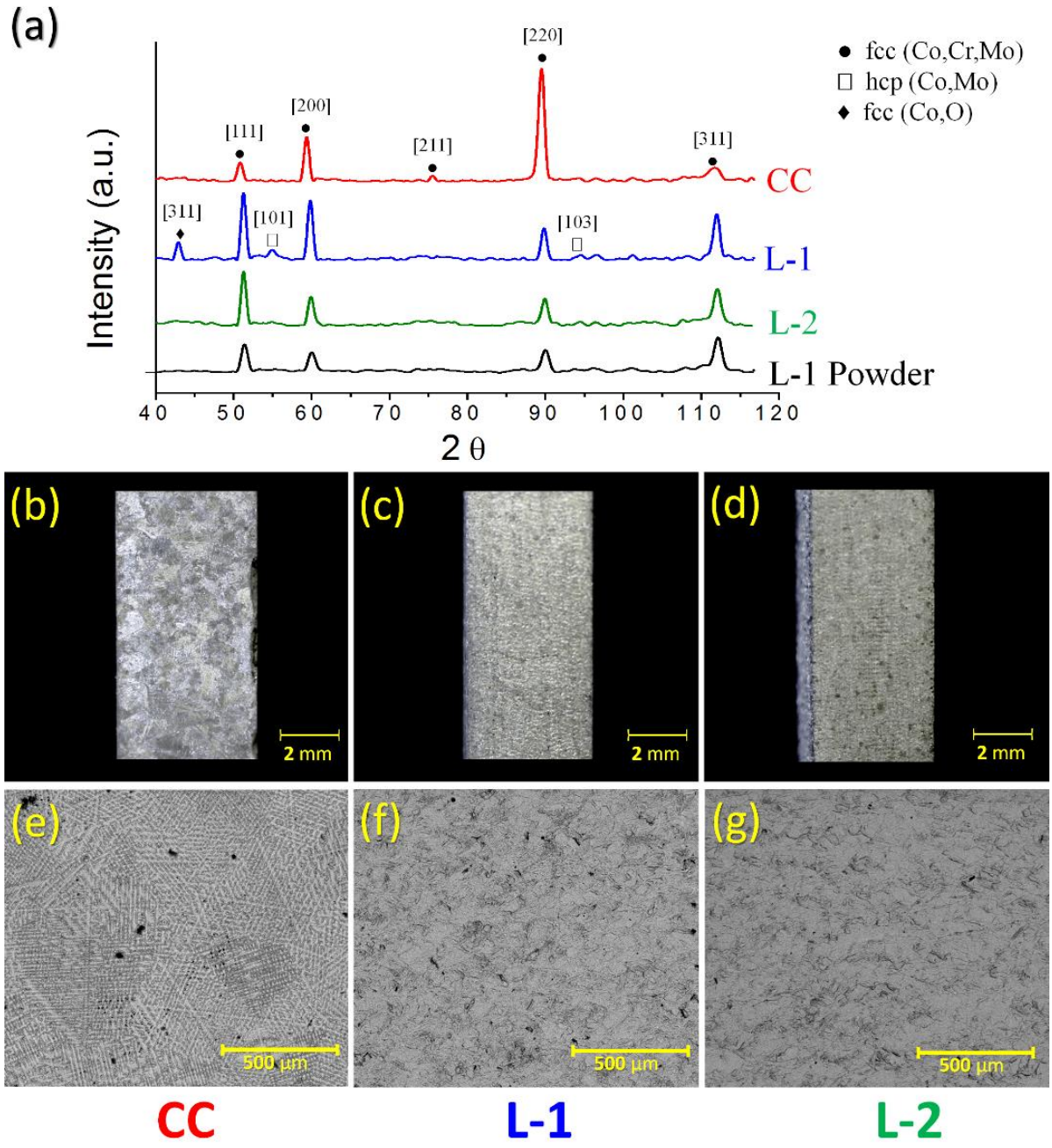


Figure 6.4. (a) Representative XRD spectra of the cast (CC), laser-sintered (L-1 and L-2) Co-Cr alloys as well as the metal powder used for laser-sintering, (b-d) digital photograph images and (e-g) SEM Back-scattering images on polished surfaces of the CC, L-1, and L-2 alloys.

The digital photographs and SEM back-scattered electron images of the polished surfaces of CC, L-1, and L-2 alloys are shown in Figure 6.4. The polished surfaces of the CC alloys revealed large grains while the polished surfaces of the L-1 and L-2 alloys exhibited a fine microstructural appearance. SEM observations of fractured samples from the different alloys are demonstrated in Figure 6.5. These SEM images revealed that the L-1 and L-2 alloys present an organized stop-like fracture path, whereas the CC alloys demonstrate an unorganized fracture path.

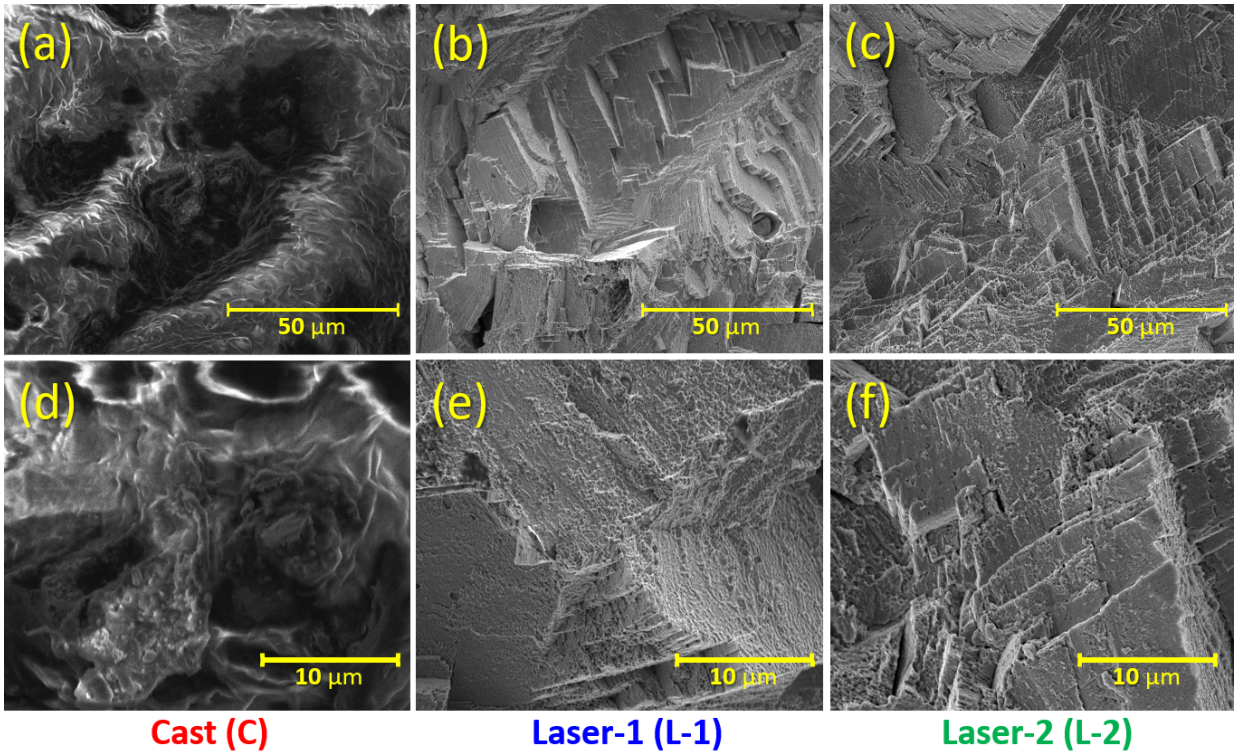


Figure 6.5. Representative SEM micrograph at the fractured surfaces of the CC, L-1, and L-2 alloys at magnifications of x2500 and x10000.

Biocompatibility assays showed that both L-1 and CC alloys had comparable behaviours (Figure 6.6, Table 6.4). Overall, the trace amounts of elements released from CC and L-1 alloys were within a small range. Both L-1 and CC alloys released comparable amounts of Co, Cr and Mo, but only the release of Co was significantly higher in the L-1 alloys when compared to the CC alloys. The percentage of cell activity (relative to control cells unexposed to metal) of the two

groups are illustrated in Figure 6.6. The viability and cytotoxicity of cells exposed to the L-1 and CC alloys declined over time up to 7 days in comparison to cells not exposed to Co-Cr alloys. However, no statistical difference was found between L-1 and CC alloys in terms of the viability and cytotoxicity. Figure 6.6 c-f shows the results of the live/dead assays of cells cultured for 24 hours and exposed to L-1 and CC alloys.

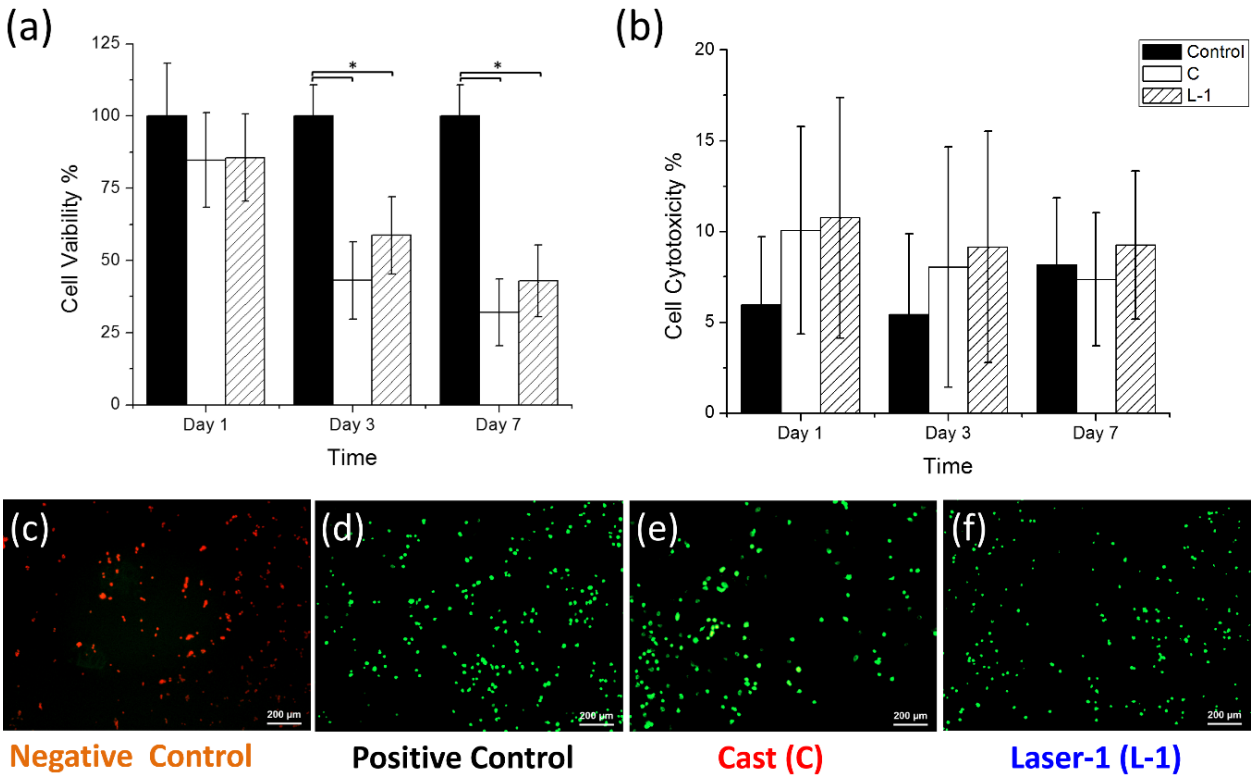


*Figure 6.6. Percentage of (a) mean cell viability, and (b) cell cytotoxicity (relative to control) of CC and L-1 alloys. Error bars represent SD and * indicates a significant difference between the different groups ($p < 0.05$). (c-f) live/dead staining results after 24 h of incubation for CC and L-1 alloys showing a higher number of live cells present (green) in compare to dead cells (red) in the negative control.*

Table 6.4. Element release in μg per cm^2 (mean ± SD) from cast (CC) and laser-sintered (L-1) Co-Cr alloys.

	Cobalt (Co)	Chromium (Cr)	Molybdenum (Mo)
CC	0.699 (±0.392)	0.007 (±0.027)	0.065 (±0.079)
L-1	1.196 (±0.044) *	0.005 (±0.025)	0.136 (±0.021)

* indicates a significant difference to cast (CC) group ($p < 0.05$).

6.5. DISCUSSION

This study was designed to characterize the mechanical, physical, and biocompatibility properties of laser-sintered RPD Co-Cr alloy and compare them to those of the cast RPD Co-Cr alloys. In this study, the Co-Cr alloys were fabricated using two commercially available systems, Phenix and EOS, which are based on two different laser-sintering methods, SLS and DMLS, respectively. The materials and processing parameters were specified for each system according to the respective manufacturer's instructions, and they were not exactly the same (Table 6.1). This might render the comparison difficult. However, the differences in chemical composition between the CC group and L-1 group were very small, and this would suggest that the characterization differences observed between the groups were most probably due to the processing approach of each system.

To the best of our knowledge, this is the first study assessing the fatigue resistance of laser-sintered Co-Cr alloys, as compared to that of the cast Co-Cr alloy. Fabricating RPDs by laser-sintering technology can have an economic impact on the way RPDs are made as well as improve the quality of RPDs. This will have a high impact on the millions of patients around the world who wearing RPDs. In fact, it was found in this study that the laser-sintering technique was 6 to 8 times more precise and 3 times more accurate than the casting technique. It was suggested that the high precision of laser-sintering technique was due to reduced number of accumulated of errors that

occur at the different steps during casting process. Although there was no a significant difference between the precision error of L-1 and L-2 techniques, the L-1 samples, which were processed by the Phenix system, tend to be more accurate than the L-2 group processed by the EOS system. The reason for these accuracy discrepancies between L-1 and L-2 could most probably be related to the features of the system since only L-1 (Phenix system) use a roller to compact the powder.

6.5.1. Mechanical properties

Clasp failure, which is the retentive elements engaging the teeth, is the most common complication of RPDs, and it is the main reason why most RPDs are replaced after 5 to 6 years of use [22, 23]. These failures are caused by the excessive and repeated stress on clasps during insertion and removal of the dentures from the mouth [22, 23]. This repeated stress might also result in fatigue failure and deformation of the RPD clasps, which eventually lead to the loss of retention [23]. For this reason, in our study, the three-point bending, and fatigue tests were performed to simulate the long-term function of the RPDs in the patients' mouth during insertion and removal of the dentures from the mouth.

The results obtained from the three-point bending test (Table 6.2) demonstrated that the L-1 alloys have a lower elastic modulus than the CC and L-2 alloys. This analysis indicates that L-1 alloys are more flexible and less stiff than CC and L-2 alloys. The elastic modulus of L-1 is closer to that of teeth (80–94 GPa [36]) than CC and L-2. This lower stiffness can be an advantage because it could minimize damage inflicted to the underlying teeth when fabricating RPDs made of L-1 alloys [37]. The reason for that is when the stiffness of an RPD framework surpasses that of supporting tissues (e.g. teeth), high-stress concentration accumulates at the metal-tissue interface resulting in fracture of the weaker component, which is the tooth in this case [38]. However, stiff Co-Cr alloys are favourable for RPDs components that require high stiffness, such as rests and

connectors, in order to prevent distortion and deflection of the dentures [22]. Therefore, L-1 alloys would be the more favourable for fabricating the RPDs' clasps, but less favourable than L-2 and CC alloys for fabricating the other RPDs components, such as rests and connectors.

The bending yield strength of both L-1 and L-2 alloys were higher than CC alloys. The bending yield strength is considered the most important mechanical property for RPDs since higher values of this strength helps to resist the plastic (permanent) deformation of RPD's clasps, and thus preventing their failure [39]. In addition, the flexural strength and fracture toughness of the L-1 group were higher than the CC and L-2 groups. Processing Co-Cr alloy by the DMLS method (L-2) involves full melting of some metal powder which makes it closer to the casting method than SLS method. This might be the reason of mechanical properties similarity between L-2 and CC alloys. Generally, L-1 alloys have better mechanical properties for RPDs than other alloys in term of the elasticity and strength, and this might be related to their porosity and microstructure which will be discussed later [40].

The fatigue resistance test was performed to simulate the insertion and removal of the dentures from the mouth for the period of 5 years (Figure 6.2 and Table 6.2) [41]. Clasps of the RPDs usually engage undercuts on the abutment tooth surface that are 0.25 mm deep [2]. To release a clasp from the abutment tooth, the clasp arm is bent to reach a deflection that equals to the depth of the undercut. This was simulated in the performed fatigue resistance test by bending the alloys to deflections of 0.1 and 0.2 mm. The results of the fatigue resistance test showed that alloys processed by laser-sintering technique (L-1 and L-2) had higher resistance to fatigue than those fabricated by the casting technique. In addition, the L-1 and L-2 alloys maintained most of their original mechanical properties after the fatigue cycles, whereas the CC alloys underwent a dramatic deformation, which was even more pronounced after fatigue cycles of larger deflections

(Figure 6.2). Based on our in vitro study, the average survival rate of the laser-sintered RPDs would be much higher than that of the cast RPDs, which is reported to have an average survival rate of 5.5 years [23].

The high fatigue resistance of L-1 and L-2 alloys is attributed to their high bending yield strength that allows higher resistance to plastic deformation when compared to CC alloys (Table 6.2) [39]. As suggested by Koutsoukis et al., the high bending yield strength and fatigue resistance of laser-sintered alloys could be attributed to their crystallinity and homogeneous microstructure (Figures 6.4 and 6.5) [6]. The SEM observation (Figure 6.5) of the CC, L-1 and L-2 alloys at the fractured surfaces showed that the L-1 and L-2 alloys were more homogenous than the CC alloys. The fine microstructure of the L-1 and L-2 alloys was due to rapid solidification of the melted powder, while the irregular microstructure of the CC alloy could be probably due to the internal defects and impurities that occur during the casting technique [6, 25]. As a result of the homogenous microstructure of the L-1 and L-2 alloys, wedge-type cracks and organized fracture paths were observed in L-1 and L-2 alloys, while unorganized fracture paths were observed in the CC alloys (Figure 6.5). Having a homogenous microstructure is beneficial for reducing the failures of RPDs clasps because it promotes homogeneous slip deformation, which in turn reduces the residual stresses and stress concentrations, as opposed to the nonhomogeneous microstructure observed in the CC samples [6].

Both types of laser-sintered alloys (L-1 and L-2) were significantly harder than the cast (CC) alloy (Table 6.2). High hardness values are desirable in the RPDs for resisting the scratching of the metallic alloy [42]. However, all the alloys tested in this study were harder than tooth enamel, and this might damage the teeth during insertion and removal the RPDs from the mouth. It should be noted that tooth damage can still occur due to the friction between the harder metal and the

softer tooth tissue even in the presence of saliva in the oral cavity that acts as a lubricant. One way to tackle this potential problem is to use metallic alloys with an appropriate elastic of modulus, since the friction depends on the force exerted by the clasp on the tooth. Accordingly, the lower elastic modulus of the L-1 alloy could palliate the negative consequence of its high hardness of Co-Cr alloys.

6.5.2. Physical properties

The porosity percentage of all the alloys tested in this study was minimal. Usually, the cast metals present high porosities and internal defects due to gas inclusion during the fabrication process [6, 22]. However, the minimal porosity of the CC alloys observed in this study could be attributed to the flat geometry of the samples and their relative small size, which might have reduced the gas inclusion. In addition, the porosity percentages for both laser-sintered alloys were minimal because of the post-heat treatment that was applied to the alloys after processing [12, 43]. It is known that the porosity of the laser-sintered alloys can be influenced and controlled by the operating parameters of the laser-sintering technology, such as layer thickness, laser power, laser wavelength, and scanning speed [6, 10]. In this study, the total porosity percentage was 1-2% higher in the L-1 alloys than in the CC and L-2 alloys (Tables 6.2 and 6.3). This could be attributed to the fusion of the metal powder during the laser-sintering process that might increase the number of internal porosities between the sintered particles and between the different layers [6, 10, 11]. Furthermore, this could explain why the L-1 alloys presented lower density than alloys fabricated using the L-2 and CC techniques.

Although the porosity of L-1 alloys was slightly higher than that of CC alloys, the majority of pores in L-1 alloys were closed. Whereas, the percentage of open porosity was similar in all alloys (Table 6.3 and Figure 6.3). It is known that open porosities can become surface sites for

crack initiation, and therefore influence the fatigue resistance of the alloys [7, 22, 25]. Therefore, the fact that closed porosity influences the fatigue resistance less than open porosity might be the reason of the relatively low elastic modulus of L-1 alloys in this study [40]. On the other hand, our results showed that the porosity in the L-1 and L-2 alloys was more isotropic than in the CC group (Figure 6.3 and Table 6.3). This indicates that the porosities in L-1 and L-2 alloys are more oriented within the same volume than in the CC samples. Therefore, the homogeneity of both porosity and microstructure of L-1 and L-2 alloys could be another factor that explains the higher fatigue resistance of the laser-sintered alloys over the cast ones, despite having similar open porosities.

XRD analysis showed that both L-2 and CC alloys yielded similar crystallographic patterns (Figure 6.4 a); however, the XRD pattern L-1 alloys exhibited peaks referring to a hexagonal close-packed (hcp) phase of Co-Mo, which is in agreement with a previous study [24]. This could be a result of the phase transformation from (fcc) to (hcp) phase during the rapid cooling of the laser-sintering process since the (fcc) phase forms at high transformation temperatures as opposed to the (hcp) phase that forms at lower temperatures [6, 42]. Indeed, unlike the DMLS method (EOS system), the metal powder of L-1 alloys processed by SLS method (Phenix system) is exposed to temperatures below its phases transition [12]. Previous studies reported that the observed (hcp) phase influences the mechanical properties of the alloys and improves their strength, wear resistance and hardness, which further confirms our results [6, 44].

