T.C.
YEDİTEPE UNIVERSITY
INSTITUTE OF HEALTH SCIENCES
DEPARTMENT OF PROSTHETIC DENTISTRY

IN VITRO EVALUATION OF THE SURFACE ROUGHNESS AND WEAR BEHAVIOR OF OCCLUSAL SPLINT MATERIALS PRODUCED BY DIGITAL AND CONVENTIONAL METHODS IN GASTRIC ACID ENVIRONMENT

DOCTOR OF PHILOSOPHY THESIS

NIHAN CEYLAN, DDS

ISTANBUL-2024

T.C.
YEDİTEPE UNIVERSITY
INSTITUTE OF HEALTH SCIENCES
DEPARTMENT OF PROSTHETIC DENTISTRY

IN VITRO EVALUATION OF THE SURFACE ROUGHNESS AND WEAR BEHAVIOR OF OCCLUSAL SPLINT MATERIALS PRODUCED BY DIGITAL AND CONVENTIONAL METHODS IN GASTRIC ACID ENVIRONMENT

DOCTOR OF PHILOSOPHY THESIS

NIHAN CEYLAN, DDS

SUPERVISOR
PROF. PINAR KURSOĞLU

ISTANBUL-2024

THESIS APPROVAL FORM

Institute : Yeditepe University Institute of Health Sciences

Programme : Doctorate Programme in Prosthodontics Department

Title of the Thesis : In Vitro Evaluation of the Surface Roughness and Wear Behavior of Occlusal Splint Materials Produced by Digital and Conventional Methods in Gastric Acid Environment

Owner of the Thesis : Nihan Ceylan, DDS, PhD.

Examination Date : 17.10.2024

This study have approved as a Doctorate Thesis in regard to content and quality by the Jury.

	Title, Name-Surname (Institution)
Chair of the Jury:	Prof. Dr. Ender KAZAZOĞLU (Yeditepe University)
Supervisor:	Prof. Dr. Pınar KURSOĞLU (Yeditepe University)
Member/Examiner:	Prof. Dr. Nuray ÇAPA YILDIRIM (Yeditepe University)
Member/Examiner:	Prof. Dr. Simel AYYILDIZ (Sağlık Bilimleri University, Gülhane Dentistry Faculty, Department of Prosthodontics)
Member/Examiner:	Doç. Dr. Sibel DİKİCİER (Sağlık Bilimleri University, Hamidiye Dentistry Faculty, Department of Prosthodontics)

APPROVAL

This thesis has been deemed by the jury in accordance with the relevant articles of Yeditepe University Graduate Education and Examinations Regulation and has been approved by Administrative Board of Institute with decision dated ..... and numbered .....

Prof. Dr. Bayram YILMAZ

Director of Institute of Health Science

DECLARATION

I hereby declare that this thesis is my own work and that, to the best of my knowledge and belief, it contains no material previously published or written by another person nor material which has been accepted for the award of any other degree except where due acknowledgment has been made in the text.

Nihan CEYLAN

DEDICATION

I dedicate my thesis to my beloved family who always support me and my dear nephew Kumsal CEYLAN.

ACKNOWLEDGEMENTS

I would like to thank my dear father, **Lütfü CEYLAN**, who is always by my side and watching over me. I would like to thank my dear mother **Leman CEYLAN**, whose daughter I am proud to be and whom I have idolized throughout my life. As I always say; I am so glad I am your daughter, I am so glad you are my mother. I would like to thank my dear brothers **Cihan CEYLAN** and **Ozan CEYLAN**, who listen to me no matter what the subject is, support me in every situation, and make me the luckiest sister in the world; and **Ceren CEYLAN**, the new precious member of our family. The best and luckiest feeling of my life was having a family like you.

I would like to express my sincere gratitude to **Prof. Pınar KURSOĞLU**, my very beloved mentor, teacher, and advisor. She has never stopped believing in me, supporting me to keep going forward, and showing me kindness as well as love. It has always been a great honor to be your student. As I said to you the day I wore my first white coat; you are my idol of my life.

My greatest gratitude is for my very much respected mentor, **Prof. Ender KAZAZOĞLU**, who always makes us feel his support. I want to sincerely thank him for all of his help, support, and guidance.

I would like to thank my dearest teacher **Prof. Nuray ÇAPA YILDIRIM**, whose academic success and knowledge fascinated me and from whose every word I learned something.

I would like to express my gratitude to my esteemed teacher, **Prof. Dr. Betül TUNCELLİ**, who was on both my PhD qualifying exam jury and my thesis committee, and whose sincerity I have always felt deeply.

Before starting my thesis, I attended the 4th METUM Days Dental Digital Production Congress in Ankara with my teacher, which inspired me for my thesis. I would like to thank my respected teacher **Prof. Dr. Simel AYYILDIZ**, who organized this congress and whom I am very proud to have met.

I want to express my gratitude coming from the deep of my heart to **Asst. Prof. Elifnaz ÖZEN SÜTÜVEN**, who was my assistant during my student life; my senior and my teacher during my doctorate period. Her kind perspective, support, sharing her knowledge and experience with calmness and sincerity always will be very precious for me. These years would have been very difficult without you.

My endless thanks are for all my teachers in our department, who never hesitate to share their knowledge and experiences during my undergraduate and doctorate education, for being very generous and warm-hearted.

I am deeply grateful to my beloved **Gürkan KESER**, who meticulously observed every step of this process, never spared his support, and was constantly at my side, ready to offer his help whenever needed. Without him, this process would be quite difficult.

I would like to thank all my friends who have been in my life since my undergraduate times, especially my precious Team Kursoğlu friends. I want to express my gratitude to all of my department friends for being such kind and encouraging individuals. It's always a pleasure working with them.

I would like to special thanks to **Dr. Elif Eda GÜNEYSU**, who has always been with me during this process, who encouraged me to study and who never got tired of giving motivational speeches.

I am very grateful to Optimal Dental Laboratory, especially to **Cdt. Hasan ALKAÇ** and **Cdt. Burak ÜNAL** as well as Simetri Dental Laboratory, especially to **Cdt. Serkan GÖK** and **Cdt. Soner NAĞAŞ** for their very kind help and support. I would like to thank **Cdt. Sami Can AKSOY** from Dentafab Dental Laboratoy for guiding me for the materials.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

LIST OF SYMBOLS AND ABBREVIATIONS

%	Percent
°	Degrees
°C	Degrees celcius
3D	Three Dimensional
AB	Awake bruxism
ADA	American Dental Association
AFM	Atomic force microscope
AM	Additive Manufacturing
ANOVA	Analysis of variance
ARS	Anterior repositioning splint
ASTM	American Society for Testing and Materials
BJT	Binder jetting
BMI	Body mass index
BN	Blumia Nervosa
CAD-CAM	Computer aided design and computer aided manufacturing
CBCT	Cone beam computed tomography
CLIP	Continuous liquid interface production
CLSM	Confocal laser scanning microscopy
CM	Conventional method
DED	Directed energy deposition
DLP	Digital light processing
DM	Digital method
EVA	Ethylene-vinyl acetate
FDM	Fused deposition
FDM	Fused Deposition Modeling
g	Gram
GERD	Gastroesophageal Reflux Disease
GI	Gastrointestinal
h	Hour

Hz	Hertz
LES	Lower esophageal sphincter
MEMS	Micro-electro-mechanical system
MEX	Material extrusion
MJM	Multi-Jet Modeling
MJT	Material jetting
mm	Milimetre
mm^2	Millimeter square, definition of area
mm^3	Milimeter cube, definition of volume
N	Newton
n	Population of samples
nm	Nanometer
NTI-tss	Nociceptive trigeminal inhibition tension suppression system
OAT	Oral appliance therapy
PBF	Powder-bed fusion
PC	Polycarbonate
PET-G	Polyethylene terephthalate-glycol
PMMA	Polymethyl methacrylate
Ra	2D surface roughness parameter, arithmetic mean of the surface
RMMA	Rhythmic masticatory muscle activity
rpm	Revolutions per minute
Rz	Mean of 5 roughness depths of five successive sample lengths of the profile
Sa	3D surface roughness parameter, arithmetic mean of the surface
SB	Sleep bruxism
SEM	Scanning electron microscope
SHL	Sheet lamination
SLA	Stereolithography
SM	Subtractive Manufacturing
SS	Stabilization splint
T	Time
TLESRs	Transient lower esophageal sphincter relaxations
TMD	Temporomandibular disorders

UV	Ultraviolet
V	Volume
VA	Vinyl Acetate
VPP	Vat photopolymerization
$\bar{x}$	Arithmetical mean

ABSTRACT

Ceylan, N. (2024). In Vitro Evaluation of the Surface Roughness and Wear Behavior of Occlusal Splint Materials Produced by Digital and Conventional Methods in Gastric Acid Environment. Yeditepe University, Institute of Health Sciences, Department of Prosthetic Dentistry PhD Thesis, İstanbul.

The current study's objective is comparing the mechanical characteristics such as; wear resistance and surface roughness of various splint materials that produced by digitally and conventionally exposed with gastric acid and control group as artificial saliva. Four polymethyl methacrylate (PMMA) based materials (Ceramill A Splint, Keysplint Hard, Probase Cold, Duracryl) that produced from different manufacturing methods are chosen. Cylindrical samples (n=104) with 16 mm diameter and 3 mm thickness were obtained. Samples were separated into two subgroups which are Gastroesophageal Reflux Disease (GERD) (pH:4) and control group (Artificial saliva) (pH:7). Three samples from each groups were chosen for scanning electron microscope (SEM). Samples had gastric acid exposure for 45 hours to simulate a reflux attack during sleep at night. Measurements were done for parameters as surface rougness (ΔRa) change by 3D optic profilometer and determining wear volume loss according to chewing simulator (20.000 cycles, load 100 N). Volume loss was computed after a 3D model scanner scanned each material sample.

The least amount of surface roughness modification was in heat-cured materials (1.20±2.32) with no statistical difference to the 3D printed (1.64±1.34), CAD-CAM (1.38±1.02) and autopolymerizing Resin (1.45±1.25) materials for artificial saliva (control) subgroup ($p>0.05$). However, for GERD subgroup lowest roughness was observed in CAD-CAM materials (0.57±0.45) with no statistical difference to the 3D printed (0.65±0.80) and autopolymerizing resin (0.89±0.91), wherease, there is statistical difference for heat-cured (2.63±2.64) materials ($p<0.05$). Lowest volumetric loss was in CAD-CAM materials (-0.11±0.35) with no statistical difference to the 3D printed (-0.721±0.94), heat-cured (-0.162±1.76) and autopolymerizing Resin (0.328±1.68) materials for artificial saliva (control) subgroup ($p>0.05$). However, for GERD subgroup lowest volumetric loss was seen in CAD-CAM materials (-0.201±0.57) with no statistical difference to the 3D printed (-1.032±0.83) and heat-cured (-0.391±1.15) materials, wherease, there is statistical difference for autopolymerizing resin (-2.277±2.72) ($p<0.05$).