XRD analysis also revealed that the Co-Cr powder and L-1 alloys had similar crystal size, while the crystal size of the CC and L-2 alloys was larger than that of the L-1 alloys, which is most probably due to the solidification of the melted metal [20]. Optical photograph and SEM back-scattering images of the polished etched surfaces of L-1 and L-2 alloys demonstrated a fine microstructure, whereas the CC alloys showed different grain boundaries within the surface

(Figure 6.4 b-g). The smaller size of crystal and grain as well as the homogeneity in the microstructure that of the L-1 and L-2 alloys have a positive impact on the mechanical and fatigue properties of the alloys [6, 45]. In summary, this study suggests that both L-1 and L-2 alloys are more suitable to be used in the fabrication of the removable partial denture than the CC alloys because of their fatigue and physical properties.

6.5.3. Biocompatibility

Biocompatibility assays revealed that both L-1 and CC alloys had similar behaviors (Figure 6.6, Table 6.4). Both alloys released cobalt (Co), chromium (Cr) and molybdenum (Mo) to the simulated saliva media (PBS). Compared to the other elements, the release of Co was found to be relatively much higher from both alloys, and this is probably because Co is the major element (64%) in the composition of the Co-Cr alloys. Even though the L-1 alloys released higher amount of Co than the CC alloys, the amount of Co released from both alloys was safe and far below the recommended daily dietary intake (i.e. Co≤50 μg/day [26]).

The viability of human gingival epithelial cells was comparable in all groups on day 1; however, the proliferation rate of cells exposed to the L-1 and CC alloys declined over time in comparison to cells not exposed to the Co-Cr alloys (Figure 6.6). This can be attributed to the fact that the released Co inhibits cell growth [46, 47]. However, the cytotoxicity assays revealed that cells exposed to the L-1 and CC Co-Cr alloys behaved similarly to cells not exposed to Co-Cr alloys. Therefore, these results suggest that laser-sintered Co-Cr alloys are biocompatible and present similar biocompatibility properties when compared to the traditional cast Co-Cr alloys that are currently commonly used in the oral cavity.

6.6. CONCLUSION

Co-Cr alloys processed by the laser-sintering techniques are more precise and present better fatigue resistance and mechanical properties for removable partial dentures than cast alloys due to their better homogeneity and small grain size. Moreover, both laser-sintered and cast Co-Cr alloys present similar biocompatibility properties. Accordingly, laser-sintered RPDs could present clinical benefits over cast ones in terms of fitting and mechanical stability.

6.7. ACKNOWLEDGEMENTS

The authors would like to acknowledge King Saud University (Riyadh, Saudi Arabia); 3DRPD Inc.; and Natural Sciences and Engineering Research Council (NSERC) Collaborative Research Development, and Canada-Discovery grant for their financial support. We also thank Dr. Ahmed Al Subaie and Ms. Linda Harrison for their technical support.

6.8. REFERENCES

1. Preshaw P, Walls A, Jakubovics N, Moynihan P, Jepson N, Loewy Z. Association of removable partial denture use with oral and systemic health. J Dent 2011;39:711-9.

2. Carr AB, Brown DT. McCracken's removable partial prosthodontics. Saint Louis, Mo.: Elsevier Mosby; 2011.

3. Hummel SK, Wilson MA, Marker VA, Nunn ME. Quality of removable partial dentures worn by the adult U.S. population. *J Prosthet Dent* 2002;88:37-43.

4. Sun S-H, Koizumi Y, Kurosu S, Li Y-P, Matsumoto H, Chiba A. Build direction dependence of microstructure and high-temperature tensile property of Co–Cr–Mo alloy fabricated by electron beam melting. Acta Biomater 2014;64:154-68.

5. Venkatesh KV, Nandini VV. Direct Metal Laser Sintering: A Digitised Metal Casting Technology. J Indian Prosthodont Soc 2013;13:389-92.

6. Koutsoukis T, Zinelis S, Eliades G, Al-Wazzan K, Rifaiy MA, Al Jabbari YS. Selective Laser Melting Technique of Co-Cr Dental Alloys: A Review of Structure and Properties and Comparative Analysis with Other Available Techniques. *J Prosthodont* 2015.

7. Vallittu PK, Miettinen T. Duration of induction melting of cobalt-chromium alloy and its effect on resistance to deflection fatigue of cast denture clasps. *J Prosthet Dent* 1996;75:332-6.

8. Puskar T, Jevremovic D, Williams RJ, Eggbeer D, Vukelic D, Budak I. A Comparative Analysis of the Corrosive Effect of Artificial Saliva of Variable pH on DMLS and Cast Co-Cr-Mo Dental Alloy. *Materials* 2014;7:6486-501

9. Lima JMC, Anami LC, Araujo RM, Pavanelli CA. Removable partial dentures: use of rapid prototyping. *J Prosthodont* 2014;23:588-91.

10. Van Noort R. The future of dental devices is digital. *Dent Mater* 2012;28:3-12.

11. Akova T, Ucar Y, Tukay A, Balkaya MC, Brantley WA. Comparison of the bond strength of laser-sintered and cast base metal dental alloys to porcelain. *Dent Mater* 2008;24:1400-4.

12. Kruth, J.-P., et al., Binding mechanisms in selective laser sintering and selective laser melting. Rapid prototyping J, 2005. **11**(1): p. 26-36.

13. Ahn, D.-G., Direct metal additive manufacturing processes and their sustainable applications for green technology: A review. Int J Precis Eng Man, 2016. **3**(4): p. 381-395.

14. Mazzoli, A., Selective laser sintering in biomedical engineering. Med Biol Eng Comput, 2013. **51**(3): p. 245-256.

15. Vayre, B., F. Vignat, and F. Villeneuve, Metallic additive manufacturing: state-of-the-art review and prospects. Mech Ind, 2012. **13**(2): p. 89-96.

16. Ahn, D.-G., Applications of laser assisted metal rapid tooling process to manufacture of molding & forming tools—state of the art. Int J Precis Eng Man, 2011. **12**(5): p. 925.

17. Cheah, C.M., et al., Rapid prototyping and tooling techniques: a review of applications for rapid investment casting. Int J Adv Manuf Tech, 2005. **25**(3): p. 308-320.

18. Khaing, M.W., J.Y.H. Fuh, and L. Lu, Direct metal laser sintering for rapid tooling: processing and characterisation of EOS parts. J Mater Process Technol, 2001. **113**(1–3): p. 269-272.

19. Hollander DA, von Walter M, Wirtz T, Sellei R, Schmidt-Rohlfing B, Paar O, Eril HJ. Structural, mechanical and in vitro characterization of individually structured Ti–6Al–4V produced by direct laser forming. *Biomaterials* 2006;27:955-63.

20. Vandenbroucke B, Kruth J-P. Selective laser melting of biocompatible metals for rapid manufacturing of medical parts. Rapid Prototyping J 2007;13:196-203.

21. Zhang B, Huang Q, Gao Y, Luo P, Zhao C. Preliminary study on some properties of Co-Cr dental alloy formed by selective laser melting technique. J Wuhan Univ Technol 2012;27:665-8.

22. Gapido CG, Kobayashi H, Miyakawa O, Kohno S. Fatigue resistance of cast occlusal rests using Co-Cr and Ag-Pd-Cu-Au alloys. *J Prosthet Dent* 2003;90:261-9.

23. Sandu L, Faur N, Bortun C. Finite element stress analysis and fatigue behavior of cast circumferential clasps. *J Prosthet Dent* 2007;97:39-44.

24. Al Jabbari YS, Koutsoukis T, Barmpagadaki X, Zinelis S. Metallurgical and interfacial characterization of PFM Co–Cr dental alloys fabricated via casting, milling or selective laser melting. *Dent Mater* 2014;30:e79-e88.

25. Xin XZ, Xiang N, Chen J, Wei B. In vitro biocompatibility of Co–Cr alloy fabricated by selective laser melting or traditional casting techniques. Mater Lett 2012;88:101-3.

26. Wataha JC. Biocompatibility of dental casting alloys: A review. *J Prosthet Dent* 2000;83:223-34.

27. Hedberg, Y.S., et al., In vitro biocompatibility of CoCrMo dental alloys fabricated by selective laser melting. Dent Mater, 2014. **30**(5): p. 525-534.

28. Bae, E.-J., et al., A comparative study of additive and subtractive manufacturing for dental restorations. J Prosthet Dent.

29. Alageel O, Abdallah M-N, Luo ZY, Del-Rio-Highsmith J, Cerruti M, Tamimi F. Bonding metals to poly(methyl methacrylate) using aryldiazonium salts. *Dent Mater* 2015;31:105-14.

30. Callister WD. Fundamentals of materials science and engineering: an integrated approach. Hoboken, NJ: John Wiley & Sons; 2005.

31. Chermant J, Osterstock F. Fracture toughness and fracture of WC-Co composites. J Mater Sci 1976;11:1939-51.

32. Brynk, T., et al., The method of fracture toughness measurement of brittle materials by means of high speed camera and DIC. Comput Mater Sci 2012. 64: p. 221-224.

33. Ghadimi E, Eimar H, Song J, Marelli B, Ciobanu O, Abdallah M-N, et al. Regulated fracture in tooth enamel: A nanotechnological strategy from nature. J Biomech 2014;47:2444-51.

34. Abdallah M-N, Eimar H, Bassett DC, Schnabel M, Ciobanu O, Nelea V, et al. Diagenesis-inspired reaction of magnesium ions with surface enamel mineral modifies properties of human teeth. Acta Biomater 2016;37:174-83.

35. Viennot S, Dalard F, Lissac M, Grosgogeat B. Corrosion resistance of cobalt-chromium and palladium-silver alloys used in fixed prosthetic restorations. Eur J Oral Sci 2005;113:90-5.

36. Rees JS, Jacobsen PH. Elastic modulus of the periodontal ligament. *Biomater* 1997;18:995-9.

37. Combe EC, Shaglouf AMS, Watts DC, Wilson NHF. Mechanical properties of direct core build-up materials. *Dent Mater* 1999;15:158-65.

38. Currey JD. How Well Are Bones Designed to Resist Fracture? J Bone Miner Res 2003;18:591-8.

39. Mahmoud A, Wakabayashi N, Takahashi H, Ohyama T. Deflection fatigue of Ti-6Al-7Nb, Co-Cr, and gold alloy cast clasps. *J Prosthet Dent* 2005;93:183-8.

40.Greiner C, Oppenheimer SM, Dunand DC. High strength, low stiffness, porous NiTi with superelastic properties. Acta Biomater 2005;1:705-16.

41. Lassila LVJ, Vallittu PK. Effect of water and artificial saliva on the low cycle fatigue resistance of cobalt-chromium dental alloy. *J Prosthet Dent* 1998;80:708-13.

42. Barucca G, Santecchia E, Majni G, Girardin E, Bassoli E, Denti L, et al. Structural characterization of biomedical Co–Cr–Mo components produced by direct metal laser sintering. Mater Sci Eng 2015;48:263-9.

43. AlMangour, B., D. Grzesiak, and J.-M. Yang, Selective laser melting of TiB2/H13 steel nanocomposites: Influence of hot isostatic pressing post-treatment. J Mater Process Technol, 2017. 244: p. 344-353.

44. Moharrami N, Langton DJ, Sayginer O, Bull SJ. Why does titanium alloy wear cobalt chrome alloy despite lower bulk hardness: A nanoindentation study? Thin Solid Films 2013;549:79-86

45. Wu L, Zhu H, Gai X, Wang Y. Evaluation of the mechanical properties and porcelain bond strength of cobalt-chromium dental alloy fabricated by selective laser melting. *J Prosthet Dent* 2014;111:51-5.

46. Berstein A, Bemauer I, Marx R, Geurtsen W. Human cell culture studies with dental metallic materials. *Biomaterials* 1992;13:98-100.

47. Bearden L, Cooke F. Growth inhibition of cultured fibroblasts by cobalt and nickel. *J Biomed Mater Res* 1980;14:289-309.

Chapter 7: Manuscript 2

The previous chapter showed that the laser sintering (laser melting) technology improved the mechanical, physical properties of the removable partial denture (RPD) alloys; thus, it would improve the quality of the RPD treatment. However, the poor RPD designs is another reason for failures and complications of the RPD treatment which result in patient dissatisfaction with the treatment. Therefore, developing a proper RPD designing guideline could improve RPD treatment. This chapter is a published manuscript aimed to develop a designing guideline for determining the optimal retention for the RPD. This manuscript including the references formatting and references list is identical to the published version.

Determining the Retention of Removable Partial Dentures

Author names and affiliations:

Omar Alageel BSc, MSc, [a] Ammar A. Alsheghri BSc, MSc, [b] Suliman Algezani BSc, MSc, [c] Eric Caron, DDS, MSc, [d] and Faleh Tamimi BDS, MSc, PhD [e]

[a] Doctoral student, Faculty of Dentistry, McGill University, Montreal, QC, Canada; and Lecturer, College of Applied Medical Sciences, King Saud University, Riyadh, Saudi Arabia.

[b] Doctoral student, Department of Mining and Materials Engineering, McGill University, Montreal, Canada.

[c] Graduate student, Faculty of Dentistry, McGill University, Montreal, QC, Canada

[d] Researcher, Dental Wings, Montreal, Canada.

[e] Associate Professor and Canada Research Chair, Faculty of Dentistry, McGill University, Montreal, Canada.

This manuscript has been accepted to be published at Journal of Prosthetic Dentistry on July 2018.

7.1. ABSTRACT

Statement of problem. Removable partial dentures (RPDs) provide a cost-effective treatment for millions of partially edentulous patients worldwide. However, they often fail because of loss of retention. One reason for this problem is lack of precise guidelines for designing retentive RPDs.

Purpose. The purpose of this study was to determine the forces produced by food and clasps during mastication to develop an algorithm for predicting RPD retention and to help determine the optimal number of clasps.

Material and methods. The forces that food exerts on acrylic resin teeth during simulated mastication and the retention forces provided by clasps (wrought wire, circumferential, and I-bar) engaging teeth were measured using a universal testing machine. Statistical analysis was performed with a 1-way analysis of variance and repeated-measures ANOVA while the developed algorithm was evaluated by using sensitivity and specificity analysis.

Results. The force exerted by food mastication on each individual tooth ranged between 1.7 and 12.2 N, depending on the type of tooth, tooth anatomy, occlusion, and food. The retention force of the clasps after cyclic testing ranged between 2.9 and 14.5 N, depending on the type of tooth abutment and clasp. Using these measurements, an algorithm was developed to predict RPD retention. The algorithm was confirmed experimentally on 36 RPDs, showing a sensitivity of 96%, a specificity of 100%, and an accuracy of 97%.

Conclusions. The forces generated by food mastication on teeth varied according to the type of tooth, occlusion, and food. The retention force of RPD clasps varied according to the type of tooth and clasp. An algorithm for predicting RPD retention and determining the optimal number of clasps was developed and validated experimentally.

CLINICAL IMPLICATIONS

The guidelines developed in this study may help predict RPD retention and determine the optimal number of clasps in an RPD.

7.2. INTRODUCTION

Removable partial dentures (RPDs) are cost-effective and functional dental prostheses that are used to restore missing teeth in partially edentulous patients.[1,2]An RPD is a treatment option that can improve the quality of life for millions of patients worldwide; over 13% of the adult population in North America and Europe wear RPDs.[3,4] However, many complications are associated with RPDs, mainly related to inadequate quality and poor design.[1,5,6] Indeed, poor RPD design results in insufficient retention, which is the main reason for treatment failure and patient dissatisfaction.[5-7]

Designing RPDs is challenging because there are 65534 possible forms of partial edentulism and the available design guidelines lack scientific evidence and do not cover all edentulism forms.[2,8,9] Therefore, RPDs are designed subjectively based on the experience of dental professionals, which could often result in inadequate designs.[10] In fact, many dentists delegate design work to dental technicians due to their extensive design experience.[11] Knowledge-based systems are available for designing RPDs that provide the most appropriate RPD design based on a database of previous patients.[10,12] However, RPD designs in the database might be inadequate and inappropriate since they were designed subjectively based on operator experience.

A properly designed RPD should provide sufficient retention to resist the dislodging forces caused during food mastication and functional muscle movements; this can be achieved by retentive elements engaging the abutment teeth, including clasps, proximal plates, and rests or by attachment on dental implants.[2,5,9,13] However, most commonly, retention in RPDs is provided by clasp designed in a variety of forms (such as I-bar, and circumferential clasp) and materials (such as wrought wire, cast metals, or acrylic resin).[2,9] Frank et al suggested that the retention of a clasp in an RPD should be between 3 and 7.5 N.[14,15] However, this can vary according to the clasp form,

location, undercut depth, composition, and guide planes.[13,14,16] Accordingly, retention can be improved by optimizing the shape, undercut depth, and fabrication process.[16-19] For instance, clasps made with laser-sintering technology present better fatigue resistance and higher precision than cast clasps.[20,21]

RPD dislodgment occurs because of the force that pulls food away from the teeth as a result of the action of adherent foods.[13,22] This force depends on factors such as patient masticatory habits, occlusion, tooth anatomy, and food characteristics such as size, shape, and texture.[13,22,23]

A common question raised in designing an RPD is determining an adequate number of clasps to provide sufficient retention to resist the dislodging forces caused during food mastication. RPDs with too few clasps could result in insufficient retention while RPD with too many clasps could cause harm to the patient. Currently, guidelines to determine the optimal number of RPD clasps are lacking, as is the optimal amount of retention needed to achieve a retentive RPD. Therefore, determining the optimal retention of any RPD design, and whether it is sufficient or not, is the key to developing better design guidelines. Accordingly, the hypothesis of this study was that for an RPD to be retentive during mastication, the retention forces provided by its clasps should be higher than the dislodging forces generated by food. The purpose of this study was to determine the forces produced by food and clasps during mastication to develop an algorithm for predicting RPD retention and help determine the optimal number of clasps. Subsequently, this study aimed to validate the new algorithm for predicting RPD retention experimentally.

7.3. MATERIAL AND METHODS

The force that food exerts on acrylic resin teeth was measured by simulated mastication using a dentoform model (Nissin Dental Products Inc) fixed on a universal testing machine (Instron

Corp) set at a constant mastication speed of 5mm/second (Figure 7.1, Supplemental Figure 7.1).[24] The dentoform model allows placing or removing each tooth on the model separately, which helped in assessing all tooth types in both arches. The force exerted by masticating caramel candy on anatomic teeth occluding in class 1 occlusion was conducted for every tooth separately on both arches with 15 repetitions per tooth. Furthermore, other types of food, tooth anatomy, and occlusion were tested for all teeth in both arches, and the tests were repeated 15 times for each type of food, tooth anatomy, and occlusion.

The types of tested food included caramel candy (Werther's original), chewing gum (Wrigley's Excel), and toast bread (Villaggio) and were chosen based on a previous study that evaluated the stickiness of 21 different food items.[25] The impact of tooth anatomy was also assessed using anatomic and nonanatomic acrylic resin teeth.

To assess the impact of occlusion upon simulated mastication, the dental arches were positioned and adjusted to be at class 1, 2, and 3 occlusions. Class 1 occurs when the maxillary teeth slightly overlap the mandibular teeth, class 2 when the maxillary teeth severely overlap the mandibular teeth, and class 3 when the mandibular teeth overlap the maxillary teeth.[26]

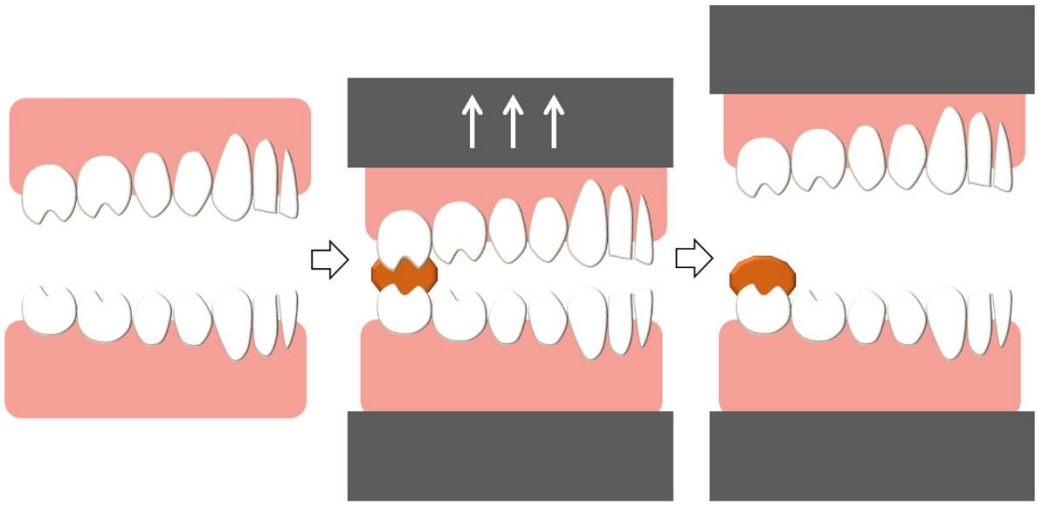


Figure 7.1. Testing of masticatory tensile forces force.