Key words: Gastroesophageal Reflux Disease (GERD), Gastric acid, Occlusal splint, Wear resistance, Surface roughness.

ÖZET

Ceylan, N. (2024). Dijital ve Konvansiyonel Yöntemlerle Üretilmiş Oklüzal Splint Materyallerinin Gastrik Asit Ortamında Yüzey Pürüzlülüğü ve Aşınma Davranışının In Vitro Olarak İncelenmesi. Yeditepe Üniversitesi, Sağlık Bilimleri Enstitüsü, Protetik Diş Tedavisi Anabilim Dalı, Doktora Tezi, İstanbul.

Bu çalışmanın amacı, dijital ve konvansiyonel yöntemlerle üretilmiş splint materyallerinin gastrik asit ve kontrol grubu olarak yapay tükürüğe maruz bırakılmasıyla aşınma direnci ve yüzey pürüzlülüğünün değerlendirilmesidir. Dört polimetil metakrilat (PMMA) materyal (Ceramill A Splint, Keysplint Hard, Probase Cold, Duracryl) üretilmiştir. 16 mm çapında ve 3 mm kalınlığındaki örnekler (n=104) elde edilmiştir. Örnekler gastroözofageal reflü hastalığı (GÖRH) (pH:4) ve kontrol grubu (yapay tükürük) (pH:7) olmak üzere 2 alt gruba ayrılmıştır. Yüzey topografisindeki değişikliklerin değerlendirilmesi için üç örnek seçilmiştir, taramalı elektron mikroskobu (SEM) ile analiz edilmiştir. Örnekler, gece uyku sırasında reflü atağının simülasyonu için 45 saat boyunca mide asidine maruz bırakılmıştır. Yüzey pürüzlülüğü (ΔRa) değişimini belirlemek için üç boyutlu (3B) optik profilometre ve aşınma hacmi kaybının belirlenmesi için çiğneme simülatörü (20.000 siklus, 100 N yük) kullanılmıştır. Tüm örnekler 3B model tarayıcı ile taranmış ve hacim kaybı hesaplanmıştır.

En düşük yüzey pürüzlülüğü değişimi, kontrol alt grubunda, 3B (1.64±1.34), CAD-CAM (1.38±1.02) ve soğuk akrilik (1.45±1.25) materyallerine göre istatistiksel olarak anlamlı bir fark olmaksızın, sıcak akrilikte (1.20±2.32) gözlendi ($p>0,05$). Bununla birlikte, GÖRH alt grubunda, en düşük yüzey pürüzlülüğü, 3B (0.65±0.80) ve soğuk akrilik (0.89±0.91) materyallerine göre istatistiksel bir fark olmaksızın CAD-CAM materyallerinde (0.57±0.45) gözlendi, buna karşın sıcak akrilik (2.63±2.64) materyaller, için istatistiksel anlamlı fark vardı ($p<0,05$). Ayrıca, en az hacimsel kayıp, kontrol alt grubunda; 3B (-0,721±0,94), sıcak akrilik (-0,162±1,76) ve soğuk akrilik (0,328±1,68) materyallerine kıyasla istatistiksel olarak farklı olmaksızın, CAD-CAM materyalinde (-0,11±0,35 mm^3) gözlendi ($p>0,05$). Bununla birlikte, GÖRH alt grubunda, en düşük hacimsel kayıp, 3B (-1,032±0,83) ve sıcak akrilik (-0,391±1,15) materyallerine kıyasla istatistiksel olarak farklı olmaksızın CAD-CAM materyallerinde (-0,201±0,57) gözlendi; buna karşın, soğuk akrilik için istatistiksel olarak anlamlı bir fark vardı (-2,277±2,72) ($p<0,05$).

Anahtar Kelimeler: Gastroözofageal reflü hastalığı (GÖRH), Gastrik asit, Oklüzal Splint, Aşınma direnci, Yüzey pürüzlülüğü.

1. INTRODUCTION AND PURPOSE

Bruxism is defined by recurrent jaw muscle activity that includes clenching or grinding of the teeth and/or bracing or thrusting of the mandible. This condition presents in two distinct circadian patterns: nocturnal bruxism, occurring during sleep, and diurnal bruxism, occurring during wakefulness. [1].

Among the possible adverse effects of sleep bruxism (SB) that have been documented in the literature include headaches upon waking, complaints of masticatory muscle and temporomandibular disorder (TMD), severe mechanical damage to teeth, and the breaking or failing of teeth, and implants as well as periodontal disease [2,3].

The clinical implications of temporomandibular disorders (TMD) on orofacial tissues are numerous and wide-ranging. Occlusal splints are the most often used therapeutic procedures, offering intriguing clinical outcomes due to their successful reduction of TMD symptoms like minimizing sensitivity, safeguarding against tooth damage, enhancing jaw movements and mouth opening, and attaining neuromuscular equilibrium through the use of a stabilization splint [4,5]. These devices are made of hard acrylic resin that provides ideal occlusion. Thanks to splint therapy, abnormal muscle activation is decreased and "neuromuscular balance" is achieved [6]. Numerous studies have been conducted in sleep laboratories on the hypothesis that an occlusal splints may lessen the degree of nocturnal bruxism as well as the volume and intensity of muscle activity at night [7].

For the purpose of management of bruxism and TMD, splints are often made of four distinct materials which are polyamide, methyl methacrylate, polymethyl methacrylate, and urethane dimethylacrylate[8]. Most commonly used material for occlusal devices is polymethyl-methacrylate (PMMA) polymers because of their superior mechanical qualities and ease of use. While the most common method for fabricating occlusal devices is still auto-, heat-, or light-polymerized resin, recent advancements have introduced the use of computer-aided design and computer-aided manufacturing (CAD-CAM) technologies, such as subtractive (milling) or additive (3D printing), for producing occlusal splints [9,10].

Although traditionally fabricated splints may offer advantages such as ease of production and lower costs, they present several drawbacks, including polymerization shrinkage, vulnerability to fractures, dimensional instability, and extended laboratory turnaround times. Conversely, CAD-CAM systems, which gather data from both jaws

using an intraoral scanner and process it through software, provide significant advantages in terms of dimensional stability, speed, enhanced retention, and improved reproducibility. However, CAD-CAM systems also come with challenges, such as the time required to become proficient in operating the software and their higher associated costs[5].

Wear refers to removal of material from surface due to mechanical interaction, liquid or solid body movement, chemical reaction, or simultaneous chemical and mechanical activity [11]. Wear is considered to be a complex phenomenon and an "overall effect" resulting from several interconnected processes, according to the studies of wear tribology and biotribocorrosion[12].

Chewing simulators that mimic chewing motions to some extent are appropriate in vitro wear simulators. In addition to wear testing, chewing simulators can be used for various tasks, such as evaluating the application of crowns and bridges to determine risk of cracks and fractures, or examining alteration in marginal integrity of restorations[13].

Gastroesophageal reflux disease (GERD), is a digestive dysfunction characterized by seepage of gastric and duodenal contents from stomach into the esophagus [14].

According to literature, it is shown that there is strong relationship between GERD and bruxism. Anxiety and depression suggest that bruxism could be a result of a dysregulation of the brain's and the gastrointestinal tract's bidirectional communication. Patients with severe bruxism frequently experience GERD symptoms for long periods of time [15].

According to Mengatto et al.[14], association between SB and GERD was evaluated and the study show participants with GERD had a greater prevalence of SB (73.7%) than did those in good health (23.1%). When bruxism and GERD are evaluated together, treatment of the condition and material selection become even more important.

Studies carried out on adults in good physical condition indicate that oesophageal acidification elevates occurrences of bruxism during sleep, as well as arousals and swallowing during sleep[16].

Gastroesophageal reflux disease (GERD) leads to splint materials' properties change because of exposure to gastric acid in patients [17]. Furthermore, bruxism or parafunctional activities result in the gradual wear of these materials, even when the splint is used for a relatively short duration[4].

The accumulation of plaque and staining agents, as well as the color stability and glossiness of the resins[18,19], are all significantly influenced by the surface roughness.

Roughening caused by intrinsic and extrinsic chemicals can change the surface properties enduring the contact time as well as material and fabrication type [20].

This current research's goal is to compare wear resistance and surface roughness of various splint materials that produced from digitally and conventionally materials and exposed to gastric acid and saliva.

1. First null hypothesis is that no statistical differences in terms of surface roughness change among materials produced by different techniques in gastric acid environment.
2. Second null hypothesis is that no statistical differences in terms of volume change among materials produced by different techniques in gastric acid environment.
3. The third null hypothesis is that no statistical differences in terms of roughness change in occlusal splint materials for gastric acid and artificial saliva environment after chewing simulator.
4. The forth null hypothesis is that no statistical differences among occlusal splint materials in terms of volume change in gastric acid and artificial saliva environment after chewing simulator.
5. The fifth null hypothesis is that no statistical differences in correlation between surface roughness and wear resistance.

2. GENERAL INFORMATIONS

2.1. Bruxism

Bruxism is characterized as a repetitive jaw-muscle movement involving the clenching or grinding of teeth and/or the bracing or thrusting of the mandible. This condition presents in two distinct circadian patterns: it can occur during sleep, referred to as sleep bruxism (SB) or nocturnal bruxism, or during wakefulness, known as awake bruxism (AB) or diurnal bruxism.[21,22].' Diurnal bruxism is characterized by masticatory muscle activity involving during daylight hours, often accompanied by recurring contact between the teeth. Conversely, SB is identified as masticatory muscle activity that occurs during sleep, which may or may not be rhythmic in nature. In healthy individuals, awake bruxism is not categorized as a movement disorder[23].

Patients diagnosed with this disorder are frequently referred to as bruxists. The condition is primarily characterized by teeth grinding and clenching, regardless of its definition, cause, or type of expression [22].