The retention forces of wrought wire, circumferential, and I-bar clasps engaging undercuts in each tooth type in both arches were measured. Because of their flexibility, wrought wire clasps usually engage deeper undercuts (0.50 mm) than Co-Cr circumferential and I-bar clasps (0.25 mm).[18,27] For wrought wire clasps, 3 test specimens per tooth type were fabricated on partially edentulous casts duplicated from a dentoform model with a silicone impression material (Exaktosil N21; Bredent GmbH) and dental stone (Figure 7.2). Each test specimen contained a pair of wrought wire clasps (17 GA; Keystone Dental Inc) placed at an undercut depth of 0.5 mm, an acrylic resin denture-base (Biocryl Resin Acrylic; Great Lakes Ortho Inc), and an attachment to the testing machine. For the circumferential and I-bar clasps, Co-Cr clasps were designed at undercut depths of 0.1 mm on the duplicated scanned model of the dentoform using a 3D scanner and a CAD software (3Series; Dental Wings Inc) and processed by direct laser-sintering technology (Phenix PXM) at the prototyping center (3DRPD Inc).[20]

To test retention forces, the specimens were attached to the upper grip of a universal testing machine (Instron Corp) and placed on the dentoform model that was fixed on the lower grip of the machine (Figure 7.2 and Supplemental Figure 7.2, 7.3). The machine applied a pull-out force at a constant speed of 5 mm/second until the clasps disengaged from the abutment teeth. The retention force was recorded, and the process was repeated 5 times for each test. Cyclic testing was applied manually by inserting and removing the clasps from the abutment teeth for up to 1200 cycles, which is the equivalent to wearing dentures for 1 year.[28-30] The retention force after 1200 cycles was then recorded as described earlier.

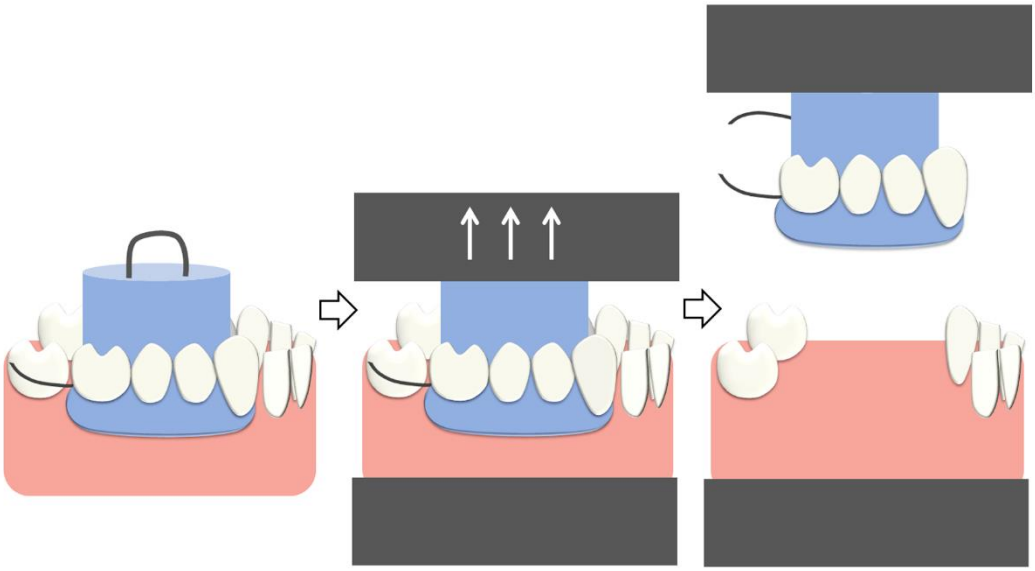


Figure 7.2. Testing of retention force for removable partial denture clasps.

An algorithm for predicting RPD retention was developed based on the hypothesis of this study by using the measurements from food mastication and clasp retention. $\text{RPD retention force} = \sum \text{clasp retention force} - \sum \text{dislodging force on replaced tooth}$ (Equation 1). This equation calculates the net retention force of any RPD design and therefore can predict its retention performance. Based on that, a net retention force greater than zero indicates sufficient retention, whereas a negative value indicates insufficient retention.

A validation test was performed to test the accuracy of the algorithm for predicting RPD retention. A total of 36 random RPDs were tested (18 per arch). The mechanism of the selection and testing is summarized in Figure 7.3. First, since there are 16384 possible forms of partial edentulism per arch, 36 random numbers were selected from 1 to 16384 by using a random number generator (www.random.org). The selected numbers were then converted to a binary number representing an edentulous arch where 1 represented missing teeth and 0 represented present teeth. The randomly edentulous partial arches with fewer than 2 remaining or missing teeth were

excluded from the study because these edentulous arches should not be treated with an RPD. The selected random numbers and randomly generated edentulous arches are shown in Table 7.4.

Next, acrylic RPDs were subjectively designed and fabricated for the selected edentulous arches. This was done on a master cast duplicated from the dentoform model as described previously. Finally, the experimental retention performances of the 36 RPDs were blind tested in simulated mastication with caramel candy as previously described. The RPDs that retained the original position during mastication were considered to have sufficient retention whereas those displaced from their position were considered to have insufficient retention. The experimental results were then compared with those generated by the algorithm for predicting RPD retention.

Statistical analysis to identify differences between teeth for the forces produced by food and clasps was done using a 1-way analysis of variance (ANOVA) followed by the post hoc Tukey honestly significant difference (HSD) test. Repeated-measures ANOVA was used to analyze the differences among the forces exerted by caramel candy mastication for class 1occlusion under different mastication conditions and between clasp retention forces after1 and 1200 cycles. Statistical software (IBM SPSS Statistics v23.0; IBM Corp) was used for the analysis (α=.05). The sample size for the validation of the algorithm for predicting RPD retention was calculated at a confidence interval of 95%, design prevalence of 10%, unit specificity of 100%, and unit and required population sensitivity of 95%. Statistical analysis for the validation of the algorithm was performed with sensitivity and specificity analysis.[31]

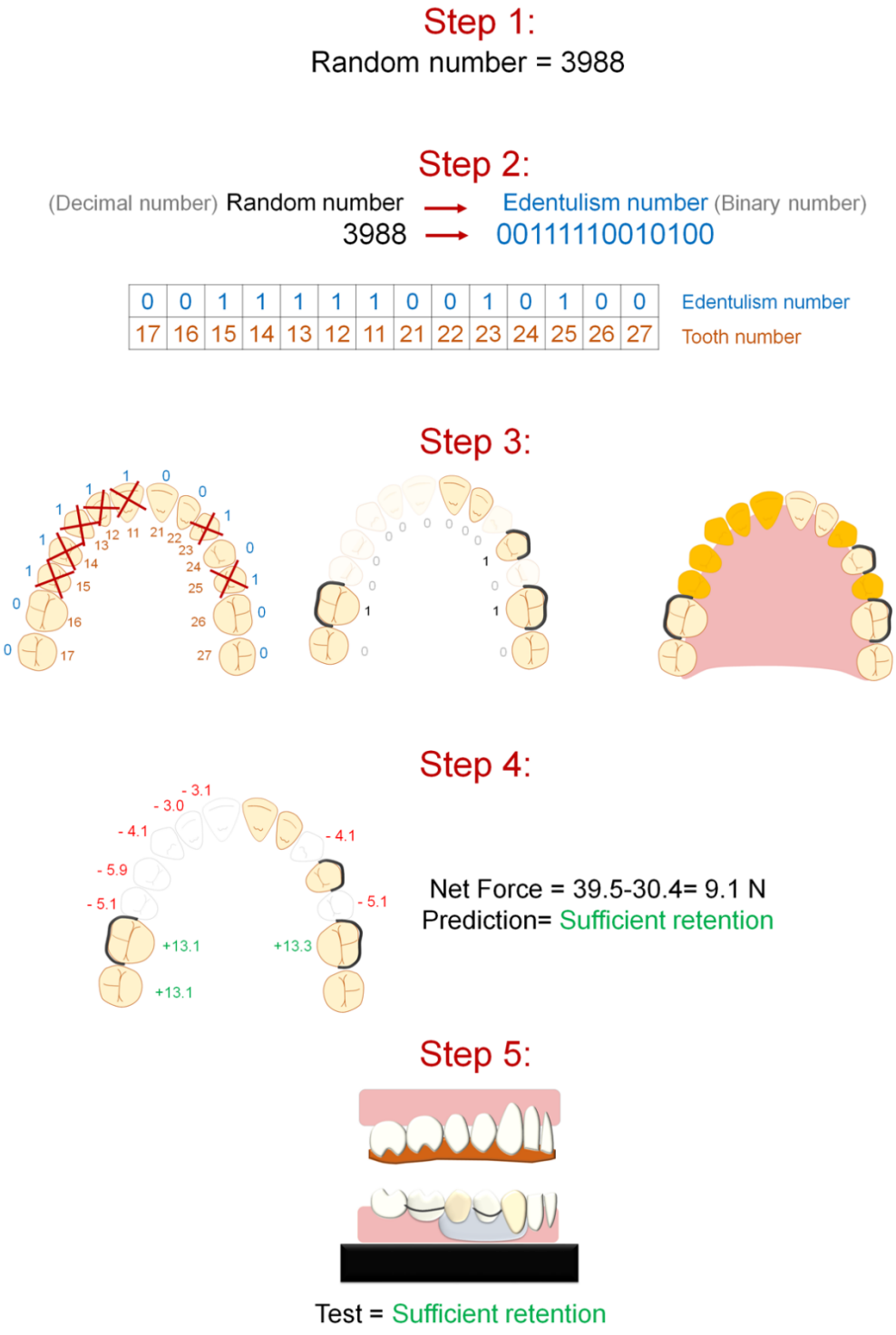


Figure 7.3. Example of selection and testing of random edentulous arches for algorithm validation. Step 1: selection of random number between 1 and 16384. Step 2: converting random number into edentulous arch (binary number) in which 1 represents missing tooth and 0 represents tooth. Step 3: determining clasp location and fabricating RPD. Step 4: determining retention prediction by using algorithm. Step 5: testing retention performance of removable partial denture.

7.4. RESULTS

The forces that food exerted on teeth varied depending on the type of tooth, occlusion, and food. The forces of caramel candy mastication on each anatomic tooth type in class 1 occlusion are shown in Table 7.1. The highest force generated by caramel candy mastication was recorded for the first molars in both the maxillary (12.0 ±0.7 N) and mandibular arches (12.2 ±1.1 N), while the lowest force was recorded on the mandibular lateral incisors (1.7 ±0.6 N). The forces exerted by caramel candy mastication were significantly different among tooth type (P<.001); molars and premolars (12.2 to 4.7 N) showed higher forces (P<.001) than canines and incisors (4.1 to 1.7 N).

Table 7.1. Masticatory tensile forces generated by mastication of caramel candy on different types of anatomic tooth at occlusion class 1 in both arches.

Arch	**Tooth**	**Masticatory tensile forces (N)**	***P* Values for post hoc comparison among tooth type**						
			Central	Lateral	Canine	1st Premolar	2nd Premolar	1st Molar	2nd Molar
Maxillary	Central	3.1 ±0.6	-	.999	.002	<.001	<.001	<.001	<.001
“	Lateral	3.0 ±0.4	.999	-	<.001	<.001	<.001	<.001	<.001
“	Canine	4.1 ±0.6	.002	<.001	-	<.001	.004	<.001	<.001
“	1st Premolar	5.9 ±0.9	<.001	<.001	<.001	-	.023	<.001	<.001
“	2nd Premolar	5.1 ±0.8	<.001	<.001	.004	.023	-	<.001	<.001
“	1st Molar	12.0 ±0.7	<.001	<.001	<.001	<.001	<.001	-	<.001
“	2nd Molar	10.1 ±0.9	<.001	<.001	<.001	<.001	<.001	<.001	-
Mandibular	Central	3.1 ±0.9	-	<.001	.709	<.001	<.001	<.001	<.001
“	Lateral	1.7 ±0.6	<.001	-	<.001	<.001	<.001	<.001	<.001
“	Canine	3.6 ±0.6	.709	<.001	-	<011	<.001	<.001	<.001
“	1st Premolar	4.7 ±0.9	<.001	<.001	<011	-	.683	<.001	<.001
“	2nd Premolar	5.3 ±0.7	<.001	<.001	<.001	.683	-	<.001	<.001
“	1st Molar	12.2 ±1.1	<.001	<.001	<.001	<.001	<.001	-	.975
“	2nd Molar	11.9 ±1.2	<.001	<.001	<.001	<.001	<.001	.957	-

The force exerted by caramel candy upon mastication in class 1 occlusion in the entire maxillary and mandibular arch (51.6 ±5.8 N in maxillary arch; 49.6 ±2.6 N in mandibular arch) was higher (P<.001) when the teeth had anatomic occlusal surfaces than when they had nonanatomic occlusal surfaces (45.4 ±1.7 N in maxillary arch; 45.2 ±2.6 N in mandibular arch)

(Table 7.2). Mastication with class 1 occlusion produced higher force than class 3 occlusion in the maxillary arch (46.8 ±4.0 N; *P*=.003) and mandibular arch (41.0 ±3.4 N; *P*<.001) than class 2 occlusion in the mandibular arch (40.0 ±2.8 N; *P*<.001). Also, the mastication of caramel candy produced higher force (*P*<.001) than the chewing gum (16.5 ±1.3 N in the maxillary arch; 15.6 ±1.2N in the mandibular arch) and bread (6.2 ±1.0 N in the maxillary arch; 5.9 ±0.9 N in the mandibular arch).

Table 7.2. Masticatory tensile forces exerted by different types of food on arch depending on tooth anatomy and type of occlusion.

Arch	**Tooth**	**Tooth anatomy**	**Occlusion**	**Food type**	**Masticatory tensile forces (N)**	***P***
Maxillary	All teeth	Anatomic	Class 1	Caramel	51.6 ±5.8	reference
“	“	“	Class 2	“	49.6 ±2.4	<.001
“	“	“	Class 3	“	46.8 ±4.0	.003
“	“	Nonanatomic	Class 1	“	45.4 ±1.7	<.001
“	“	Anatomic	“	Gum	16.5 ±1.3	“
“	“	“	“	Bread	6.2 ±1.0	“
Mandibular	All teeth	Anatomic	Class 1	Caramel	49.6 ±2.6	reference
“	“	“	Class 2	“	40.0 ±2.8	<.001
“	“	“	Class 3	“	41.0 ±3.2	“
“	“	Nonanatomic	Class 1	“	45.2 ±2.6	“
“	“	Anatomic	“	Gum	15.6 ±1.2	“
“	“	“	“	Bread	5.9 ±0.9	“

The retention forces of wrought wire, circumferential, and I-bar clasps engaging teeth are shown in Table 7.3. The highest retention force with wrought wire and circumferential clasps was recorded on molars (14.5 ±1.7 N and 6.8 ±1.0 N), while the lowest retention force was recorded on incisors (8.5 ±1.6 N and 2.9 ±1.2 N). The retention forces of wrought wire and circumferential clasps were significantly (*P*<.001) different depending on the type of tooth (Table 7.3 and Supplemental Tables 7.1, 7.2). I-bar clasps provided similar (*P*=.33 for maxillary arch and *P*=.15

for mandibular arch) retention force values on all teeth (3.6 ±0.9 to 4.8 ±1.3 N) (Supplemental Table 7.3). Fatigue cycling significantly decreased the retention forces of wrought wire clasps on all teeth except incisors and of circumferential clasps on mandibular premolars and molars but did not affect the retention of I-bar clasps (Table 7.3).

Table 7.3. Retention forces of wrought wire and circumferential and I-bar clasps engaging on abutment teeth before and after fatigue

Arch	**Tooth**	**Retention forces (N) of Wrought wire clasps**		**Circumferential clasps**		**I-bar clasps**	
		1 Cycle	**1200 Cycles**	**1 Cycle**	**1200 Cycles**	**1 Cycle**	**1200 Cycles**
Maxillary	Central	9.3 ±1.1	9.1 ±1.1	3.4 ±0.8	2.9 ±1.2	4.2 ±1.7	3.9 ±1.2
"	Lateral	8.7 ±1.9	8.5 ±1.6	4.1 ±0.7	3.4 ±1.5	4.0 ±1.2	3.6 ±0.9
"	Canine	15.8 ±2.2	13.3 ±2.1*f	4.5 ±1.0	4.1 ±1.6	4.3 ±1.3	4.2 ±0.8
"	1st Premolar	13.7 ±1.9	11.3 ±1.8 *a	4.6 ±1.1	4.4 ±1.5	5.3 ±1.4	4.8 ±0.7
"	2nd Premolar	13.8 ±3.0	10.7 ±2.1 *e	5.3 ±1.2	4.5 ±1.1	5.2 ±1.3	4.8 ±1.2
"	1st Molar	14.4 ±2.4	13.1 ±0.7 *d	6.9 ±0.7	6.8 ±1.0	5.0 ±0.5	4.8 ±1.3
"	2nd Molar	14.2 ±2.7	13.1 ±1.5 *h	6.8 ±1.0	6.1 ±0.9	5.0 ±0.4	4.7 ±0.9
Mandibular	Central	11.4 ±1.7	11.0 ±1.2	3.7 ±0.8	3.0 ±1.4	4.9 ±1.8	3.6 ±0.7
"	Lateral	9.8 ±2.7	9.5 ±1.6	3.8 ±1.0	3.4 ±1.5	4.9 ±1.9	3.6 ±0.7
"	Canine	15.9 ±2.4	14.0 ±1.7 *a	5.2 ±2.2	4.3 ±1.0	4.8 ±2.2	4.3 ±0.9
"	1st Premolar	13.3 ±1.2	12.2 ±1.2 *c	6.3 ±0.8	5.0 ±0.8*e	5.1 ±1.4	4.8 ±0.5
"	2nd Premolar	13.3 ±1.7	11.9 ±1.4 *a	6.5 ±0.9	5.1 ±0.6*h	4.7 ±1.3	4.5 ±0.3
"	1st Molar	18.3 ±3.0	14.5 ±1.7 *a	7.2 ±1.0	6.2 ±0.9*h	4.6 ±0.5	4.5 ±0.6
"	2nd Molar	17.3 ±3.2	13.6 ±2.6 *b	7.2 ±0.7	6.2 ±1.0*g	5.1 ±0.9	4.6 ±1.4

** Indicates significant difference between the retention forces of 1 cycle and 1200 cycles P: a<.001, b=.002, c=.003, d=.008, e=.01, f =.02, g=.03, h=.04.*

Based on the data collected, an algorithm for predicting RPD retention was generated:

$$\text{RPD retention force} = \sum \text{clasp retention force} - (\text{Ka Kb Kc} \sum \text{disloding force on replaced tooth}),$$ (Equation 2).

where Ka is a constant for tooth surface anatomy, Kb is a constant for occlusion type, and Kc is a constant for food type (Supplemental Table 7.4).

A working version of the algorithm has been made available online at www.ebhnow.com/apps/0160. The deviation of this equation was ±8.1 N. The algorithm was validated experimentally on 36 randomly selected RPDs (Table 7.4). A total of 24 RPDs were

predicted by the algorithm to provide sufficient retention and presented sufficient retention during the experimental retention test. In addition, 11 of 12 RPDs were predicted to provide insufficient retention and presented insufficient retention experimentally. Only 1of 36 RPDs tested did not follow the prediction. Accordingly, the algorithm had a sensitivity of 96%, a specificity of 100%, and an accuracy of 97%.

Table 7.4. Experimental retention performances of random RPDs in comparison with retention performances predicted by algorithm for predicting RPD retention.

	Arch	Random number	Edentulism number	Clasps position	Algorithm prediction		Experimental test
					Net force (N)	Sufficient Retention	Sufficient Retention
1	Mandibular	11389	10111110001101	01000001010010	0.7	Yes	Yes
2	“	11389	10111110001101	01000001000010	-13.0	No	No
3	“	11389	10111110001101	01000001000000	-27	“	“
4	“	109	10110110000000	01001000000001	15.0	Yes	Yes
5	“	109	10110110000000	01000000000001	1.5	“	“
6	“	109	10110110000000	01000000000000	-10.0	No	No
7	“	11283	11001000001101	00110100010010	13.0	Yes	Yes
8	“	11283	11001000001101	00110000010010	3.5	“	“
9	“	11283	11001000001101	00110000000010	-10.0	No	Yes *
10	“	358	01100110100000	10010000010001	29.5	Yes	Yes
11	“	358	01100110100000	10010000000001	15.0	“	“
12	“	358	01100110100000	10000000000001	3.3	“	“
13	“	6669	10110000010110	01001000001001	21.3	“	“
14	“	6669	10110000010110	01001000000001	8.0	“	“
15	“	6669	10110000010110	00001000000001	-10.0	No	No
16	“	9096	00010001110001	10001000001010	21.3	Yes	Yes
17	“	9096	00010001110001	10000000001010	15.3	“	“
18	“	9096	00010001110001	10000000000010	-10.0	No	No
19	Maxillary	3988	00101001111100	01010100000010	15.0	Yes	Yes
20	“	3988	00101001111100	01010000000010	7.2	“	“
21	“	3988	00101001111100	00010000000010	-6.0	No	No
22	“	285	00000100011101	00001010100010	20.0	Yes	Yes
23	“	285	00000100011101	00001000100010	6.0	“	“
24	“	285	00000100011101	00001000000010	-1.6	No	No
25	“	5880	00011111011010	00100000100101	11.0	Yes	Yes
26	“	5880	00011111011010	00100000100001	-0.3	No	No
27	“	5880	00011111011010	00100000100001	-10.0	“	“
28	“	2770	01001011010100	10100000001010	16.8	Yes	Yes
29	“	2770	01001011010100	10000000001010	5.7	“	“
30	“	2770	01001011010100	10000000001000	-7.0	No	No
31	“	10846	01111010010101	10000101101010	24.0	Yes	Yes
32	“	10846	01111010010101	10000100101010	6.7	“	“
33	“	10846	01111010010101	10000100101000	1.9	“	“
34	“	12444	00111001000001	01000110100100	12.4	“	“
35	“	12444	00111001000001	01000100100100	6.5	“	“
36	“	12444	00111001000001	01000100000100	-4.2	No	No

True positive	24
True Negative	11
False positive	0
False Negative	1
Sensitivity	96%
Specificity	100%
Accuracy	97%

*Sufficient retention ability to resist dislodging forces caused during food mastication; * Indicates difference between predicted and tested RPD retention.*

7.5. DISCUSSION

The hypothesis of this study was confirmed. For an RPD to withstand mastication without being dislodged, the sum of the retention forces provided by each clasp should be higher than the sum of the dislodging forces generated by food mastication on each replaced missing tooth. By confirming the hypothesis, this study established a new approach for predicting and optimizing RPD retention using experimental data of forces produced by food and clasps during mastication (Equation 2). The authors are unaware of a previous engineering model that predicts the retention of any RPD. This could help dental professionals better determine the appropriate number and positions of clasps in RPDs and subsequent automatization of the designing process. Accordingly, the model developed in this study has the potential to enhance the quality of life for millions of patients worldwide by providing them with more predictable treatments.[7,25]

This study indicated that the forces exerted by food mastication depended on the tooth, occlusion, and food (Tables 7.1, 7.2). First, each tooth type and tooth anatomy generated a specific dislodging force. As reported previously,[13] the larger surface areas of posterior or anatomic teeth provided higher forces than the smaller surface areas in anterior or nonanatomic teeth. Moreover, the type of occlusion also affected the dislodging forces generated by food mastication, which might be related to the contact area between maxillary and mandibular teeth during mastication.[26] In addition, among the different food types tested, caramel exerted the highest forces followed by chewing gum, while bread provided the lowest forces. This was due to the variable in stickiness among the different food types as reported previously.[25] Therefore, the algorithm for predicting RPD retention (Equation 2) must take into account the unique characteristic of each tooth and has to be adjusted for constants related to tooth anatomy, occlusion type, and food type.