2.1.1. Sleep Bruxism Etiology

Biological, psychological also lifestyle are all part of the multifactorial of SB. Studies on gene analysis and family member observations suggest that both genetic and environmental variables may contribute to the occurrence of SB. Additionally, the onset of rhythmic masticatory muscle activity (RMMA) and sleep bruxism (SB) may be impacted by an imbalance in centrally active neurotransmitters, such as serotonin and dopamine. It has been proposed that a number of psychosocial factors, including stress, anxiety, depression, and maladaptive coping mechanisms, raise the risk of SB. It's also been proposed that certain lifestyle choices, such as drinking alcohol and smoking, raise the risk of SB [2].

2.1.2. Sleep Bruxism Adverse Effects

Among the possible adverse effects of SB that have been documented in the literature include headaches upon waking, complaints of masticatory muscle and temporomandibular disorder (TMD), tooth wear, impairment of teeth, restorations, and implants as well as periodontal disease [2,3].

2.2. Temporomandibular Disorder (TMD)

The term TMD refers to diverse range of musculoskeletal and neuromuscular problems that affect the TMJ as well as the surrounding osseous and musculature. Up to 15% of adults suffer from TMD, which peaks between the ages of 20 and 40. There are two types of TMD: extraarticular and intraarticular. Earaches, headaches, facial pain, and jaw pain or dysfunction are typical symptoms. There are numerous contributing factors to the multifactorial etiology of TMD, such as environmental, social, emotional, and cognitive causes. Most typically, a physical examination and history are used to make a diagnosis [24].

2.2.1. Temporomandibular Disorder (TMD) Management

Many management strategies have been used to lessen TMD's negative effects. Oral appliance therapy (OAT) is a common and well-known method; in particular, the stabilizing appliance is gold standart process and utilized in clinical procedures. Other treatment protocols were also used such as conservative procedures (counselling, biofeedback, physical therapy, pharmacotherapy) as well as irreversible and more invasive procedures, such as occlusal adjustments, orthodontics, arthroscopy, and surgical interventions[3,25,26].

2.2.1.1. Occlusal Splint Therapy

An occlusal splint, often termed an orthopaedic or bite splint, is a versatile and detachable intraoral device utilized for therapeutic applications and for assessing a newly established vertical dimension before undertaking comprehensive full mouth rehabilitation. Although it is effective in lessening pain intensity, enhancing mandibular kinetics, the specific cause of these outcomes is yet unknown[26].

The functionality of an occlusal splint is dictated by its design, which may vary from a partial coverage apparatus to a flat plane that encompasses every occlusal surfaces. Alternatively, it can be more intricate, incorporating occlusal ramps or indentations, contingent on the clinical indications[27].

The literature has reports on a range of occlusal splints used to treat TMDs and commonly used splints are stabilization splints[25].

Non-occluding splints, hard and soft stabilization splint, mini anterior splints like nociceptive trigeminal inhibition tension suppression system (NTI-tss), anterior repositioning splint are used from past to present [25,28].

2.2.1.1.A. Stabilization Splint (SS)

Stabilization splints (SS) are one specific kind of occlusal splint. This splint is constructed from hard acrylic material, serves to provide a temporary and removable optimal occlusion. By facilitating optimal occlusion through splint therapy, abnormal muscle activation is reduced, thereby achieving "neuromuscular equilibrium. As the splint encourages the lower jaw to adopt a new position, it necessitates adjustments over multiple visits to accommodate the relaxation of the masticatory muscles, thereby achieving a stable jaw alignment. These adjustments can be made by selectively grinding certain points on the splint's surface. Patients should be scheduled for regular follow-up appointments. For a long time, the stabilization appliance has been regarded gold standard in the management of TMD pain [6,29].

Different results were obtained in studies comparing the treatment effectiveness of soft and hard stabilization splints. It was found that hard splint reduces muscle activity more than soft splint; one study also reported that soft splints increased muscle activity [29]. It is determined that low-cost soft splints, when used in conjunction with self treatment, did not offer significant additional benefits compared to self-care treatment alone. While promoting short-term appliance therapy use is the primary objective, it's crucial to assess the long-term results as well because TMD is frequently a chronic, recurrent pain problem [30].

2.2.1.1.B. Anterior Repositioning Splint (ARS)

ARS is a widespread conservative method is used for treatment of disc displacement-related TMD. An anterior guidance ramp is typically used to maintain the protrusion status of an ARS, that can be fixed on the maxilla or mandible. The mandibular protrusion modifies the disc condyle relationship and is a commonly employed therapeutic modality for intra-articular TMD. Pain symptoms can be significantly decreased with using this technique. As such, it can also be applied to TMDs that cause discomfort. Upon the completion of therapy, the splint is progressively reduced. This allows the mandible to revert to its original alignment, thereby restoring the disc-condyle complex to its proper location. Most experts advocate for a continuous wear period of 24 hours daily, spanning 3 to 6 months[31].

2.2.1.1.C. Non-occluding Splints

As the name suggests, nonoccluding OAs are unable to immediately change vertical dimension or condylar posture because they lack an occlusal platform that makes contact with the opposing dentition [32].

For both myogenous and arthrogenous TMDs, this appliances performed better than the control. According to those research, a positive effect and the lingual flange, which arrange tongue position could be the cause of the outcome favoring non-occluding splints. It is impossible to ignore this beneficial placebo effect, especially in those with myogenous TMDs. In conclusion, there was insufficient data to support the claim that non-occluding splints can effectively lessen discomfort in people with arthrogenous TMDs [29,30]. Research shows the electromyographic (EMG) activity of the masseter muscle in sleep bruxism (SB) patients has demonstrated that both types of oral appliances (OAs), whether occluding or nonoccluding, were equally effective in temporarily mitigating nocturnal muscle activity in some individuals. [32].

2.2.1.2. Materials used in occlusal splint construction and their historical development

Several metal materials have been used to fabricate occlusal splints in the past, such as gold, silver, or even lead. Today, most splints are made using heat- or light-curing acrylic [33]. On the other hand, different materials are utilized for TMD treatment, i.e., polymethyl methacrylate (PMMA), polyethylene terephthalate-glycol (PET-G), ethylene-vinyl acetate (EVA), polyolefin, polyetheretherketone (PEEK), polycarbonate (PC) [26].

2.2.1.2.A. Polymethyl Methacrylate (PMMA)

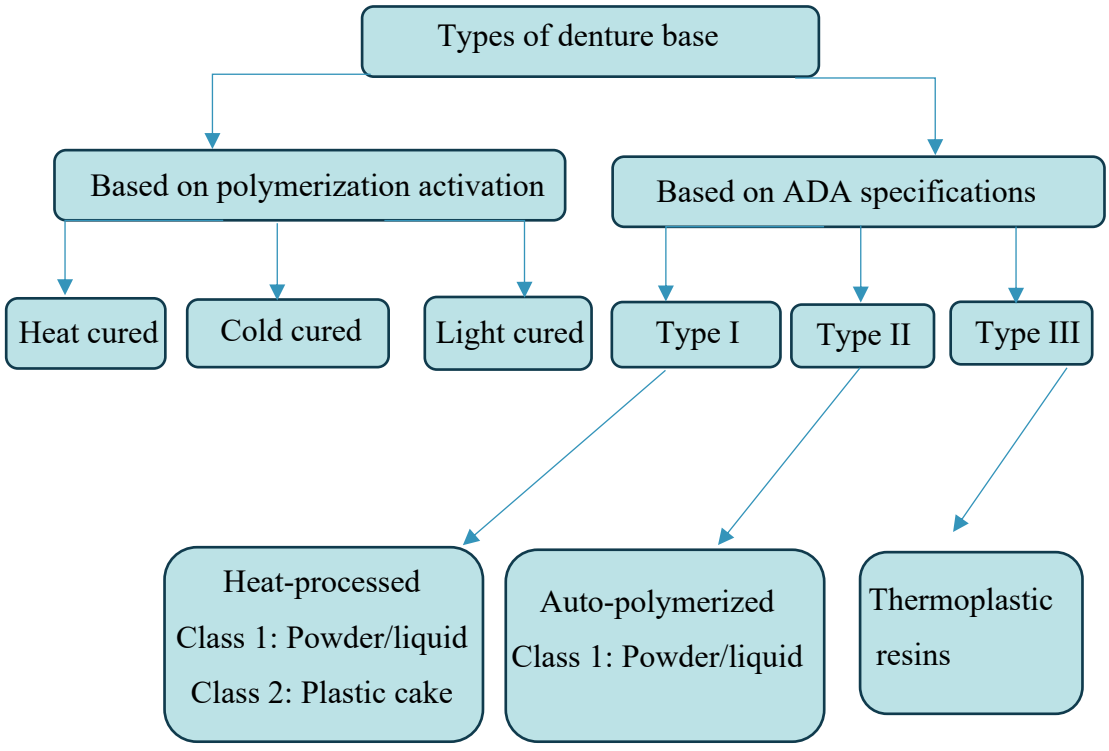
PMMA was initially described by Redtenbacher in 1843 [34]. In clinical dentistry, polymers of all kinds are frequently employed for a variety of purposes. Among them, dental and clinics like relining of dentures and temporary crowns, also industry (for the creation of artificial teeth) are the places where polymethyl methacrylate, or PMMA, is most frequently utilized. PMMA is typically offered as a powder–liquid combination, regardless of the intended use. The powder is made of a transparent polymer (PMMA), but to replicate the appearance and physical characteristics of oral tissues (such as the mucosa and gums). A methyl methacrylate monomer, inhibitors, and cross-linking agents are present in the liquid component. Due to PMMAs special qualities—such as its low

density, attractiveness, affordability, simplicity of manipulation, and mechanical and physical propertie—it has become more and more widespreas for dental applications. [26,35].

- **Chemistry and Types of PMMA**

Poly(methyl methacrylate) (PMMA), also known as poly[1-(methoxy carbonyl)-1-methyl ethylene], is a synthetic polymer formed by polymerizing methyl methacrylate (C5O2H8) with the involvement of free radicals, producing poly(methyl methacrylate) (C5O2H8). According to the specifications set forth by the American Dental Association (ADA) in Specification No. 12, denture base polymers are categorized into distinct types and classes. These denture base polymers are primarily classified into three main categories based on the activation mechanisms of the polymerization reaction, each differing in their polymerization reactions and compositions (see Table 2.1). Furthermore, beyond types I, II, and III as defined by ADA specifications, the ISO 20795-1:2013 standard has introduced additional categories, specifically light-activated (type IV) and microwave-cured (type V) polymers. [36].

Table 2.1. Classification of denture base polymers based on polymerization activation and according to the ADA specifications.