The retention forces provided by wrought wire and circumferential clasps varied according to tooth type and clasp length. Wrought wire and circumferential clasps on larger teeth such as molars presented higher retention forces than on smaller teeth such as incisors. The friction surface area between the wrought wire or circumferential clasp and the tooth might be the reason for the higher retention force on large teeth despite their having longer and more flexible clasps.[14,15] The retention forces of I-bar clasps did not vary substantially among teeth possibly because their retention surface area and friction are similar across the different tooth types. Thus, the algorithm must take into account the differences among types of tooth and clasp. In addition, wrought wire clasps in this study showed higher retention forces than circumferential and I-bars probably because they possessed different undercuts. The wrought wire clasps were placed at undercuts of 0.5 mm, while circumferential and I-bar clasps were placed at undercuts of 0.1 mm due to the path of insertion and removal of the testing specimens. Thus, the retention forces of circumferential and I-bar clasps engaged in deeper undercuts would be comparable to the retention of Co-Cr clasps reported in the literature.[2,14,32]

Clasps undergo repeated bending caused by mastication, insertion, and removal of the RPD and therefore are vulnerable to loss of retention. The retention of clasps usually changes after wearing the RPD for a time[28]; thus, cyclic fatigue testing of clasps was also assessed in this study. The retention forces of all types of clasps decreased after cyclic fatigue, which could be due to clasp deformation on the wear between the crown and the inner surface of the clasp.[14,29] This can decrease the friction coefficient between the clasp and the abutment tooth and lead to loss of retention.[32] The loss of retention was more pronounced on long wrought wire clasps than on short ones, which agrees with a previous study.[17] Surprisingly, even though wrought wire clasps are known to maintain much of their retention after cyclic testing because of their flexibility, both I-

bar and circumferential clasps in this study outperformed the wrought wire clasps.[30] This is probably because the I-bar and circumferential clasps were prepared by laser sintering technology and were engaging smaller undercuts.[20]

The algorithm for predicting RPD retention was validated experimentally in a blinded test to avoid bias. Only 1 of 36 RPDs tested did not follow the prediction; this RPD showed higher experimental retention than predicted, which might be related to the friction of the clasps. Accordingly, the sensitivity of the algorithm was 96%, and the specificity was 100%; this means that all the RPDs predicted to have sufficient retention by the algorithm presented sufficient retention during food mastication, while 96% of the RPDs predicted to have insufficient retention presented insufficient retention during food mastication. Generally, the algorithm for predicting RPD retention was confirmed with an accuracy of 97%.

Limitations in this study should be considered in future studies to improve the clinical performance of the algorithm. For example, parameters that may vary among patients were not tested. This includes tooth anatomy, height of tooth crown, mastication mechanics (such as mastication speed, angle, and food volume), and the path of insertion and removal of the RPDs.[13] Another limitation was that the clasp retention experiments were performed in a dry ambient condition with acrylic resin teeth; this might underestimate clasp retention force in the oral environment because of the adhesive effect of saliva in tooth-clasp interactions.[18,33] In addition, variations in the fabrication process of RPDs (such as clasps materials, thickness, length, and undercut depth) between dental clinics and laboratories might limit the validity of the algorithm in clinical practices.[14,18] Moreover, other clasp types and other retentive features of RPDs such as rotational partial dentures were not addressed in this algorithm; including these permutations would acknowledge deviations in the algorithm.[16,19,34] Nevertheless, with the arrival of computer-

aided design and computer-aided manufacturing (CAD-CAM) technology and the digitalization of the RPD fabrication process, these limitations can be overcome.

7.6. CONCLUSIONS

Within the limitations of this in vitro study, the following conclusions were drawn:

1. The force generated by food mastication on teeth varied according to the type of tooth, occlusion, and food.

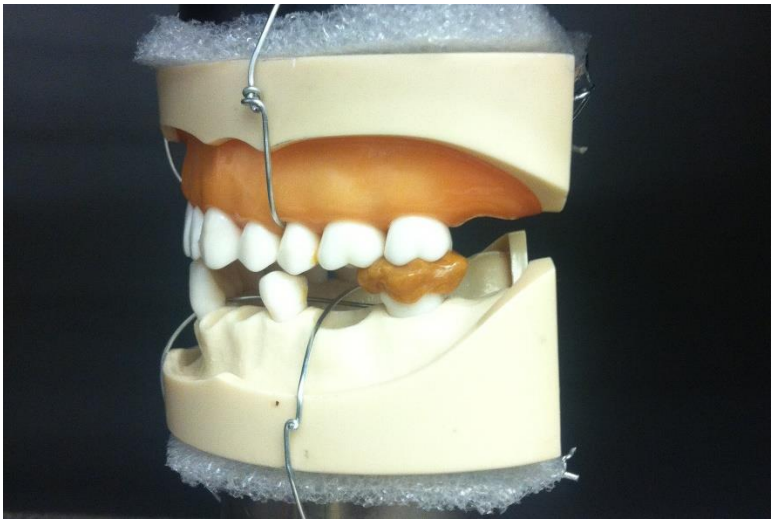
2. The retention force of RPD clasps varied according to the type of tooth and clasp.

3.An algorithm for predicting RPD retention and determining the optimal number of clasps was developed and validated experimentally.

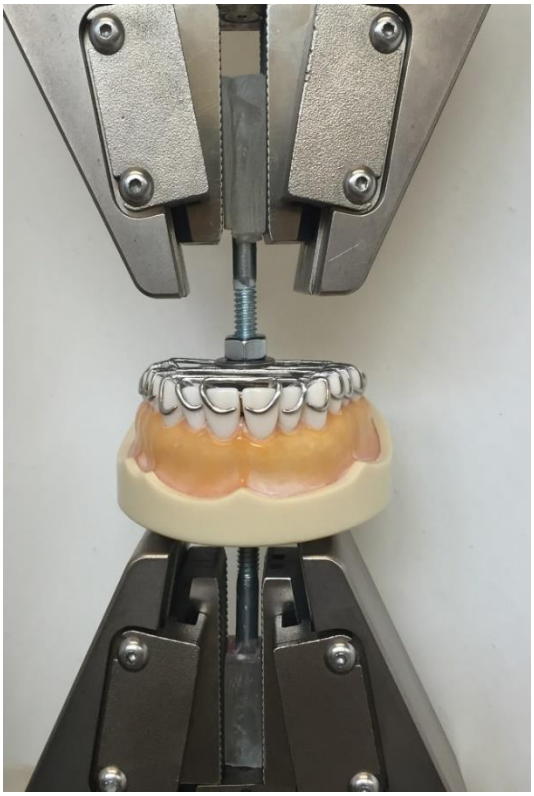
7.7. ACKNOWLEDGMENTS

The authors would like to acknowledge King Saud University (Riyadh, Saudi Arabia), 3DRPD Inc. (Montreal, Canada), and the Natural Sciences and Engineering Research Council (NSERC) Collaborative Research Development and Canada-Discovery grant for their financial support.

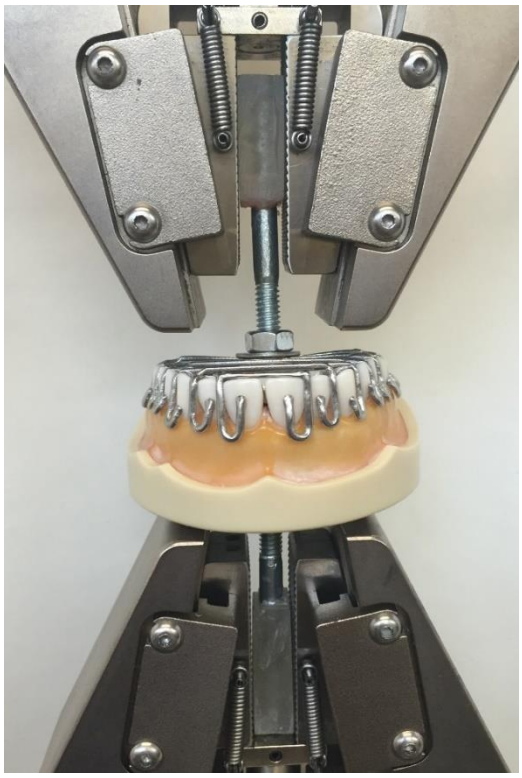
7.8. SUPPLEMENTAL MATERIALS



Supplemental Figure 7.1. Experimental design for measuring forces exerted by food on acrylic resin teeth.



Supplemental Figure 7.2. Experimental design for measuring retention force of circumferential clasps engaging teeth.



Supplemental Figure 7.3. Experimental design for measuring retention force of I-bar clasps engaging teeth.

Supplemental Table 7.1. Retention forces of wrought wire clasps after fatigue for post hoc comparison between tooth types.

Arch	**Tooth**	**Retention Forces (N) of Wrought wire clasp**	***P* Values for post hoc comparison between tooth type**						
			Central	Lateral	Canine	1st Premolar	2nd Premolar	1st Molar	2nd Molar
Maxillary	Central	9.1 ±1.1	-	.824	<.001	.003	.086	<.001	<.001
"	Lateral	8.5 ±1.6	.824	-	<.001	<.001	<.001	<.001	<.001
"	Canine	13.3 ±2.1	<.001	<.001	-	<.001	<.001	1.0	1.0
"	1st Premolar	11.3 ±1.8	.003	<.001	.006	-	.942	.013	.019
"	2nd Premolar	10.7 ±2.1	.086	.001	<.001	.945	-	<.001	<.001
"	1st Molar	13.1 ±0.7	<.001	<.001	1.0	.013	<.001	-	1.0
"	2nd Molar	13.1 ±1.5	<.001	<.001	1.0	.019	<.001	1.0	-
Mandibular	Central	11.0 ±1.2	-	.172	<.001	.117	.433	<.001	<.001
"	Lateral	9.5 ±1.6	.172	-	<.001	<.001	<.001	<.001	<.001
"	Canine	14.0 ±1.7	<.001	<.001	-	<.001	<.001	.926	.988
"	1st Premolar	12.2 ±1.2	.117	<.001	<.001	-	.994	<.001	.015
"	2nd Premolar	11.9 ±1.4	.433	<.001	<.001	.994	-	<.001	<.001
"	1st Molar	14.5 ±1.7	<.001	<.001	.926	<.001	<.001	-	.503
"	2nd Molar	13.6 ±2.6	<.001	<.001	.988	.015	<.001	.50	-

Supplemental Table 7.2. Retention forces of circumferential clasps after fatigue for post hoc comparison between tooth types.

Arch	**Tooth**	**Retention Forces (N) of circumferential clasp**	***P* Values for post hoc comparison between tooth types**						
			Central	Lateral	Canine	1st Premolar	2nd Premolar	1st Molar	2nd Molar
Maxillary	Central	2.9 ±1.2	-	.998	.752	.526	.450	.001	.008
"	Lateral	3.4 ±1.5	.998	-	.963	.842	.780	.004	.031
"	Canine	4.1 ±1.6	.752	.963	-	1.0	1.0	.041	.217
"	1st Premolar	4.4 ±1.5	.526	.842	1.0	-	1.0	.092	.393
"	2nd Premolar	4.5 ±1.1	.450	.780	1.0	1.0	-	.118	.465
"	1st Molar	6.8 ±1.0	.001	.004	.041	092	.118	-	.981
"	2nd Molar	6.1 ±0.9	.008	.031	.217	.393	.465	.981	-
Mandibular	Central	3.0 ±1.4	-	.997	.562	.138	.099	.003	.003
"	Lateral	3.4 ±1.5	.997	-	.879	.369	.283	.011	.014
"	Canine	4.3 ±1.0	.562	.879	-	.970	.934	.170	.197
"	1st Premolar	5.0 ±0.8	.138	.369	.970	-	1.0	.631	.681
"	2nd Premolar	5.1 ±0.6	.099	.283	.934	1.0	-	.729	.775
"	1st Molar	6.2 ±0.9	.003	.011	.170	.631	.729	-	1.0
"	2nd Molar	6.2 ±1.0	.003	.014	.197	.681	.775	1.0	-

Supplemental Table 7.3. Retention forces of I-bar clasps after fatigue for post hoc comparison between tooth types.

Arch	**Tooth**	**Retention Forces (N) of I-bar clasp**	***P* Values for post hoc comparison between tooth types**						
			Central	Lateral	Canine	1st Premolar	2nd Premolar	1st Molar	2nd Molar
Maxillary	Central	2.9 ±1.2	-	.999	1.0	.834	.770	.803	.862
"	Lateral	3.4 ±1.5	.999	-	.979	.584	.507	.546	.623
"	Canine	4.1 ±1.6	1.0	.979	-	.965	.937	.953	.975
"	1st Premolar	4.4 ±1.5	.834	.584	.965	-	1.0	1.0	1.0
"	2nd Premolar	4.5 ±1.1	.770	.507	.937	1.0	-	.118	.465
"	1st Molar	6.8 ±1.0	.803	.546	.953	1.0	1.0	-	1.0
"	2nd Molar	6.1 ±0.9	.862	.623	.975	1.0	1.0	1.0	-
Mandibular	Central	3.0 ±1.4	-	1.0	.806	.265	.647	.659	.548
"	Lateral	3.4 ±1.5	1.0	-	.741	.216	.572	.585	.475
"	Canine	4.3 ±1.0	.806	.741	-	.961	1.0	1.0	.999
"	1st Premolar	5.0 ±0.8	.265	.216	.961	-	.993	.992	.998
"	2nd Premolar	5.1 ±0.6	.647	.572	1.0	.993	-	1.0	1.0
"	1st Molar	6.2 ±0.9	.659	.585	1.0	.992	1.0	-	1.0
"	2nd Molar	6.2 ±1.0	.548	.475	.999	.998	1.0	1.0	-

Supplemental Table 7.4. Constant factor in algorithm for predicting RPD retention, equation 2, for mastication of different type of food on different tooth anatomy and occlusion.

Constant	Condition	Maxillary arch	Mandibular arch
Ka: Constant for tooth anatomy	Anatomical teeth	1.0	1.0
	Non-anatomical teeth	0.89	0.91
Kb: Constant for occlusion type	Class 1 occlusion	1.0	1.0
	Class 2 occlusion	0.96	0.81
	Class 3 occlusion	0.91	0.83
Kc: Constant for food type	Caramel candy	1.0	1.0
	Chewing gum	0.32	0.31
	Bread	0.12	0.12

7.9. REFERENCES

1. Campbell SD, Cooper L, Craddock H, Hyde TP, Nattress B, Pavitt SH, et al. Removable partial dentures: The clinical need for innovation. J Prosthet Dent 2017;118:273-80.

2. Carr AB, Brown DT. McCracken's removable partial prosthodontics. 13th ed. St. Louis, Mosby/Elsevier; 2016. p. 21-4,67-92,112-27.

3. Preshaw P, Walls A, Jakubovics N, Moynihan P, Jepson N, Loewy Z. Association of removable partial denture use with oral and systemic health. J Dent 2011;39:711-9.

4. Hummel SK, Wilson MA, Marker VA, Nunn ME. Quality of removable partial dentures worn by the adult U.S. population. J Prosthet Dent 2002;88:37-43.

5. Saito M, Notani K, Miura Y, Kawasaki T. Complications and failures in removable partial dentures: a clinical evaluation. J Oral Rehabil 2002;29:627-33.

6. Bural C, Buzbas B, Ozatik S, Bayraktar G, Emes Y. Distal extension mandibular removable partial denture with implant support. Eur J Dent 2016;10:566-70.

7. Chaiyabutr Y, Brudvik JS. Removable partial denture design using milled abutment surfaces and minimal soft tissue coverage for periodontally compromised teeth: a clinical report. J Prosthet Dent 2008;99:263-6.

8. Beaumont AJ Jr.Microcomputer-aided removable partial denture design: The next evolution. J Prosthet Dent 1989;62:551-6.

9. Phoenix RD, Cagna DR, DeFreest CF, Stewart KL. Stewart's clinical removable partial prosthodontics. 4th ed. Hanover Park, IL, Quintessence Pub; 2008. p. 1-51,205-48.

10. Chen Q, Wu J, Li S, Lyu P, Wang Y, Li M. An ontology-driven, case-based clinical decision support model for removable partial denture design. Sci Rep 2016;6:27855.

11. Hammond P, Davenport J, Potts A. Knowledge-based design of removable partial dentures using direct manipulation and critiquing. J Oral Rehabil1993;20:115-23.

12. Mendonça EA. Clinical decision support systems: perspectives in dentistry. J Dent Educ 2004;68:589-97.

13. Cabrera CAG, Pinzan A, Cabrera MdC, Henriques JFC, Janson G, Freitas MRd. Biometric study of human teeth. Dental Press J Orthod2011;16:111-22.

14. Tannous F, Steiner M, Shahin R, Kern M. Retentive forces and fatigue resistance of thermoplastic resin clasps. Dent Mater 2012;28:273-8.

15. Frank RP, Nicholls JI. A study of the flexibility of wrought wire clasps. J Prosthet Dent 1981;45:259-67.

16. Khan S, Geerts G. Aesthetic clasp design for removable partial dentures: a literature review. SADJ 2005;60:190-4.

17. Vallittu PK. Fatigue resistance and stress of wrought-steel wire clasps. J Prosthodont1996;5:186-92.

18. Arda T, Arikan A. An in vitro comparison of retentive force and deformation of acetal resin and cobalt-chromium clasps. J Prosthet Dent 2005;94:267-74.

19. Tran C, LaBarre E, Landesman HM. A removable partial denture using an esthetically designed round-rest distal clasp on maxillary anterior abutment teeth: A clinical report. J Prosthet Dent 2009;102:286-9.

20. Alageel O, Abdallah MN, Alsheghri A, Song J, Caron E, Tamimi F. Removable partial denture alloys processed by laser-sintering technique. J Biomed Mater Res B Appl Biomater 2018;106:1174-1185.

21. Almufleh B, Emami E, Alageel O, de Melo F, Seng F, Caron E, et al. Patient satisfaction with laser-sintered removable partial dentures: A crossover pilot clinical trial. J Prosthet Dent 2018;119:560-7.

22. Ledley RS. Dental forces and mastication. J Texture Stud 1971;2:3-17.

23. Chen J. Food oral processing—A review. Food Hydrocolloids 2009;23:1-25.

24. Xu WL, Torrance JD, Chen BQ, Potgieter J, Bronlund JE, Pap JS. Kinematics and experiments of a life-sized masticatory robot for characterizing food texture. IEEE T Ind Electron 2008;55:2121-32.

25. Kashket S, Van Houte J, Lopez LR, Stocks S. Lack of correlation between food retention on the human dentition and consumer perception of food stickiness. J Dent Res 1991;70:1314-9.

26. Grist F. Basic guide to orthodontic dental nursing. Chichester, West Sussex: Blackwell Pub; 2010. p.1-11.

27. VandenBrink JP, Wolfaardt JF, Faulkner MG. A comparison of various removable partial denture clasp materials and fabrication procedure for placing clasps on canine and premolar teeth. J Prosthet Dent1993;70:180-8.

28. Matheson GR, Brudvik JS, Nicholls JI. Behavior of wrought wire clasps after repeated permanent deformation. J Prosthet Dent 1986;55:226-31.

29. Suresh S. Fatigue of materials. 2nd ed. Cambridge: Cambridge University Press; 1998. p. 1-131,423-34.

30. Kim D, Park C, Yi Y, Cho L. Comparison of cast Ti-Ni alloy clasp retention with conventional removable partial denture clasps. J Prosthet Dent 2004;91:374-82.

31. Zhu W, Zeng N, Wang N. Sensitivity, specificity, accuracy, associated confidence interval and ROC analysis with practical SAS implementations. NESUG proceedings: health care and life sciences, Baltimore, MD 2010;19:67.

32. Cheng H, Xu M, Zhang H, Wu W, Zheng M, Li X. Cyclic fatigue properties of cobalt-chromium alloy clasps for partial removable dental prostheses. J Prosthet Dent 2010;104:389-96.

33. Ghazal M, Kern M. The influence of antagonistic surface roughness on the wear of human enamel and nanofilled composite resin artificial teeth. J Prosthet Dent 2009;101:342-9.

34. Sharma D, Bhat BS, Arora H. Restoring anterior aesthetics by a rotational path cast partial denture: an overlooked technique. J Clin Diagn Res 2016;10:ZD11-3.

Chapter 8: Manuscript 3

The previous chapter discussing the developing of a designing guideline for predicting the retention performance of any RPD design. Then, the developed model was tested and validated in the laboratory using 36 RPDs. However, the clinical evidence is still lacking. For this reason, this chapter is to validate the developed model for predicting RPDs retention clinically. In addition, this clinical study also aimed to investigate the factors related to RPD retention that affect patient satisfaction. This chapter is a manuscript submitted for publication and this manuscript including the references formatting and references list is similar to the submitted version with some modifications according to the examiners' reviews.

Design-driven Prediction of Removable Partial Denture Retention Is Associated with Patient Satisfaction.

Author names and affiliations:

Omar Alageel BSc, MSc [a], Nida Ashraf [b], Marion Bessadet DDS, MSc, PhD [c], Emmanuel Nicolas DDS, MSC, PhD, HDR [d], and Faleh Tamimi BDS, MSc, MClinDent, PhD [e]

[a] Doctoral student, Faculty of Dentistry, McGill University, Montreal, Quebec, Canada; and Lecturer, College of Applied Medical Sciences, King Saud University, Riyadh, Saudi Arabia.

[b] Undergraduate student, Faculty of Dentistry, McGill University, Montreal, Quebec, Canada.

[c] Senior Lecturer and Hospital practitioner, Université Clermont Auvergne, CROC, Clermont-Ferrand, France; CHU de Clermont-Ferrand, Service d'Odontologie, Clermont-Ferrand, France.

[d] Professor and Hospital practitioner, Université Clermont Auvergne, CROC, Clermont-Ferrand, France; CHU de Clermont-Ferrand, Service d'Odontologie, Clermont-Ferrand, France.

[e] Associate Professor and Canada Research Chair, Faculty of Dentistry, McGill University, Montreal, QC, Canada.

This manuscript has been submitted to Journal of Prosthetic Dentistry on October 2018.

8.1. ABSTRACT

Statement of problem. Removable partial denture (RPD) is a cost-effective treatment designed to replace missing teeth for partially edentulous patients. However, RPD often presents insufficient retention which results in treatment failure and patient dissatisfaction.

Purpose. The purpose of this study was to investigate the factors related to RPD retention that affect patient satisfaction, to clinically validate a newly published model for predicting RPDs retention based on the number and position of missing teeth and clasps, and to identify the predictions of patient satisfaction in order to improve the guideline for RPD design.

Material and methods. Seventy- five patients treated with 107 RPDs delivered at McGill University Dental Clinic (Montreal, Canada) and Estaing University Hospital (Clermont-Ferrand, France) participated in this study. Data on RPD design was collected from patients' clinical records and the retention of each RPD was tested with the mathematical model designed for predicting RPD retention. Data on patients' satisfaction with their RPDs was collected using a standardized questionnaire (McGill Denture Satisfaction Instrument). Statistical analysis of factors related to RPD retention and patient satisfaction was performed using the chi-square test and Mann-Whitney test, while the developed model for predicting RPD retention was evaluated using sensitivity and specificity analysis.

Results. The average satisfaction score for all RPDs was 8.2±1.7 out of 10. Patients were more satisfied with RPDs in the maxillary arch, tooth-bounded, or retained by ≥3 clasps than with RPDs in the mandibular arch, with free-end saddle, or retained by <3 clasps. The materials used for RPD fabrication (metal-based or acrylic-based), the number of missing teeth, and the presence of indirect retention were not associated with patient satisfaction. Patients were significantly more satisfied with RPD designs predicted by the developed mathematical model to have sufficient

retention than with RPD designs predicted to have insufficient retention. The mathematical model for predicting the RPDs retention showed a clinical specificity of 83% in predicting patient satisfaction.

Conclusion. RPD retention predicted from the number and position of clasps and missing teeth can determine patient satisfaction. In addition, patients' satisfaction with RPDs is influencing by the arch type, the presence of free-end saddle, and the number of the clasp.

CLINICAL IMPLICATIONS.

The model for calculating RPD retention may help design better RPDs with more predictable treatment outcomes.

8.2. INTRODUCTION

Removable partial dentures (RPDs) are effective and affordable treatments for tooth replacement of partially edentulous patients.[1-4] This type of treatment is used by millions of patients worldwide, however, it is reported that 66% of patients wearing RPDs are dissatisfied with their treatment, and around 40% of them stopped wearing them within the first few years of usage.[5-7] Therefore, there is a pressing need to improve this treatment modality. There are various complications associated with RPDs mainly related to inadequate quality and poor design.[1,4,6,7] Improvement in the fabrication process of RPDs, such as the introduction of direct metal laser sintering technology, has improved treatment quality and patient satisfaction.[8,9] However, poor RPD design remains the main reason for patient dissatisfaction as it often results in loss of retention, one of the main reasons for RPD's failure.[1,5,7]

Designing an RPD is very challenging because there are 65534 possible presentations of partial edentulism in each jaw, and no guidelines available to determine the optimal RPD design for each presentation.[10-12] Thus, RPDs are usually designed subjectively based on the preference and experience of dental professionals and may result in poor designs.[12] Recently, it was shown that it is possible to accurately predict the retention of an RPD using a simple calculation based on the RPD design.[12] This mathematical model determines the amount of retention provided by the clasps as well as the dislodging forces generated by food during mastication. By calculating the difference between these 2 values (Figure 8.1), it is possible to predict whether or not an RPD design will provide sufficient retention during mastication.[12] Although the model was validated experimentally in the laboratory, the clinical evidence is still lacking.

Patient satisfaction is an important tool for evaluating the effectiveness and the success of a treatment.[1,13,14] In fact, partial dentures should not only be assessed based on clinical outcomes but also based on patients' feedback and opinions because patient dissatisfaction with the treatment will lead to underuse and subsequent treatment failure.[1,13,15] Although patients' satisfaction with their dentures is subjective and may vary amongst patients, it provides an indication of the treatment success in different aspects such as comfort, appearance, speech, hygiene, mastication ability, and retention of the denture.[8,13-19] One simple way of measuring patient satisfaction is using a self-administered instrument such as the visual analog scale (VAS) questionnaire which has been validated and used effectively in many clinical studies.[8,13-18]

Patient satisfaction with RPD treatments can be influenced by clinical and social factors such as age, gender, experience with previous dentures, socio-economic status, general health, and lifestyle.[13-15,20,21] For instance, patient satisfaction is reported to be greater in patients with a positive previous denture experience, in patients with a greater number of natural teeth in

occlusion, or in older patients, whereas satisfaction is lower in patients with poor health or when an RPD occludes with a denture in the opposing arch.[14,20,21] However, Zlataric et al found that there was no relationship between patients' satisfaction with their RPDs and their age, gender, general health, lifestyle, and socioeconomic status.[15] Awad et al acknowledged that patient satisfaction is affected by their expectations and evaluation of different aspects of their denture such as denture appearance and function.[13]

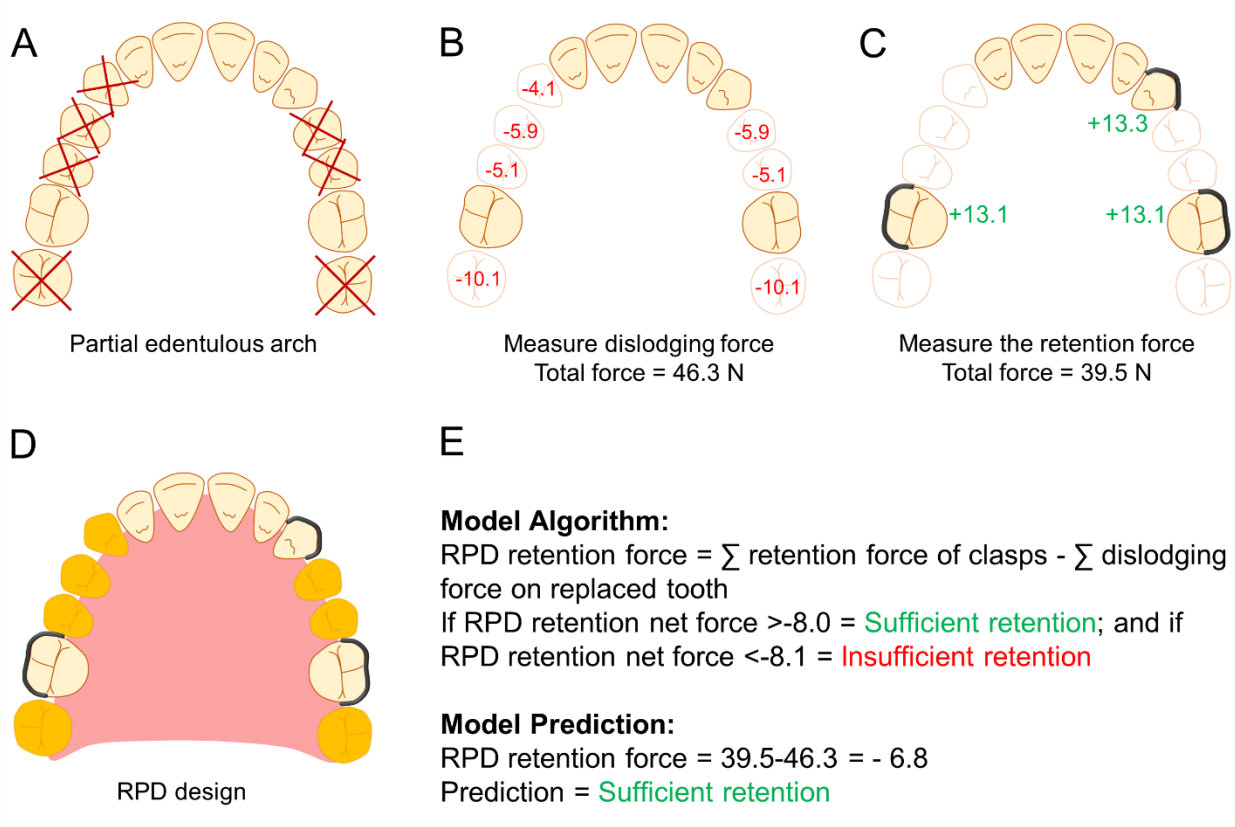


Figure 8.1. Example of testing the RPD retention by model for prediction RPDs retention; (A) partial edentulous arch, (B) determining dislodging force by food mastication on missing teeth, (C) determining retention force of RPD clasps, (D) RPD design, (E) model algorithm and prediction result.[12]

Factors related to RPD retention such as number and location of clasps or missing teeth might also influence patient satisfaction since they can determine the mechanical performance of the RPD.[21] In fact, Wetherell et al found that most RPD failures occurred on free-end saddle configuration than tooth-bounded RPD configurations.[22] This gives us an indication about the possible impact of location of missing teeth on patient satisfaction. Surprisingly, very few studies have investigated the relationship between patient satisfaction and the location of missing teeth, and their findings were dissimilar. Koyama et al found that patient satisfaction correlated with the type of edentulism.[14] On the contrary, Zlataric et al found no relationship between patient satisfaction and location of missing teeth, but concluded that the number of missing teeth on the mandibular arch could play an important role in patient satisfaction.[15] Therefore, investigating factors related to RPD retention, such as the number and location of missing could help to improve the design of RPDs.

Patient subjective reporting on a lack of retention has been associated with unsatisfactory treatment; however, the effect of functional objective retention of RPDs on patient satisfaction has never been investigated because of the lack of adequate tools. Now, the mathematical model for predicting retention of RPDs allows the calculation of RPDs objective retention, but it has not been investigated yet as a predictor of patient satisfaction.[12] The hypothesis of this study was that patient satisfaction with the RPD depends on its estimated retention. Accordingly, the aim of this study was to investigate the factors related to RPD retention that affect patient satisfaction, to clinically validate a newly published model for predicting RPDs retention based on the number and position of missing teeth and clasps. The ultimate goal of this study was to identify the predictions of patient satisfaction in order to improve the guidelines for RPD design.

8.3. MATERIAL AND METHODS

The research protocol was ethically approved by the Institutional Review Board of the Faculty of Medicine at McGill University (A01-B07-16B) and by the local ethical committee of Clermont-Ferrand University Hospital (CE-CIC GREN-09-12; IRB number 5044). Partial edentulous patients treated with a removable partial denture (RPD) at McGill University Dental Clinic (Montreal, Quebec, Canada) between 2012 and 2017 and at Estaing University Hospital (Clermont-Ferrand, France) between 2014 and 2017 were selected for the study. The study design is shown in Figure 8.2. Patients who meet the inclusion criteria were invited to participate in the study. The research study and procedures were explained to the patients who agreed to participate in the study either during their follow-up appointments or by phone. Finally, written consent or verbal consent were obtained from the participants.

For inclusion in the study, adult patients (age above 18 years) of any gender had to be a partial edentulous and have been using RPDs for at least 12 months. The 12-month time period was chosen because RPD clasps usually present with fatigue deformation after few months of usage.[23,24] No restrictions on the type of RPD, type of edentulism, tooth anatomy, and occlusion were imposed. Records of patients who were no longer alive or did not answer the phone or patients with no RPD design available in their records were excluded from the study. In addition, the participants who did not agree to participate or who are not able to provide written or oral consent and questionnaire in English or French were also excluded from the study.

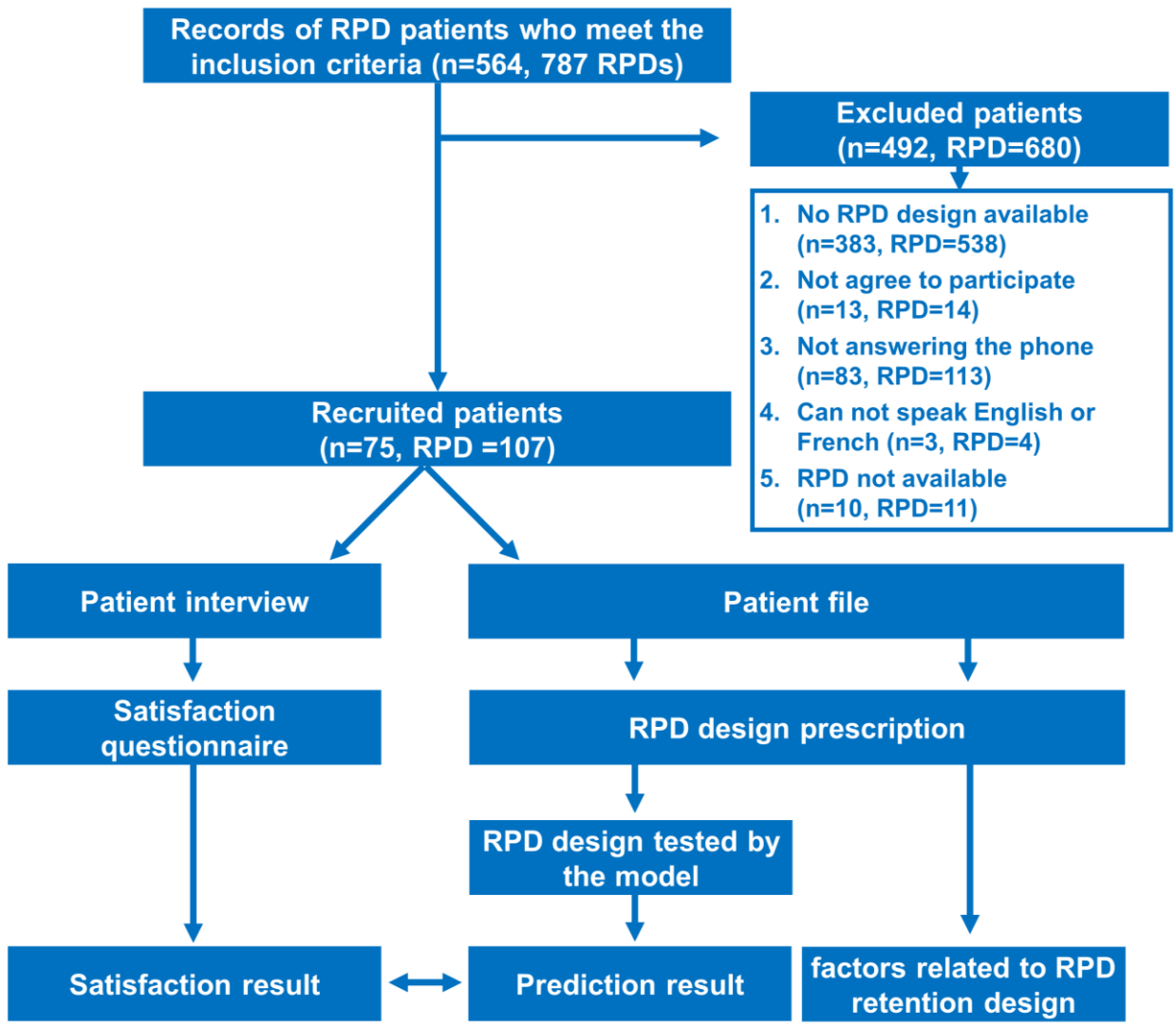


Figure 8.2. Flow diagram showing study design.

The participants were asked to answer the McGill Denture Satisfaction questionnaire (Table 8.1).[25] This questionnaire used 26 questions to evaluate 8 categories related to patient satisfaction about their RPDs such as ease of cleaning (Q1), general satisfaction (Q2), ability to speak (Q3), comfort (Q4), esthetics (Q5), retention and stability (Q6-11), function and ability to masticate different types of food (Q12-25), and general oral condition (Q26). Patients were asked to answer each question from 0 to 100, where zero meant not satisfied and 100 meant very satisfied.

The scale was then converted to a 10-point scale. An average was calculated from all the answers by combining the responses of the 8 satisfaction categories of the questionnaire in order to create an overall satisfaction measurement for each patient.[21] Then, the degree of satisfaction was created by dichotomizing the overall satisfaction measurement using a cut-point of 8.0 (close to the median value of 8.4), and the patients were divided into two categories satisfied (score 8 and above) and dissatisfied (score below 8).[21]

Table 8.1. The list of items of the patient satisfaction questionnaire that was used in this study.

N	Question statement
1	How difficult is it to clean your prosthesis?
2	In general, are you satisfied with your prosthesis?
3	How difficult is it for you to speak because of your prosthesis?
4	Are you satisfied with the comfort of your prosthesis?
5	Are you satisfied with the appearance of your prosthesis?
6	Are you satisfied with the retention (tightness) of your prosthesis?
7	Are you satisfied with the easiness to remove your prosthesis?
8	Does your denture rock forward and backward when you chew?
9	Do you find that your denture comes out easily while chewing?
10	Do you find that your denture comes out easily while speaking?
11	Do you find that your denture comes out easily with your tongue?
12	In general, do you find it difficult to chew food because of your prosthesis?
13	How difficult is it to eat fresh white bread because of your prosthesis?
14	How difficult it is for you to eat hard cheese because of your prosthesis?
15	How difficult it is for you to eat raw carrots because of your prosthesis?
16	How difficult it is for you to eat sliced steak because of your prosthesis?
17	How difficult it is for you to eat raw apples because of your prosthesis?
18	How difficult it is for you to eat lettuce because of your prosthesis?
19	In general, is your food well chewed before swallowing?
20	Are pieces of fresh white bread well chewed before swallowing?
21	Are pieces of hard cheese well chewed before swallowing?
22	Are pieces of raw carrot well chewed before swallowing?
23	Are pieces of sliced steak well chewed before swallowing?
24	Are pieces of raw apple well chewed before swallowing?
25	Are pieces of lettuce well chewed before swallowing?
26	In general, are you satisfied with your oral condition?

The RPD design prescriptions and data for the factors related to RPD retention were retrieved from the participants' files. The data was categorized into groups according to arch type (maxillary or mandibular), RPD type (acrylic or metallic), type of edentulism (free-end saddle or tooth-bounded), number of missing teeth (≤5 or ≥6), number of clasps (2 or ≥3), and presence of indirect retention (present or not). The associations between results from patient's satisfaction questionnaire and factors related to RPD retention were tested.

Using the RPD designs that are available in patients' files, the retention of the selected RPDs were calculated using the mathematical model designed to predict whether or not an RPD will provide sufficient retention during mastication (Figure 8.1).[12] The mathematical model has been made available online at www.ebhnow.com. In order to calculate the retention of each RPD using the mathematical model, the missing teeth were first indicated on the arch shown by the homepage of the model. Then, the type and location of the RPDs clasps were indicated on the arch. Next, the mathematical model calculates the amount of retention force (net force) on the RPD which is the sum of the retention force provided by all the clasps on the RPD subtracted by the dislodging force generated on all the missing teeth.[12] In addition, the model provides the prediction of the RPD retention where it can be sufficient retention if the net force of the RPD was higher than -8 or insufficient retention if the net force of the RPD was lower than -8. Finally, the prediction results from the model for predicting RPD retention were compared with the results from the questionnaire.

In order to calculate the sample size for this study, the correlation coefficient for this study was anticipated to be 0.5 (r) this gives a sample size of at least 64 patients for a power of 90%, type I error α=.05, and type II error β =.01. Chi-square test was used to determine statistical significance in the relationships between the outcomes of patient satisfaction and the factors that

might be related to RPD retention. Moreover, the Mann-Whitney U test was used to test for significant differences between the categorized groups for each satisfaction item. In addition, multiple regression analysis was used to identify the predictions of RPDs retention with the factors associated with patient satisfaction. Statistical analysis for the validation of the model for predicting RPDs retention was tested with sensitivity and specificity analyses that evaluate the accuracy of the model from predicting the retention of the RPD. Statistical software (IBM SPSS Statistics v23.0; IBM Corp) was used for all the statistical analyses (α=.05).