- **Heat-cured PMMA**

Curing with heat PMMA materials are widely utilized to make denture bases and dentures [37]. They are available in liquid and powder form. After the two components are mixed, the polymerization reaction begins, and the initiator needs heat energy (from a water bath, for example) to be activated. Benzoyl peroxide is used as an initiator in these materials; upon heating, it separates into carbon dioxide (CO_2) and produces free radicals [38]. Good physical qualities are the consequence of high polymerization; but, polymerization and inadequate adaptation are still the primary issues [37]. In contrast to rapid-heat-polymerized PMMA, which has a much shorter curing time (20 min at 100 °C), the conventional heat curing cycle requires a longer curing period (~9 h at 74 °C) [39].

- **Cold-cured PMMA**

Cold-cured PMMA does not need thermal energy and has a different composition and polymerization mechanism[37]. The cold-cured PMMA is combined with a tertiary amine initiator, such as dimethyl-p-toluidine, to activate the benzyl peroxide and start the polymerization process by chemically producing free radicals[38]. However, compared to cold-cured PMMA has a noticeably lower degree of polymerization, leaving leftover uncured monomers in the polymerized substance that have a tendency to leach away from them[40]. The primary benefits of cold-cured PMMA over heat-cured PMMA are its improved dimensional stability and adaptability, which leads to minor polymerization shrinkage but a lesser degree of polymerization[38]. The oxidation of the amine initiator over time results in discoloration, reduced color stability, and a decrease in the glass transition temperature. These effects are accompanied by monomer leaching and suboptimal mechanical properties. Currently, only applications for cold-cured PMMA materials are denture repair and manufacturing custom trays and temporary partial dentures [37,38,40,41].

- **Light-cured PMMA**

Visible-light-cured, or light-cured, PMMA functions similarly to composites made of resin that cure when exposed to visible light [42,43]. Conventional initiator in PMMA has been switched out for a photo-sensitive substance called camphorquinone, which when exposed to light becomes active and releases free radicals. The pre-mixed form of the light-cured PMMA materials comes with silica, acrylic resin monomers, urethane dimethacrylate matrix, and PMMA fillers. PMMA needs to be exposed to visible

light for the necessary amount of time after the cast is adjusted and the teeth are positioned in order to cure fully. After polymerization, light-cured PMMA can be polished and finished in a way similar to that of heat-cured PMMA[40].

The benefits of light-cured PMMA include easier manufacture and complete control over the curing process, giving plenty of opportunity for modification and adaption prior to starting the polymerization process [44]. The possible advantages of light-cured PMMA also include decreased bacterial adherence, polymerization shrinkage, and residual monomer presence in comparison to heat- and cold-cured PMMA materials[44,45]. However, due to its limitations, including their high cost, sensitivity to technique, and restricted curing depth, light-cured PMMA materials are not commonly used. Because light-cured PMMA materials' mechanical qualities are marginally worse than those of conventional PMMA, their uses are restricted to denture base relining and repair, as well as the creation of customized base plates for dentures [35].

- **Microwave-cured PMMA**

For polymerizing, a non-metallic denture flask and a microwave energy source are needed in place of a traditional water bath [44]. The main benefit is that it saves a lot of time because it only requires three minutes at 500 watts in a microwave, as opposed to hours of heating followed by a cooling phase in a standard heat curing cycle[39,46]. The physical characteristics of microwave-cured PMMA materials are similar to those of traditional heat-cured PMMA, including dimensional accuracy. According to a findings, injection-molded, microwave-cured PMMA was shown to have lower impact and flexural strengths than conventional heat-cured PMMA [46]. The primary drawback of the PMMA materials that are microwave-cured is their weak adhesion to acrylic teeth, which restricts their use in prosthodontics [44].

2.2.1.2.B. Polyethylene terephthalate-glycol (PET-G)

Thermoplastic PET-G polymer is found in thermoforming sheets and is utilized in the production of aligners, orthodontic retainers, and occlusal splints. Though they can be colored, the sheets are usually translucent [47]. A thermoforming unit composed of an infrared heat source and a vacuum reserve is needed to create this kind of equipment. A combined occludator (Occluform 3 by Erkodent) is included with some of these devices, enabling the opposite jaw to be positioned as desired. Despite being advertised as appropriate for the creation of occlusal splints, there is a few information available

regarding the use of PET-G in this context, and the majority of research focuses on its orthodontic purposes. For instance, it has been shown that thicker PET-G aligners provide better stress distribution, extending the appliance's lifespan. Furthermore, it has been hypothesized that using multi-layer vacuum formed sheets can result in retainers that are more durable [26,48].

2.2.1.2.C. Ethylene-vinyl acetate (EVA)

Heat facilitates the easy manipulation of EVA, or poly (vinyl acetate)-polyethylene, a thermoplastic flexible polymer. Furthermore, this is widely accessible on the market and thought to be non-toxic[49,50]. On the other hand, it can be also utilized to fabricate soft occlusal splints [32]. Vinyl Acetate (VA) ratio is one of the elements that affects its mechanical qualities. A material that has a greater VA ratio is harder, more transparent, and more flexible. Furthermore, dark colors fit and retain better than translucent ones. The most common figure for sports mouthguards is a VA ratio of 28% [50]. The use of EVA as mouthguards has raised some issues, though, mostly because of its high water sorption, which causes it to expand in volume when it comes into contact with the oral cavity and jeopardize athletes' comfort [26].