8.4. RESULTS

In total, 107 RPDs were delivered to 75 patients. Table 8.2 shows the distribution of 6 factors that related to RPD retention and their association with patient satisfaction. A total of 72 RPDs (67%) were found to be satisfied with their patients. A significantly larger proportion of patients satisfied with maxillary RPDs (n=47) than with mandibular RPDs (n=60) (P=.026), with tooth-bounded (n=23) than with free-end saddle (n=84) (P=.023), and with RPDs retained by ≥3 clasps (n=65) than by only 2 clasps (n=42) (P=.026). However, there were no statistically significant difference between patient satisfaction and RPD type (metal-based RPDs, n=97; and acrylic-based RPDs, n=10), the number of missing teeth (≥6, n=51; and ≤5, n=56), and presence of indirect retention (present, n=51; and not, n=56).

The average satisfaction score based on the satisfaction questionnaire for all RPDs was 8.2±1.7. Maxillary RPDs were found to have a higher satisfaction score than mandibular RPDs when inquired about general satisfaction (P=.035) (Table 8.3). Patients were more satisfied with the comfort (P=.002), appearance (P=.025), retention (P=.007), mastication ability (P=.004), and oral condition (P=.021) of tooth-bounded RPDs than that of free-end saddle RPDs. Patients with

≤5 missing teeth were more satisfied than patients with ≥6 missing teeth in terms of ease of cleaning (*P*=.017) and the ability of speech (*P*=.028). In addition, RPDs that have ≥3 clasps provided a higher satisfaction score than RPDs with 2 clasps in terms of retention (*P*=.024) and mastication ability (*P*=.003).

The model for predicting RPD retention was validated clinically on the 107 RPDs (Table 8.4). Patients were more satisfied with RPDs predicted by the model to provide sufficient retention than those predicted to provide insufficient retention in terms of comfort (*P*=.007), appearance (*P*=.010), retention and stability (*P*=.008), and mastication ability (*P*<.001) (Table 8.3). In addition, among the 43 RPDs predicted to have acceptable retention by the calculator model, 37 were reported to be satisfactory by the patients. Only 6 RPDs did not follow the positive prediction. Similarly, 29 of 64 RPDs predicted by the model to have insufficient retention were considered dissatisfactory by the patients. However, the other 35 RPDs predicted by the model to have insufficient retention were found satisfactory by the patients. Thus, the model for predicting RPD retention had a specificity of 83%, a sensitivity of 51%, a positive predictive value of 86%, negative predictive value of 45%, and accuracy of 62%.

There were statistically significant associations between the predictions for RPD retention generated by the mathematical model and patients' satisfaction (*P*=.015), type of edentulism (*P*=<.001), the number of clasps (*P*=.002), and the number of missing teeth (*P*=<.001) (Table 8.5). There was no association between the predictions of the RPDs retention and arch location of the RPD (*P*=.861), the type of RPD (*P*=.176), the and presence of indirect retention (*P*=.486).

Table 8.2. Distribution of categorical variables in the study, and the association between patient satisfaction and 6 factors that might be related to RPD retention. Satisfaction degree set at score of 8 and above of overall satisfaction score.

Factors	**Average satisfaction score**	**Satisfied patient N (%)**	**Dissatisfied patient N (%)**	***P***	**OR (95% CI)**
Arch:					
Mandibular	7.8±1.7	35 (58%)	25 (42%)		1
Maxillary	8.5±1.5	37 (79%)	10 (21%)	.026*	2.64 (1.11-6.28)
RPD type:					
Acrylic	7.8±1.8	5 (50%)	5 (50%)		1
Metallic	8.1±1.6	67 (70%)	30 (30%)	.221	2.23 (0.60-8.29)
Type of edentulism:					
Free-end saddle	8.0±1.6	52 (62%)	32 (38%)		1
Tooth-bounded	9.0±1.2	20 (87%)	3 (13%)	.023*	4.10 (1.13-14.92)
Number of missing teeth:					
≤5	8.3±1.5	42 (75%)	14 (25%)		1
≥6	7.9±1.7	30 (59%)	21 (41%)	.075	0.48 (0.21-1.10)
Number of clasps:					
2	7.7±1.6	23 (55%)	19 (45%)		1
≥3	8.4±1.6	49 (75%)	16 (15%)	.026*	2.53 (1.10-5.80)
Indirect retention:					
Present	8.4±1.4	36 (65%)	20 (35%)		1
Not	8.0±1.9	36 (70%)	15 (30%)	.488	1.16 (0.49-2.75)

** Indicates a significant difference at P<.05.*

Table 8.3. Results of average satisfaction score (± SD) of each item in questionnaire according to categorical groups of 6 factors that might be related to RPDs retention and outcomes of model prediction.

	Q1: Cleaning	**Q2: General satisfaction**	**Q3: Speech**	**Q4: Comfort**	**Q5: Appearance**	**Q6-11: Retention**	**Q12-25: Mastication**	**Q26: Oral condition**
All RPDs	9.0±1.8	8.0±2.4	8.2±2.6	7.8±2.4	9.0±1.7	8.3±2.1	7.8±2.0	8.3±2.1
Arch:								
Mandibular	8.9±1.9	7.5±2.6***j**	8.0±2.8	7.5±2.6	8.7±2.1	8.0±2.3	7.7±2.1	8.1±2.1
Maxillary	9.2±1.6	8.5±2.0***j**	8.5±2.2	8.3±2.0	9.3±0.9	8.6±1.7	8.3±1.7	8.6±1.9
RPD type:								
Acrylic	8.4±1.9	7.9±2.1	7.4±2.7	8.6±1.7	8.6±1.4	8.0±1.7	7.4±2.2	7.8±2.3
Metallic	9.1±1.8	8.0±2.5	8.3±2.5	7.8±2.4	9.0±1.8	8.3±2.1	8.3±2.0	8.2±1.7
Type of edentulism:								
Free-end saddle	8.9±2.0	7.8±2.6	8.1±2.7	7.5±2.4***b**	8.8±1.8***i**	8.0±2.2***e**	7.7±2.0***d**	8.0±1.6***h**
Tooth-bounded	9.6±0.6	8.7±1.7	8.9±1.7	9.0±1.6***b**	9.6±0.7***i**	9.2±1.0***e**	8.9±1.6***d**	9.1±1.5***h**
Number of missing teeth:								
≤5	9.5±1.0***h**	8.0±2.3	8.8±2.2***i**	7.8±2.4	9.1±1.6	8.5±1.8	8.2±1.9	8.4±2.0
≥6	8.5±2.3***h**	8.0±2.6	7.7±2.8***i**	7.9±2.3	8.9±2.3	8.0±2.3	7.7±2.1	8.2±2.1
Number of clasps:								
2	8.8±1.9	7.9±2.3	8.1±2.7	7.6±2.3	8.9±1.5	8.0±1.7***h**	7.3±2.1***c**	8.1±2.1
≥3	9.2±1.7	8.0±2.5	8.3±2.5	8.0±2.4	9.0±1.7	8.4±2.3***h**	8.4±1.8***c**	8.4±2.0
Indirect retentions:								
Present	8.8±2.2	8.2±2.0	8.8±2.2	8.1±2.2	8.9±1.8	8.6±1.6	8.2±1.9	8.7±2.3
Not	9.5±1.1	7.8±2.9	7.9±2.8	7.4±2.6	9.2±1.7	7.9±2.6	7.8±2.1	8.0±2.3
Calculation retention:								
Insufficient	8.8±2.0	7.7±2.6	7.9±2.7	7.4±2.4***e**	8.7±1.8***g**	7.9±2.2***f**	7.5±2.0***a**	8.0±2.2
Sufficient	9.4±1.5	8.5±2.1	8.8±2.2	8.4±2.2***e**	9.4±1.5***g**	8.7±1.7***f**	8.7±1.7***a**	8.7±1.8

** Indicates significant difference between categorized groups P: a<.001, b=.002, c=.003, d=.004, e=007, f=.008, g=.01, h=.02, i=.03, j=.04.*

Table 8.4. Validity of predicting RPDs retentions by mathematical model of predicting RPD retention at different satisfaction degrees of overall patient satisfaction and for different type of edentulism.

Cut-points	**Specificity (95% CI)**	**Sensitivity (95% CI)**	**Positive predictive value (95% CI)**	**Negative predictive value (95% CI)**	**Accuracy (95% CI)**
Satisfaction score of 6.0	73% (45-92%)	45% (34-55%)	91% (81-96%)	18% (13-23%)	48% (39-58%)
Satisfaction score of 7.0	76% (53-92%)	48% (37-59%)	89% (79-95%)	26% (21-33%)	53% (43-63%)
Satisfaction score of 8.0	83% (66-93%)	51% (39-63%)	86% (74-93%)	45% (39-52%)	62% (52-71%)
Satisfaction score of 8.5	78% (63-88%)	58% (45-71%)	76% (64-84%)	61% (53-69%)	67% (58-76%)
Satisfaction score of 9.0	70% (57-80%)	63% (46-78%)	53% (42-64%)	77% (69-84%)	67% (58-76%)
Satisfaction score of 8.0 for tooth-bounded	NC	95% (75-100%)	86% (85-85%)	NC	83% (61-95%)
Satisfaction score of 8.0 for free-end saddle	91% (75-98%)	35% (22-49%)	86% (66-95%)	46% (40-52%)	56% (45-67%)

CI: Confidence interval; NC: Cannot be calculated.

Table 8.5. Results of association between the predictions of model for predicting RPD retention and outcomes of overall patient satisfaction set at score of above 8 and the factors related to RPD retention.

Factors	Average satisfaction score	Calculated Retention N		Crude		Adjusted **	
		Sufficient	Insufficient	*P*	OR (95% CI)	*P*	OR (95% CI)
Satisfied patients:							
No	7.7±1.7	6	29		1		1
Yes	8.8±1.3	37	35	.001*	5.11 (1.89-13.79)	.015*	4.63 (1.35-15.92)
Arch:							
Mandibular	7.8±1.7	21	39		1		1
Maxillary	8.5±1.5	22	25	.150	1.63 (0.75 -3.57)	.861	0.92 (0.34-2.47)
Type of edentulism:							
Free-end saddle	8.0±1.6	21	63		1		1
Tooth-bounded	9.0±1.2	22	1	<.001*	66.0 (8.39-519.9)	<.001*	67.70 (8.41-544.89)
Number of clasps:							
2	7.7±1.6	4	38		1		1
≥3	8.4±1.6	39	26	<.001*	14.25 (4.54-44.70)	.002*	6.47 (1.94-21.45)
RPD type:							
Acrylic	7.8±1.8	3	7		1		1
Metallic	8.1±1.6	40	57	.370	1.63 (0.40-6.72)	.176	5.76 (0.45-73.03)
Number of missing teeth:							
≤5	8.3±1.5	35	21		1		1
≥6	7.9±1.7	8	43	<.001*	0.11 (0.04-0.28)	<.001*	0.45 (0.01-0.22)
Indirect retention:							
Yes	8.4±1.4	19	32		1		1
No	8.0±1.9	21	25	.264	0.70 (0.31-1.56)	.486	1.46 (0.50-4.26)

** Indicates a significant difference at P<.05; CI: Confidence interval; OR: Odd Ratio; ** Ratio adjusted for the type of arch and the type of the edentulism.*

8.5. DISCUSSION

The hypothesis of this study was confirmed by showing that patient satisfaction with their RPD could be influenced by its estimated retention. The study showed that other factors can influence the patient's satisfaction with their RPDs, such as arch location, the presence of free-end saddle, and the number of clasps.

Previous studies showed that there is a great need to improve the quality of RPD design by reducing the biomechanical problems associated with them.[1,6,21] Hence, a new engineering model for predicting and optimizing the retention of RPDs was established.[12] This retrospective study is the first one to evaluate and clinically validate this model for predicting RPD retention. Thus, this study opens the door for further digitalization of the fabrication process and design optimization of RPDs, and consequently, has the potential to enhance the quality of life for millions of patients worldwide.[1,8,9]

The results of this study concluded that 67% of RPDs were considered satisfactory by patients. This finding disagrees with previous studies reporting that only one-third of RPDs were found satisfactory by the patients.[6] This difference might be related to the superior quality of RPDs in this study resulting from the use of the laser-sintering technique for the fabrication of most of the RPDs. Laser-sintered RPDs have been shown to offer enhanced patient satisfaction over conventional ones.[8] In addition, less satisfied patients were less likely to respond or agree to participate in the study which may also lead to overestimation of the level of satisfaction of patients with their RPDs.

In the satisfaction questionnaire, ease of cleaning and appearance scored higher than other items since RPDs generally provide good access for cleaning and have an aesthetically pleasing appearance. However, comfort and ability to masticate different foods scored lower than other

items indicating the challenges faced by patients who wear RPDs. Thus, some of these challenges may be addressed by improving the retention of RPDs.

The results of this study showed that several factors influence the retention of RPDs and patient satisfaction (Tables 8.2 and 8.3). Increased satisfaction of patients with maxillary RPDs might be related to the superior stability provided by the palatal support while the mandibular arch often presents reduced stability because of residual bone resorption and tongue movement.[14,21] The superior stability of maxillary RPDs results in enhanced speech, comfort, retention, mastication ability, and oral condition for the patients. In addition, maxillary teeth are generally more visible than mandibular teeth, and restoring maxillary teeth is more important for patients in terms of appearance and esthetics.

The study showed that patients who wore tooth-bounded RPDs were more satisfied than patients with free-end saddle RPDs. This was in agreement with a previous study showing that patients who had tooth-bounded RPDs were more likely to continue using their RPDs after 5 years of delivery than patients with free-end saddle RPDs.[14] One of the reasons for this is likely because of the biomechanical problems associated with free-end saddle RPDs.[21] Previous studies indicated that free-end saddle RPDs present higher failure rates of the clasps than tooth-bounded RPDs due to the stress on the denture-base that results in loss of retention.[3,22] Moreover, free-end saddle RPDs present problems of discomfort and pain, especially the mandibular RPDs, which might explain the reasons for low satisfaction scores in comfort and oral condition.[14]

Clasps play an important role in patient satisfaction since they are an important element in RPD retention and resist dislodging forces generated by mastication and functional muscle movements.[1,2] In this study, it was found that patients who had RPDs retained by ≥3 clasps were more satisfied than patients who had RPDs retained by 2 clasps because of the additional retention

and improved function provided by the greater number of clasps. Furthermore, there was no association found between the presence of indirect retentions with patient satisfaction in this study. However, the number of rests including indirect retentions can influence the fit, comfort, appearance, and chewing ability of the patient.[14]

The model for predicting the retention of RPDs was validated clinically that 86% of RPDs predicted to have sufficient retention presented sufficient retention during food mastication. There were only 6 of 43 RPDs that gave a false positive and did not follow the positive prediction. The reason for this might be because 4 of these 6 RPDs were made with cast frameworks that present less precision and poor mechanical properties than laser-sintered RPDs.[9] In fact, patients who wore those 4 cast RPDs were dissatisfied with their retention; while, patients who wore the 2 laser-sintered frameworks were satisfied with their retention but not with other satisfaction categories such as comfort, speech ability, and oral condition.

A total of 51% of RPDs predicted to provide insufficient retention were actually satisfactory for their patients. It was found that some patients in this group were not satisfied with the retention of the RPDs, as predicted by the model, but they were generally satisfied with the RPDs because of other factors contributing to the patients' satisfaction. This meant that this model is excellent for predicting the satisfactory RPDs but not as for predicting the dissatisfactory RPDs. Therefore, this model is an excellent tool for designing the retention of RPDs.

The sensitivity and accuracy of the model for predicting patients' satisfaction for tooth-bounded RPDs were much higher than for free-end saddle RPDs (Table 8.4). The low sensitivity and accuracy with free-end saddle RPDs were probably related to other important satisfaction items than the retention such as pain, comfort, and oral condition. In fact, most dissatisfied patients

with their RPDs had free-end saddle RPDs, and the free-end saddle RPDs often causes pain and discomfort that may explain the issue.[14]

The predictions of RPD retention were associated with patients' satisfaction indicating the efficiency of the model for predicting RPD retention. Patients with RPDs predicted by the model to provide sufficient retention were more satisfied than patients with RPDs predicted to have insufficient retention (Table 8.5). The high satisfaction levels are known to depend on retention and mastication ability, and they are also probably related to the enhanced comfort, appearance, and overall satisfaction shown by patients using RPDs with predicted satisfactory retention (Table 8.3). As expected, the predictions of the RPDs retention were associated with the number of clasps in the RPDs and with the number of missing teeth according to the model algorithm.[12]

Limitations in this study should be considered in future studies to improve the model further for better designing RPDs with predictable treatment outcome. Currently, the model of predicting RPDs retention is only considering the retention of the RPDs but not for the other satisfaction items. The model should also take into account other satisfaction items such as comfort, aesthetic, oral condition. In addition, the accuracy of the model for predicting the retention performance for free-end saddle RPDs is lower than for tooth-bounded RPDs. One important reason is that the free-end saddle RPDs are supported by the residual ridges and the indirect retainers that prevent RPD denture base movement.[4] However, these specials characteristics of the free-end saddle RPDs were not accounted for in the prediction model. Hence, changes should be made to the model to take into account these factors related to the secondary retention of free-end saddle bases.

8.6. CONCLUSION

Within the limitations of this study, the following conclusions were drawn:

1. RPD retention can determine patient satisfaction.

2. The model for predicting RPD retention is an excellent tool for designing the retention of RPDs, and it can predict the patient's perception of RPD retention.

3. Patient satisfaction was greater with RPDs that were on the maxillary arch, tooth-bounded, or retained by 3 clasps and more.

4. Patients are likely to be satisfied with RPDs designed according to the predictions of the new model designed for optimizing RPD retention.

8.7. ACKNOWLEDGEMENTS

The authors would like to acknowledge King Saud University (Riyadh, Saudi Arabia), 3DRPD Inc. (Montreal, Canada), and the Natural Sciences and Engineering Research Council (NSERC) Collaborative Research Development and Canada-Discovery grant for their financial support. The authors would like to acknowledge Dr. Jeffrey Myers for helping us access the patient's records at McGill dental clinics.

8.8. REFERENCES

1. Campbell SD, Cooper L, Craddock H, Hyde TP, Nattress B, Pavitt SH, et al. Removable partial dentures: The clinical need for innovation. J Prosthet Dent 2017;118:273-80.

2. Carr AB, Brown DT. McCracken's removable partial prosthodontics. 13th ed. St. Louis, Mosby/Elsevier; 2016. p. 21-4,67-92,112-27

3. Vermeulen AHBM, Keltjens HMAM, van't Hof MA, Kayser AF. Ten-year evaluation of removable partial dentures: Survival rates based on retreatment, not wearing and replacement. J Prosthet Dent 1996;76:267-72.

4. Sakar O. Removable Partial Dentures A Practitioners' Manual. 1st ed. Switzerland, Springer; 2015. p. 3-17,25-73.

5. Bilhan H, Erdogan O, Ergin S, Celik M, Ates G, Geckili O. Complication rates and patient satisfaction with removable dentures. J Adv Prosthodont 2012;4:109-15.

6. Hummel SK, Wilson MA, Marker VA, Nunn ME. Quality of removable partial dentures worn by the adult U.S. population. J Prosthet Dent 2002;88:37-43.

7. Saito M, Notani K, Miura Y, Kawasaki T. Complications and failures in removable partial dentures: a clinical evaluation. J Oral Rehabil 2002;29:627-33.

8. Almufleh B, Emami E, Alageel O, de Melo F, Seng F, Caron E, et al. Patient satisfaction with laser-sintered removable partial dentures: A crossover pilot clinical trial. J Prosthet Dent 2018;119:560-7.

9. Alageel O, Abdallah MN, Alsheghri A, Song J, Caron E, Tamimi F. Removable partial denture alloys processed by laser-sintering technique. J Biomed Mater Res B Appl Biomater 2018;106:1174-85.

10. Beaumont AJ Jr.Microcomputer-aided removable partial denture design: The next evolution. J Prosthet Dent 1989;62:551-6.

11. Bonnet G, Lance C, Bessadet M, Tamini F, Veyrune JL, Francois O, et al. Teaching removable partial denture design:‘METACIEL’, a novel digital procedure. Int J Med Educ 2018;9:24.

12. Alageel O, Alsheghri A, Algezani S, Tamimi F. Determining the Retention of Removable Partial Dentures. In. J Prosthet Dent 2018.

13. Awad MA, Feine JS. Measuring patient satisfaction with mandibular prostheses. Community Dent Oral Epidemiol 1998;26:400-5.

14. Koyama S, Sasaki K, Yokoyama M, Sasaki T, Hanawa S. Evaluation of factors affecting the continuing use and patient satisfaction with removable partial dentures over 5 years, J Prosthodont Res 2010;54:97-101.

15. Zlatarić DK, Čelebić A, Valentić-Peruzović M, Jerolimov V, Pandurić J. A survey of treatment outcomes with removable partial dentures. J Oral Rehabil 2003;30:847-54.

16. Sato Y, Hamada S, Akagawa Y, Tsuga K. A method for quantifying overall satisfaction of complete denture patients. J Oral Rehabil 2000;27:952-7.

17. Feine JS, Lund JP. Measuring chewing ability in randomized controlled trials with edentulous populations wearing implant prostheses. J Oral Rehabil 2006;33:301-8.

18. Michaud P-L, de Grandmont P, Feine JS, Emami E. Measuring patient-based outcomes: Is treatment satisfaction associated with oral health-related quality of life? J Dent 2012;40:624-31.

19. Armellini D, Von Fraunhofer JA. The shortened dental arch: a review of the literature. The J Prosthet Dent 2004;92:531-35.

20. Weinstein M, Schuchman J, Lieberman J, Rosen P. Age and denture experience as determinants in patient denture satisfaction. J Prosthet Dent 1988;59:327-9.

21. Frank RP, Milgrom P, Leroux BG, Hawkins NR. Treatment outcomes with mandibular removable partial dentures: a population-based study of patient satisfaction. J Prosthet Dent 1998;80:36-45.

22. Wetherell JD, Smales RJ. Partial denture failures: A long-term clinical survey. J Dent 1980;8:333-40.

23. Kim D, Park C, Yi Y, Cho L. Comparison of cast Ti-Ni alloy clasp retention with conventional removable partial denture clasps. J Prosthet Dent 2004;91:374-82.

24. Matheson GR, Brudvik JS, Nicholls JI. Behavior of wrought wire clasps after repeated permanent deformation. J Prosthet Dent 1986;55:226-31.

25. de Albuquerque Junior RF, Lund JP, Tang L, Larivée J, de Grandmont P, Gauthier G, et al. Within-subject comparison of maxillary long-bar implant-retained prostheses with and without palatal coverage: patient-based outcomes. Clin Oral Implants Res 2000;11:555-65.

Chapter 9: General Conclusion

The findings of this work proved our hypothesis by confirming that the digital technology including the laser sintering/melting technology and the advance algorithm can improve the design, accuracy, property, and clinical performance of the removable partial denture treatments.

Specifically, Co-Cr alloys processed by the laser-sintering techniques are more precise and present better fatigue resistance and mechanical properties for removable partial dentures than cast alloys due to their better homogeneity and small grain size. In addition, both laser-sintered/melted Co-Cr alloys are biocompatible and present similar biocompatibility properties to the cast Co-Cr alloy. Accordingly, laser-sintered/melted RPDs could present clinical benefits over the cast ones in terms of fitting and mechanical stability.

In this thesis, it was investigated that the force generated by food mastication on teeth varied according to the type of tooth, occlusion, and food. Whereas, the retention force of RPD clasps varied according to the type of tooth and clasp. This thesis suggested that for an RPD to withstand mastication without being dislodged, the sum of the retention forces provided by each clasp should be higher than the sum of the dislodging forces generated by food mastication on each replaced missing tooth. Accordingly, a model for predicting RPD retention and determining the optimal number of clasps was developed using the advanced algorithm, and this model was validated experimentally and tested clinically. The model for predicting RPD retention may help design better RPDs with more predictable treatment outcomes. Also, it was found that patients' perception of RPD retention can be predicted mathematically by the developed model.

It was investigated that in this thesis, patient satisfaction was greater with RPDs that were on the maxillary arch, tooth-bounded, or retained by 3 clasps and more. Also, RPD retention can

determine patient satisfaction. Patients are likely to be satisfied with RPDs designed according to the predictions of the new model designed for optimizing RPD retention.

Chapter 10: Limitations and Future Directions

The model for predicting RPDs retention presented limitations, and some of these limitations are related to the differences between patients and to the setup of the experimental part. First, many parameters that vary among patients were not tested such as tooth anatomy, the height of tooth crown, mastication mechanics (such as mastication speed, angle, and food volume), and the path of insertion and removal of the RPDs. In fact, these different parameters might change the dislodging forces that were considered in our developed model. Also, the retention forces of the clasps that were calculated in this thesis can be different in the clinic. Clasp retention experiments were performed in a dry ambient condition with acrylic resin teeth; this might underestimate clasp retention force in the oral environment because of the adhesive effect of saliva in tooth-clasp interactions.

The simulated mastication that was used in the thesis for measuring the masticatory forces was done with an angle of 90° which might be not similar to the mastication mechanics of patients. However, we think that the dislodging forces that calculated by our setup experiment will be similar or lower than the forces at different mastication angle since the dislodging force that pulls food away from the teeth. These forces generate during the contacts between food and teeth, and the contact is only occurring at a small distance which probably will not be affected by the mastication angle. In addition, based on the physics principle, the force at a diagonal angle will be lower than the force at the vertical angle, and this has no negative effects on predictions from the developed model.

The variations in the fabrication process of RPDs (such as clasps materials, thickness, length, and undercut depth) between dental clinics and laboratories might limit the validity of the algorithm in clinical practices. Moreover, other clasp types and other retentive features of RPDs

such as rotational partial dentures were not addressed in this algorithm; including these permutations would acknowledge deviations in the algorithm.

Currently, the model of predicting RPDs retention is only considering the retention of the RPDs but not for the other satisfaction items. In addition, the accuracy of the model for predicting the retention performance for free-end saddle RPDs is lower than for tooth-bounded RPDs. One important reason is that the free-end saddle RPDs are supported by the residual ridges and the indirect retainers that prevent RPD denture base movement. However, these specials characteristics of the free-end saddle RPDs were not accounted for in the prediction model.

Limitations in this study should be considered in future studies to improve the model further for better designing RPDs with predictable treatment outcome. In future studies, the developed model should be improved further for better designing RPDs with predictable treatment outcome. For example, the model should include other factors related to the RPD retention including other clasp types, other retentive features, and the secondary retention for the free-end saddle bases. The model should also take into account other factors that are important for denture satisfaction such as comfort and aesthetic. Finally, automatization of the designing process of the RPD could help dental professionals better determining the optimal RPD design.

Chapter 11: Appendix

10.1. Ethical Approval

McGill

Faculty of Medicine
3655 Promenade Sir William Osler #633
Montreal, QC H3G 1Y6

Faculté de médecine
3655, Promenade Sir William Osler #633
Montréal, QC H3G 1Y6

Fax/Télécopieur: (514) 398-3870
Tel/Tél: (514) 398-3124

January 22, 2016

Dr. Faleh Tamimi
Faculty of Dentistry
3640 University – Room M-64
Montreal, Quebec H3A 0A7

RE: IRB Study Number A01-B07-16B
Does the design of removable partial dentures (RPDs) influence patients' satisfaction?

Dear Dr. Tamimi,

Thank you for submitting the above-referenced study for an ethics review, on behalf of your students, Omar Alageel and Nida Ashraf.

As this study involves no more than minimal risk, and in accordance with Articles 2.9 and 6.12 of the 2nd Edition of the Canadian Tri-Council Policy Statement of Ethical Conduct for Research Involving Humans (TCPS 2) and U.S. Title 45 CFR 46, Section 110 (b), paragraph (1), we are pleased to inform you that ethics approval for the protocol, study instruments and consent form (December 22, 2015) is provided via an expedited review by the IRB Co-Chair on January 22, 2016. The ethics approval is valid until **January 2017**. The study proposal will be presented for corroborative approval at the next meeting of the Committee and a certification document will be issued to you at that time.

Before initiating the study, please remove "Dose" from the title in all the study documents and replace with "Does"

A review of all research involving human subjects is required on an annual basis in accord with the date of initial approval. The annual review should be submitted at least one month before **January 2017**. Please inform the IRB promply of any modifications that may occur to the study over the next twelve months.

Sincerely,
Carolyn Ells
Carolyn Ells, PhD
Co-Chair
Institutional Review Board

cc: Omar Alageel
Nida Ashraf
A01-B07-16B

McGill

Faculty of Medicine
3655 Promenade Sir William Osler #633
Montreal, QC H3G 1Y6

Faculté de médecine
3655, Promenade Sir William Osler #633
Montréal, QC H3G 1Y6

Fax/Télécopieur: (514) 398-3870
Tél/Tel: (514) 398-3124

17 October 2017

Dr. Faleh Tamimi
Faculty of Dentistry
2001 avenue McGill-College, 5th Floor
Montreal QC H3A 1G1

RE: IRB Study Number A01-B07-16B
Does the design of removable partial dentures (RPDs) influence patients' satisfaction?

On 16 October 2017, at a meeting of the Institutional Review Boad, the following amendment received a full Board review and approval:

- Protocol Amendment (dated September 22, 2107);
- Subject Information and Consent Form, IRB dated September 2017.

The Investigator is reminded of the requirement to report all IRB approved protocol and consent form modifications to the Research Ethics Offices (REOs) for the participating study sites, when applicable. Please contact the individual research ethics offices for instructions on how to proceed. Research funds may be withheld, and/or the study's data may be revoked for failing to comply with this requirement.

Regards,

Roberta Palmour, PhD
Chair
Institutional Review Board

cc: A01-B07-16B

McGill

Faculty of Medicine
3655 Promenade Sir William Osler #633
Montreal, QC H3G 1Y6

Faculté de médecine
3655, Promenade Sir William Osler #633
Montréal, QC H3G 1Y6

Fax/Télécopieur: (514) 398-3870
Tél/Tel: (514) 398-3124

January 16, 2018

Dr. Faleh Tamimi
Faculty of Dentistry
3640 University – Room M-64
Montreal, Quebec H3A 0A7

RE: IRB Study Number A01-B07-16B
Does the design of removable partial dentures (RPDs) influence patients' satisfaction?

Dear Dr. Tamimi,

Thank you for submitting an application for Continuing Review for the above-referenced study.

The study progress report was reviewed and an expedited re-approval was provided on January 15, 2018. The ethics certification renewal is valid from **January 8, 2018 to January 7, 2019**.

The Investigator is reminded of the requirement to report all IRB approved protocol and consent form modifications to the Research Ethics Offices (REOs) for the participating hospital sites. Please contact the individual hospital REOs for instructions on how to proceed. Research funds may be withheld and / or the study's data may be revoked for failing to comply with this requirement.

Should any modification or unanticipated development occur prior to the next review, please notify the IRB promptly. Regulation does not permit the implementation of study modifications prior to IRB review and approval.

Sincerely,

Carolyn Ells, PhD
Co-Chair
Institutional Review Board

cc: Omar Alageel
Nida Ashraf
A01-B07-16B

10.2. Consent Form

SUBJECT INFORMATION AND CONSENT FORM

STUDY TITLE: Does the design of removable partial dentures (RPDs) influence patients' satisfaction?

Principal Investigators (PIs):

Dr. Faleh Tamimi
Faculty of Dentistry, McGill University
Room M-64, Strathcona Anatomy & Dentistry,
3640 University Street
Montreal, Quebec H3A 2B2
Tel.: 514-398-7203 ext 09654 | Fax: 514-398-8900
Email: faleh.tamimimarino@mcgill.ca

Students:

Omar Alageel, Ph.D student
Faculty of Dentistry, McGill University
Strathcona Anatomy and Dentistry
3640 University Street
Montreal, Quebec H3A 2B2
omar.alageel@mail.mcgill.ca

Nida Ashraf, undergraduate DMD student
Faculty of Dentistry, McGill University
Faculty of Dentisty
2001 McGill College Ave
Montreal, Quebec H3A 1G1
Email: nida.ashraf@mail.mcgill.ca

December 22, 2015

ORAL CONSENT SCRIPT

Hello, my name is (Nida Ashraf/Omar Alageel). I am working on a study being conducted by McGill University, Faculty of dentistry.

Do you have a few minutes to discuss the study?

- If yes, continue below.
- If no, but he/she is interested to participate in the study, determine a better time to call back.
- If no, thank them for their time.

You have receiving a removable partial dentures (RPDs) at McGill undergraduate dental clinic. So, we invite you to take part in a research study at McGill undergraduate dental clinic which seeks to get more information about your satisfaction with the treatment.

Taking part in this study is entirely voluntary. We are going to give you information about this research and invite you to be part of this research. Before deciding to participate and make a decision, you should understand the full content of this study, and take your time to make a decision. You can ask questions if there is anything you do not understand.

The purpose of this study is to investigate the relationship between removable partial dentures (RPDs) design and patient's satisfaction. This research will involve only filling a questionnaire for one time. You will be asked 28 questions about your satisfaction with the treatment. We will use the questionnaire to draw a comparison between the patients' satisfaction of the dentures and the dentures design. Your participation will last about [7 minutes].

You will not be paid for participating in this study, and there will be no cost to you to participate in this study. Also, there are no harms or risks associated with your participation in this study.

McGill University will respect the confidentiality and no information that discloses your identity will be released or published unless required by law. We will not collect any personal information that is related you. We will only be recording the file number present on your file but the number will be coded for your privacy. The results from this study may be published, but no information that discloses the identity of any patient will be released or published except when required by law.

Your participation in this study is voluntary. You may refuse to participate or discontinue your participation at any time without explanation.

December 22, 2015

Do you have any questions?

Do you need time to think about participating in this study?

- If yes, determine a better time to call back.
- If no, continue below.

Do you agree to participate in this study?

☐ Yes: Document oral consent below and continue with questionnaire.

☐ No: Thank them for their time

_____	_____	_____
Participant code	Date	

Person Obtaining Consent

I have read this form to the participant. I explained the study and answered all questions from the participant. In my judgment, the subject has demonstrated comprehension of the information. The subject has provided oral consent to participate in this study.

_____

Name and Title (Print)

_____	_____
Signature of Person Obtaining Consent	Date

December 22, 2015

10.3. Questionnaire

VAS PRACTICE QUESTIONAIRE

ASSESSMENT OF PROSTHESIS

Patient's code:

Date: / / Phase I Phase II

y y m m d d

We would like to know how satisfied you are with your **present** prosthesis. Read each of the following questions and draw a vertical line on the horizontal line, where you think your answer best fits. In the case where a question doesn't apply to you, for example if you don't eat a certain type of food, write a brief explanation on the line.

1. Ease of cleaning

Please indicate how difficult it is to clean your prosthesis and mouth

Extremely difficult ——— Not at all difficult 1. □□□

2. General satisfaction

In general, are you satisfied with your prosthesis?

Not at all satisfied ——— Extremely satisfied 2. □□□

3. Ability to speak

Please indicate how difficult it is for you to speak because of your prosthesis?

Extremely difficult ——— Not at all difficult 3. □□□

4. Comfort

Are you satisfied with the comfort of your prosthesis?

Not at all satisfied ——— Extremely satisfied 4. □□□

5. Aesthetics

Are you satisfied with the appearance of your prosthesis?

1/4

© McGill University, 2015

Not at all satisfied	_____	Extremely satisfied

5. ☐☐☐

6. Retention and stability

Are you satisfied with the retention (tightness) of your prosthesis?

Not at all satisfied	_____	Extremely satisfied

6. ☐☐☐

Are you satisfied with the easiness to remove your prosthesis?

Not at all satisfied	_____	Extremely satisfied

7. ☐☐☐

Does your prosthesis rocks forward and backward in your mouth when you chew?

All of the time	_____	Never

8. ☐☐☐

Do you find that your prosthesis comes out easily while **chewing**?

Always	_____	Never

9. ☐☐☐

Do you find that your prosthesis comes out easily while **speaking**?

Always	_____	Never

10. ☐☐☐

Do you find that your prosthesis comes out easily **with your tongue**?

Always	_____	Never

11. ☐☐☐

7. Ability to chew

In general, do you find it difficult to chew food because of your prosthesis?

Extremely difficult	_____	Not at all difficult

12. ☐☐☐

Please indicate how difficult it is for you to eat **fresh white bread** because of your prosthesis?

Extremely difficult	_____	Not at all difficult

13. ☐☐☐

Please indicate how difficult it is for you to eat **hard cheese** because of your prosthesis?

Extremely difficult	_____	Not at all difficult

14. ☐☐☐

© McGill University, 2015

Please indicate how difficult it is for you to eat **raw carrots** because of your prosthesis?

Extremely difficult _____ Not at all difficult 15. ☐☐☐

Please indicate how difficult it is for you to eat **sliced steak** because of your prosthesis?

Extremely difficult _____ Not at all difficult 16. ☐☐☐

Please indicate how difficult it is for you to eat **raw apples** because of your prosthesis?

Extremely difficult _____ Not at all difficult 17. ☐☐☐

Please indicate how difficult it is for you to eat **lettuce** because of your prosthesis?

Extremely difficult _____ Not at all difficult 18. ☐☐☐

8. Function

In general, is your food well chewed before swallowing?

Badly chewed _____ Very well chewed 19. ☐☐☐

Are pieces of **fresh white bread** well chewed before swallowing?

Badly chewed _____ Very well chewed 20. ☐☐☐

Are pieces of **hard cheese** well chewed before swallowing?

Badly chewed _____ Very well chewed 21. ☐☐☐

Are pieces of **raw carrot** well chewed before swallowing?

Badly chewed _____ Very well chewed 22. ☐☐☐

Are pieces of **sliced steak** well chewed before swallowing?

Badly chewed _____ Very well chewed 23. ☐☐☐

Are pieces of **raw apple** well chewed before swallowing?

© McGill University, 2015

Badly chewed _____ Very well chewed 24. ☐☐☐

Are pieces of **lettuce** well chewed before swallowing?

Badly chewed _____ Very well chewed 25. ☐☐☐

9. Oral condition

In general, are you satisfied with your oral condition?

Not at all satisfied _____ Extremely satisfied 26. ☐☐☐

Do you believe that your oral condition has a negative effect on your general health? 27.

No ○$_0$ Yes ○$_1$

If yes, why?

_____

_____

_____

_____

****

Is there any kind of problem with your upper or lower prosthesis that you would like to report? 28.

No ○$_0$ Yes ○$_1$

If yes, please, describe?

_____

_____

_____

_____

© McGill University, 2015

10.4. Publications

Published Articles by the Candidate during Ph.D. Studies. The The collection of articles includes the articles published as a first author and as a co-author.

Article I

ARTICLE IN PRESS

THE JOURNAL OF PROSTHETIC DENTISTRY

RESEARCH AND EDUCATION

Determining the retention of removable partial dentures

Omar Alageel, BSc, MSc,[a] Ammar A. Alsheghri, BSc, MSc,[b] Suliman Algezani, BSc, MSc,[c] Eric Caron, DDS, MSc,[d] and Faleh Tamimi, BDS, MSc, PhD[e]

ABSTRACT

Statement of problem. Removable partial dentures (RPDs) provide a cost-effective treatment for millions of partially edentulous patients worldwide. However, they often fail because of loss of retention. One reason for this problem is lack of precise guidelines for designing retentive RPDs.

Purpose. The purpose of this in vitro study was to determine the forces produced by food and clasps during mastication to develop an algorithm for predicting RPD retention and to help determine the optimal number of clasps.

Material and methods. The forces that food exerts on acrylic resin teeth during simulated mastication and the retention forces provided by clasps (wrought wire, circumferential, and I-bar) engaging on teeth were measured using a universal testing machine. A statistical analysis was performed with a 1-way ANOVA and repeated-measures ANOVA while the developed algorithm was evaluated by using sensitivity and specificity analysis.

Results. The force exerted by food mastication on each individual tooth ranged between 1.7 and 12.2 N, depending on the type of tooth, tooth anatomy, occlusion, and food. The retention force of the clasps after cyclic testing ranged between 2.9 and 14.5 N, depending on the type of tooth abutment and clasp. Using these measurements, an algorithm was developed to predict RPD retention. The algorithm was confirmed experimentally on 36 RPDs, showing a sensitivity of 96%, specificity of 100%, and an accuracy of 97%.

Conclusions. The forces generated by food mastication on teeth varied according to the type of tooth, occlusion, and food. The retention force of RPD clasps varied according to the type of tooth and clasp. An algorithm for predicting RPD retention and determining the optimal number of clasps was developed and validated experimentally. (J Prosthet Dent 2018;■:■-■)

Removable partial dentures (RPDs) are cost-effective and functional dental prostheses that are used to restore missing teeth in partially edentulous patients.[1,2] An RPD is a treatment option that can improve the quality of life for millions of patients worldwide; over 13% of the adult population in North America and Europe wear RPDs.[3,4] However, many complications are associated with RPDs, mainly related to inadequate quality and poor design.[1,5,6] Indeed, poor RPD design results in insufficient retention, which is the main reason for treatment failure and patient dissatisfaction.[5-7]

Designing RPDs is challenging because there are 65 534 possible forms of partial edentulism, and the available design guidelines lack scientific evidence and do not cover all edentulism forms.[2,8,9] Therefore, RPDs are designed subjectively based on the experience of dental professionals, which could often result in inadequate designs.[10] In fact, many dentists delegate design work to dental technicians due to their extensive design experience.[11] Knowledge-based systems are available for designing RPDs that provide the most appropriate RPD design based on a database of previous patients.[10,12] However, RPD designs in the database might be inadequate and inappropriate because they were designed subjectively based on operator experience.

The authors would like to acknowledge King Saud University (Riyadh, Saudi Arabia), 3DRPD Inc (Montreal, Canada), and the Natural Sciences and Engineering Research Council (NSERC) Collaborative Research and Development, Canada, Discovery grant for their financial support.

[a]Doctoral student, Faculty of Dentistry, McGill University, Montreal, Canada; and Lecturer, College of Applied Medical Sciences, King Saud University, Riyadh, Saudi Arabia.
[b]Doctoral student, Department of Mining and Materials Engineering, McGill University, Montreal, Canada.
[c]Graduate student, Faculty of Dentistry, McGill University, Montreal, Canada.
[d]Researcher, Dental Wings, Montreal, Canada.
[e]Associate Professor and Canada Research Chair, Faculty of Dentistry, McGill University, Montreal, Canada.

Article II

Society For Biomaterials

Removable partial denture alloys processed by laser-sintering technique

Omar Alageel,[1,2] Mohamed-Nur Abdallah ,[1] Ammar Alsheghri,[3] Jun Song,[3] Eric Caron,[4] Faleh Tamimi[1]

[1]Faculty of Dentistry, McGill University, Montreal, Quebec, Canada
[2]College of Applied Medical Sciences, King Saud University, Riyadh, Saudi Arabia
[3]Department of Mining and Materials Engineering, McGill University, Montreal, Quebec, Canada
[4]3DRPD Inc., Montreal, Quebec, Canada

Received 9 November 2016; revised 3 May 2017; accepted 12 May 2017
Published online 31 May 2017 in Wiley Online Library (wileyonlinelibrary.com). DOI: 10.1002/jbm.b.33929

Abstract: Removable partial dentures (RPDs) are traditionally made using a casting technique. New additive manufacturing processes based on laser sintering has been developed for quick fabrication of RPDs metal frameworks at low cost. The objective of this study was to characterize the mechanical, physical, and biocompatibility properties of RPD cobalt–chromium (Co–Cr) alloys produced by two laser-sintering systems and compare them to those prepared using traditional casting methods. The laser-sintered Co–Cr alloys were processed by the selective laser-sintering method (SLS) and the direct metal laser-sintering (DMLS) method using the Phenix system (L-1) and EOS system (L-2), respectively. L-1 and L-2 techniques were 8 and 3.5 times more precise than the casting (CC) technique ($p < 0.05$). Co–Cr alloys processed by L-1 and L-2 showed higher ($p < 0.05$) hardness (14–19%), yield strength (10–13%), and fatigue resistance (71–72%) compared to CC alloys. This was probably due to their smaller grain size and higher microstructural homogeneity. All Co–Cr alloys exhibited low porosity (2.1–3.3%); however, pore distribution was more homogenous in L-1 and L-2 alloys when compared to CC alloys. Both laser-sintered and cast alloys were biocompatible. In conclusion, laser-sintered alloys are more precise and present better mechanical and fatigue properties than cast alloys for RPDs. © 2017 Wiley Periodicals, Inc. J Biomed Mater Res Part B: Appl Biomater, 106B: 1174–1185, 2018.

Key Words: laser-sintering, cobalt–chromium (Co–Cr), removable partial dentures, fatigue resistance, biocompatibility

How to cite this article: Alageel O, Abdallah MN, Alsheghri A, Song J, Caron E, Tamimi F. 2018. Removable partial denture alloys processed by laser-sintering technique. J Biomed Mater Res Part B 2018:106B:1174–1185.

INTRODUCTION

Removable partial dentures (RPDs) are simple and cost-effective prostheses that can restore missing teeth in partially edentulous patients, and thus improving their quality of life.[1,2] This type of treatment has an important impact on the life of millions of patients in the world; indeed, over 13% of the adult population in North America and Europe wear RPDs.[1,3] RPD frameworks are commonly made of cobalt–chromium (Co–Cr) alloys because of their suitable cost and mechanical properties, and their excellent corrosion resistance and biocompatibility.[4]

RPD frameworks are traditionally fabricated using the casting (lost-wax) technique that has been used in dentistry for more than a century.[5,6] The casting technique is a very laborious manual process that involves making a wax replica of the object, making a mold of the object, and then cast the melted metal into the mold. Owing to its complexity, this technique is strongly influenced by the skill of the dental technician.[5,7] Moreover, producing RPDs by casting technique not only is time consuming and costly but may also generates low precision and ill-fitting frameworks.[7,8]

Different methods were introduced in the last few decades for fabricating RPD frameworks without using casting techniques.[6,9,10] A new additive manufacturing (AM) process based on laser-sintering has been developed for processing 3-D metal objects. The laser-sintering technique combines computer-aided design (CAD) of any products and their subsequent fabrication using a high-power laser that fuses metal powder in a layer-by-layer pattern.[5,6,10–12] The laser-sintering technique enables the fabrication of complex 3-D objects quickly with high precision (20 μm) and at low cost.[10–15]

Laser-sintering technology can be described using different terminologies, such as selective laser melting (SLM), selective laser-sintering (SLS), or direct metal laser-sintering (DMLS).[6,9,12,13] SLM involves full melting of the metal

Correspondence to: F. Tamimi, BDS, MSc, MClinDent, PhD; e-mail: faleh.tamimimarino@mcgill.ca
Contract grant sponsors: King Saud University (Riyadh, Saudi Arabia), 3DRPD Inc., and Natural Sciences and Engineering Research Council (NSERC) Collaborative Research Development
Contract grant sponsor: Fonds de recherche du Québec – Nature et technologies (FRQNT)
Contract grant sponsor: Fondation de l'Ordre des dentists du Quebec (FODQ)

© 2017 WILEY PERIODICALS, INC.

Article III

Acta Biomaterialia 80 (2018) 425–434

Contents lists available at ScienceDirect

Acta Biomaterialia

journal homepage: www.elsevier.com/locate/actabiomat

Full length article

Bio-inspired and optimized interlocking features for strengthening metal/polymer interfaces in additively manufactured prostheses

Ammar A. Alsheghri [a], Omar Alageel [b,c], Mohamed Amine Mezour [b], Binhan Sun [a], Stephen Yue [a], Faleh Tamimi [b], Jun Song [a,*]

[a] Department of Mining and Materials Engineering, McGill University, Montreal, QC, Canada
[b] Faculty of Dentistry, McGill University, Montreal, QC, Canada
[c] College of Applied Medical Sciences, King Saud University, Riyadh, Saudi Arabia

ARTICLE INFO

Article history:
Received 27 June 2018
Received in revised form 28 August 2018
Accepted 13 September 2018
Available online 21 September 2018

Keywords:
Metal/polymer interface
Biomimetic design
Interlocking features
Finite Element Analysis (FEA)
Additive manufacturing

ABSTRACT

Biomedical and dental prostheses combining polymers with metals often suffer failure at the interface. The weak chemical bond between these two dissimilar materials can cause debonding and mechanical failure. This manuscript introduces a new mechanical interlocking technique to strengthen metal/polymer interfaces through optimized additively manufactured features on the metal surface. To reach an optimized design of interlocking features, we started with the bio-mimetic stress-induced material transformation (SMT) optimization method. The considered polymer and metal materials were cold-cured Poly(methyl methacrylate) (PMMA) and laser-sintered Cobalt-Chromium (Co-Cr), respectively. Optimal dimensions of the bio-inspired interlocking features were then determined by mesh adaptive direct search (MADS) algorithm combined with finite element analysis (FEA) and tensile experiments such that they provide the maximum interfacial tensile strength and stiffness while minimizing the stress in PMMA and the displacement of PMMA at the Co-Cr/PMMA interface. The SMT optimization process suggested a Y-shape as a more favorable design, which was similar to mangrove tree roots. Experiments confirmed that our optimized interlocking features increased the strength of the Co-Cr/PMMA interface from 2.3 MPa (flat interface) to 34.4 ± 1 MPa, which constitutes 85% of the tensile failure strength of PMMA (40.2 ± 1 MPa).

Statement of Significance

The objective of this study was to improve metal/polymer interfacial strength in dental and orthopedic prostheses. This was achieved by additive manufacturing of optimized interlocking features on metallic surfaces using laser-sintering. The interlocking design of the features, which was a Y-shape similar to the roots of mangrove trees, was inspired by a bio-memetic optimization algorithm. This interlocking design lowered the PMMA displacement at the Co-Cr/PMMA interface by 70%, enhanced the interfacial strength by more than 12%, and increased the stiffness by 18% compared with a conventional bead design, meanwhile no significant difference was found in the toughness of both designs.

© 2018 Acta Materialia Inc. Published by Elsevier Ltd. All rights reserved.

1. Introduction

Biomedical devices combining polymers with metal alloys often suffer from mechanical failure at the metal/polymer interface [1–3]. For example, dental prostheses and orthopedic devices, in which metal frameworks are joined with acrylic resin bases, present a weak bond between these components that can cause debonding and subsequent catastrophic mechanical failure at the metal/polymer interface [3,4].

Poly(methyl methacrylate) (PMMA) is extensively used for dental prostheses as well as in orthopedic devices because of its biocompatibility, excellent esthetic, and mechanical properties [5]. Metal frameworks are commonly made from Cobalt-Chromium (Co-Cr) alloys because of their suitable cost, excellent mechanical properties, corrosion resistance, and biocompatibility [6]. The tensile strength between PMMA and grit-blasted Co-Cr is only about 5 MPa [7], possibly due to the weak acid-base bond PMMA forms with metal oxide [8]. Therefore, mechanical and chemical methods

* Corresponding author at: Department of Mining and Materials Engineering, McGill University, 3610 University Street, Montreal, QC H3A 2B2, Canada.
E-mail address: jun.song2@mcgill.ca (J. Song).

https://doi.org/10.1016/j.actbio.2018.09.029
1742-7061/© 2018 Acta Materialia Inc. Published by Elsevier Ltd. All rights reserved.

Article IV

DENTAL MATERIALS 34 (2018) 1474–1482

Available online at www.sciencedirect.com

ScienceDirect

journal homepage: www.intl.elsevierhealth.com/journals/dema

An analytical model to design circumferential clasps for laser-sintered removable partial dentures

Ammar A. Alsheghri[a], *Omar Alageel*[b,c], *Eric Caron*[d], *Ovidiu Ciobanu*[b], *Faleh Tamimi*[b], *Jun Song*[a,*]

[a] *Department of Mining Materials Engineering, McGill University, Montreal, QC, Canada*
[b] *Faculty of Dentistry, McGill University, Montreal, QC, Canada*
[c] *College of Applied Medical Sciences, King Saud University, Riyadh, Saudi Arabia*
[d] *3DRPD Inc., Montreal, QC, Canada*

ARTICLE INFO

Article history:
Received 10 August 2017
Received in revised form
5 April 2018
Accepted 7 June 2018

Keywords:
Removable partial dentures (RPDs)
Laser-sintering
Circumferential clasp design
Cobalt–chromium (Co–Cr)
Plastic deformation
Fatigue failure
Finite element analysis (FEA)
Undercut
Retention force
Stress

ABSTRACT

Objective. Clasps of removable partial dentures (RPDs) often suffer from plastic deformation and failure by fatigue; a common complication of RPDs. A new technology for processing metal frameworks for dental prostheses based on laser-sintering, which allows for precise fabrication of clasp geometry, has been recently developed. This study sought to propose a novel method for designing circumferential clasps for laser-sintered RPDs to avoid plastic deformation or fatigue failure.

Methods. An analytical model for designing clasps with semicircular cross-sections was derived based on mechanics. The Euler–Bernoulli elastic curved beam theory and Castigliano's energy method were used to relate the stress and undercut with the clasp length, cross-sectional radius, alloy properties, tooth type, and retention force. Finite element analysis (FEA) was conducted on a case study and the resultant tensile stress and undercut were compared with the analytical model predictions. Pull-out experiments were conducted on laser-sintered cobalt–chromium (Co–Cr) dental prostheses to validate the analytical model results.

Results. The proposed circumferential clasp design model yields results in good agreement with FEA and experiments. The results indicate that Co–Cr circumferential clasps in molars that are 13 mm long engaging undercuts of 0.25 mm should have a cross-section radius of 1.2 mm to provide a retention of 10 N and to avoid plastic deformation or fatigue failure. However, shorter circumferential clasps such as those in premolars present high stresses and cannot avoid plastic deformation or fatigue failure.

Significance. Laser-sintered Co–Cr circumferential clasps in molars are safe, whereas they are susceptible to failure in premolars.

© 2018 The Academy of Dental Materials. Published by Elsevier Inc. All rights reserved.

* *Corresponding author at*: Department of Mining and Materials Engineering, McGill University, 3610 University Street, Montreal, QC, H3A 2B2, Canada.
E-mail addresses: ammar.alsheghri@mail.mcgill.ca (A.A. Alsheghri), omar.alageel@mail.mcgill.ca (O. Alageel), eric.caron@dental-wings.com (E. Caron), ovidiu.ciobanu@mail.mcgill.ca (O. Ciobanu), faleh.tamimimarino@mcgill.ca (F. Tamimi), jun.song2@mcgill.ca (J. Song).
https://doi.org/10.1016/j.dental.2018.06.011
0109-5641/© 2018 The Academy of Dental Materials. Published by Elsevier Inc. All rights reserved.

Article V

DENTAL MATERIALS 33 (2017) e393–e404

Available online at www.sciencedirect.com

ScienceDirect

journal homepage: www.intl.elsevierhealth.com/journals/dema

Metal-composite adhesion based on diazonium chemistry

Yara Oweis[a,1], *Omar Alageel*[a,b,1], *Paige Kozak*[a], *Mohamed-Nur Abdallah*[a], *Jean-Marc Retrouvey*[a], *Marta Cerruti*[c], *Faleh Tamimi*[a,*]

[a] *Faculty of Dentistry, McGill University, Montreal, QC, Canada*
[b] *College of Applied Medical Sciences, King Saud University, Riyadh, Saudi Arabia*
[c] *Department of Mining and Materials Engineering, McGill University, Montreal, QC, Canada*

ARTICLE INFO

Article history:
Received 16 November 2016
Received in revised form
11 June 2017
Accepted 13 July 2017

Keywords:
Diazonium
Composite
Adhesive
Metal-composite adhesive
Bis-GMA

ABSTRACT

Objective. Composite resins do not adhere well to dental alloys. This weak bond can result in failure at the composite-metal interface in fixed dental prostheses and orthodontic brackets. The aim of this study was to develop a new adhesive, based on diazonium chemistry, to facilitate chemical bonding between dental alloys and composite resin.

Methods. Samples of two types of dental alloys, stainless steel and cobalt chromium were primed with a diazonium layer in order to create a surface coating favorable for composite adhesion. Untreated metal samples served as controls. The surface chemical composition of the treated and untreated samples was analyzed by X-ray photoelectron spectroscopy (XPS) and the tensile strength of the bond with composite resin was measured. The diazonium adhesive was also tested for shear bond strength between stainless steel orthodontic brackets and teeth.

Results. XPS confirmed the presence of a diazonium coating on the treated metals. The coating significantly increased the tensile and shear bond strengths by three and four folds respectively between the treated alloys and composite resin. Conclusion: diazonium chemistry can be used to develop composite adhesives for dental alloys.

Significance. Diazonium adhesion can effectively achieve a strong chemical bond between dental alloys and composite resin. This technology can be used for composite repair of fractured crowns, for crown cementation with resin based cements, and for bracket bonding.

© 2017 The Academy of Dental Materials. Published by Elsevier Ltd. All rights reserved.

1. Introduction

Composite resins are widely used in modern dental practice due to their ability to bind to acid etched tooth structure, and their good handling, mechanical, and esthetic properties. Composites are used in a variety of dental applications such as temporary and permanent restorations, cementation and repair of indirect restorations as well as bonding of orthodontic brackets [1,2]. However, the weak adherence of dental composites to dental alloys frequently leads to clinical problems such as failures at the resin-metal interface in fixed dental prostheses and de-bonding of orthodontic bracket [1,3,4].

* *Corresponding author at*: Faculty of Dentistry, McGill University, 3640 University Street, Montreal, QC H3A 0C7, Canada.
E-mail address: faleh.tamimimarino@mcgill.ca (F. Tamimi).
[1] Equal contribution.
http://dx.doi.org/10.1016/j.dental.2017.07.017
0109-5641/© 2017 The Academy of Dental Materials. Published by Elsevier Ltd. All rights reserved.

Article VI

DENTAL MATERIALS 31 (2015) 105–114

Available online at www.sciencedirect.com

ScienceDirect

journal homepage: www.intl.elsevierhealth.com/journals/dema

Bonding metals to poly(methyl methacrylate) using aryldiazonium salts

Omar Alageel[a,b], *Mohamed-Nur Abdallah*[a], *Zhong Yuan Luo*[c], *Jaime Del-Rio-Highsmith*[d], *Marta Cerruti*[c,*], *Faleh Tamimi*[a,**]

[a] *Faculty of Dentistry, McGill University, Montreal, QC, Canada*
[b] *College of Applied Medical Sciences, King Saud University, Riyadh, Saudi Arabia*
[c] *Department of Mining and Materials Engineering, McGill University, Montreal, QC, Canada*
[d] *Faculty of Dentistry, Complutense University of Madrid, Madrid, Spain*

ARTICLE INFO

Article history:
Received 24 January 2014
Received in revised form
15 July 2014
Accepted 3 November 2014

Keywords:
Dental prosthesis
Bonding diazonium
Poly(methyl methacrylate)
Titanium
Stainless steel

ABSTRACT

Objectives. Many dental devices, such as partial dentures, combine acrylic and metallic parts that are bonded together. These devices often present catastrophic mechanical failures due to weak bonding between their acrylic and metallic components. The bonding between alloys and polymers (e.g. poly(methyl methacrylate), PMMA) usually is just a mechanical interlock, since they do not chemically bond spontaneously. The aim of this study was to develop a new method to make a strong chemical bond between alloys and polymers for dental prostheses based on diazonium chemistry.

Methods. The method was based on two steps. In the first step (primer), aryldiazonium salts were grafted onto the metallic surfaces. The second step (adhesive) was optimized to achieve covalent binding between the grafted layer and PMMA. The chemical composition of the treated surfaces was analyzed with X-ray photoelectron spectroscopy (XPS), and the tensile or shear bonding strength between metals and poly(methyl methacrylate) was measured.

Results. XPS and contact angle measurements confirmed the presence of a polymer coating on the treated metallic surfaces. Mechanical tests showed a significant increase in bond strength between PMMA and treated titanium or stainless steel wire by 5.2 and 2.5 folds, respectively, compared to the untreated control group ($p < 0.05$).

Significance. Diazonium chemistry is an effective technique for achieving a strong chemical bond between alloys and PMMA, which can help improve the mechanical properties of dental devices.

© 2014 Academy of Dental Materials. Published by Elsevier Ltd. All rights reserved.

* *Correspondence to*: Department of Mining and Materials Engineering, McGill University, Room 2M020, 3610 University Street, Montreal, QC H3A 0C5, Canada. Tel.: +1 514 398 5496; fax: +1 514 398 4492.
** *Correspondence to*: Faculty of Dentistry, McGill University, Room M64, 3640 University Street, Montreal, QC, H3A 0C7, Canada. Tel.: +1 514 398 7203x09654; fax: +1 514 398 8900.
E-mail addresses: marta.cerruti@mcgill.ca (M. Cerruti), faleh.tamimimarino@mcgill.ca (F. Tamimi).

http://dx.doi.org/10.1016/j.dental.2014.11.002
0109-5641/© 2014 Academy of Dental Materials. Published by Elsevier Ltd. All rights reserved.

Article VII

THE JOURNAL OF PROSTHETIC DENTISTRY

CLINICAL RESEARCH

Patient satisfaction with laser-sintered removable partial dentures: A crossover pilot clinical trial

Balqees Almufleh, BDS, MSc,[a] Elham Emami, BDS, MSc, PhD,[b] Omar Alageel, BSc, MSc,[c] Fabiana de Melo, DDS, MSc,[d] Francois Seng, DDS, MSc,[e] Eric Caron, DDS, MSc,[f] Samer Abi Nader, DDS, MSc,[g] Ashwaq Al-Hashedi, BDS, MSc,[h] Rubens Albuquerque, DDS, MSc, PhD,[i] Jocelyne Feine, DDS, PhD,[j] and Faleh Tamimi, BDS, PhD[k]

ABSTRACT

Statement of problem. Clinical data regarding newly introduced laser-sintered removable partial dentures (RPDs) are needed before this technique can be recommended. Currently, only a few clinical reports have been published, with no clinical studies.

Purpose. This clinical trial compared short-term satisfaction in patients wearing RPDs fabricated with conventional or computer-aided design and computer-aided manufacturing (CAD-CAM) laser-sintering technology.

Material and methods. Twelve participants with partial edentulism were enrolled in this pilot crossover double-blinded clinical trial. Participants were randomly assigned to wear cast or CAD-CAM laser-sintered RPDs for alternate periods of 30 days. The outcome of interest was patient satisfaction as measured using the McGill Denture Satisfaction Instrument. Assessments was conducted at 1, 2, and 4 weeks. The participant's preference in regard to the type of prosthesis was assessed at the final evaluation. The linear mixed effects regression models for repeated measures were used to analyze the data, using the intention-to-treat principle. To assess the robustness of potential, incomplete adherence, sensitivity analyses were conducted.

Results. Statistically significant differences were found in patients' satisfaction between the 2 methods of RPD fabrication. Participants were significantly more satisfied with laser-sintered prostheses than cast prostheses in regard to general satisfaction, ability to speak, ability to clean, comfort, ability to masticate, masticatory efficiency, and oral condition ($P<.05$). At the end of the study, 5 participants preferred the laser-sintered, 1 preferred the cast RPD, and 3 had no preference.

Conclusions. The use of CAD-CAM laser-sintering technology in the fabrication of removable partial dentures may lead to better outcomes in terms of patient satisfaction in the short term. The conclusion from this pilot study requires confirmation by a larger randomized controlled trial. Clinical Trial: ClinicalTrials.gov. A study about patient satisfaction with laser-sintered removable partial dentures; NCT02769715. (J Prosthet Dent 2018;119:560-7)

Funded by 3DRPD Company and King Saud University, Riyadh, Saudi Arabia.

[a]Doctoral student, Faculty of Dentistry, McGill University, Montreal, Quebec, Canada; and Lecturer, Department of Prosthetic Dental Sciences, College of Dentistry, King Saud University, Riyadh, Saudi Arabia.
[b]Associate Dean of Knowledge Transfer and Internationalization and Associate Professor, Department of Restorative Dentistry, Department of Social and Preventive Medicine, Faculty of Dentistry, School of Public Health, University of Montreal, Montreal, Quebec, Canada.
[c]Doctoral student, Faculty of Dentistry, McGill University, Montreal, Quebec, Canada; and Teaching assistant, College of Applied Medical Sciences, King Saud University, Riyadh, Saudi Arabia.
[d]Doctoral student, Department of Periodontology, School of Dentistry, School of Dentistry, Federal University of Rio Grande do Sul, Rio Grande do Sul, Brazil.
[e]Faculty Lecturer and Clinical Instructor, Department of Prosthodontics, McGill University, Montreal, Quebec, Canada.
[f]Researcher, Dental Wings, Montreal, Quebec, Canada.
[g]Associate Professor, Department of Prosthodontics, McGill University, Montreal, Quebec, Canada.
[h]Faculty of Dentistry, Division of Restorative Dentistry, McGill University, Montreal, Quebec, Canada.
[i]Associate professor, Department of Dental Materials and Prosthodontics, Faculty of Dentistry of Ribeirão Preto, University of São Paulo, São Paulo, Brazil.
[j]Professor, Oral Health Sciences, Faculty of Dentistry, and Associate Member, Department of Epidemiology and Biostatistics, Department of Oncology, Faculty of Medicine, McGill University, Montreal, Quebec, Canada.
[k]Faculty of Dentistry, Division of Restorative Dentistry, McGill University, Montreal, Quebec, Canada.