2.2.1.2.D. Polyolefin

Polyolefin has better tensile and tear strengths, less water sorption, and a similar shock-absorbing effect than EVA. Comparative clinical research is still necessary, nevertheless, to establish which material is better in terms of mechanical qualities, athlete comfort, and performance. A study has shown that using pressure-forming rather than vacuum thermopressing results in a better fit for the polyolefin mouthguard [26,51].

2.2.1.2.E. Polyetheretherketone (PEEK)

PEEK is a popular subject for new material research because of its favorable biocompatibility, high temperature resistance, superior fatigue properties, high toughness, resistance to aging and corrosion, ease of processing, and color stability[52]. It is opaque and has a natural tooth color (brown or grey), although metal, fiber, or ceramic particles can change its mechanical characteristics and color [53]. Furthermore, PEEK continues to be the preferred material to manufacture digital occlusal splints, while further research is required to completely comprehend additive manufacturing techniques [54]. PEEK needs a certain polishing process, and the best one hasn't been found yet. Poorly

polished PEEK can accelerate wear on adjacent surfaces and absorb stains that will eventually contrast with the opaque portion of the splint[53].

2.2.1.2.F. Polycarbonate (PC)

It exhibits less polymerization shrinkage than conventional PMMA, which leads to a better fit. Due to its increased flexibility, the splint's fracture resistance can be maintained even when its thickness is reduced to 0.3 mm. Patients tend to adjust to thinner splints more quickly, so this could lead to improved patient compliance. Furthermore, a thinner layer enables the application of "double splints," one for each arch, which improves testing accuracy for a recently determined vertical dimension of occlusion [55].

Comparing PC splints to PMMA materials either digitally (additive and subtractive) or conventionally, an in vitro investigation found a lower surface hardness and more surface wear. These findings restrict the usege of polycarbonate splints and render them ineffective for individuals undergoing significant masticatory stresses or who are bruxers. The expensive price of PCs is another drawback [55,56].

2.3. Computer Aided Design-Computer Aided Manufacturing (CAD/CAM)

In dentistry, usage of CAD/CAM, has grown over the past few years. It has been utilized for decades in the industry for everything from casts and impression production to temporary fabrication and final restorations [41,57]. CAD/CAM systems are made up of a scanner, processing software, manufacturing devices which turns data into an original appliance, denture, or restoration. This "digital workflow," which captures both dentitions, enables dental expert to evaluate the preparation of the tooth and build a restoration that adheres to treatment. To facilitate fast contact with the technician and enable any necessary adjustments before moving on to the next phase, a digital file can be uploaded to a cloud server. Most of time, the procedure saves time and requires no impression of the materials.

There are numerous scanning technologies on the market at present. For using an oxide powder was necessary to improve the scan's quality. In order to record of tooth preparation, scanning is processed using either a stream of video images or photos. Each system has proprietary designing software that enables the doctor or technician to plan the appliance or restoration in respect to the opposing dentition. After processing, data is produced chairside, in lab, or in central production facility [58]. Manufacturing process may be subtractive or additive [59,60].

2.3.1. CAD/CAM Advantages and Disadvantages

When it involves dental restorations, CAD/CAM technology offers a lot of benefits over conventional methods. These benefits include quality, quickness, and convenience of use. Compared with conventional impressions, digital scans may be quicker and simpler because they do not require casting, wax-ups, investing, casting, or firing. On the same day that they arrive, patients can have their permanent restoration, saving them trouble of scheduling a second appointment, thanks to the on-site milling equipment. Provisional restorations are not necessary because they take time to fabricate and fit [61]. Costs might be decreased by saving time and labor. Patients ought to be attracted to the possibility of obtaining faster, better restorations, and they should be relieved that no gag-inducing impressions are required. Another benefit is that all of the scans may be stored on the computer; traditional stone models need to be stored and can fracture or break if they are not properly conserved[62].

CAD/CAM systems are not without drawbacks, though. The equipment and software have a high initial cost, and the practitioner must invest time and money in training[61]. Dentists who do not perform a sufficient number of restorations may have a difficult time making their investment pay off. The dentist must get an exact recording of the teeth when taking an optical scan, much like with conventional impressions. Since digital impression systems require numerous processes, it's possible that they don't save as much time as they do. For instance, dentists using specific scanners are required to send the images for first, and then a dental technician sets the margins. After being reviewed in the dentist's laboratory, the photos are returned for model milling. Finally, the dentist's laboratory receives the models and dies so that the restoration may be made [63].

2.3.2. Subtractive Manufacturing (SM)

Use milling, drilling, or grinding techniques in subtractive manufacturing that are predicated on removing material or cutting solid blocks or bars[64].

Kanazawa et al.[65] tried to enhance and shorten CAD/CAM denture manufacture procedure by utilizing subtractive manufacturing. In order collect data regarding the patient's mucosa and centric relation, they performed cone beam computed tomography (CBCT) scans on a set of prosthetic teeth as well as the patient's previous complete denture. Three-dimensional CAD software was used to create the virtual denture [66].

In 2012, Goodacre et al. [67] presented the first clinical report of a patient treated with proof-of-concept CAD/CAM dentures fabricated of transparent plastic using a three-axis milling machine. At the milling technique, denture teeth were fixed manually. Authors also provided a second clinical case in which a 5 axis milling machine CAD/CAM dentures from prepolymerized PMMA. Once more, the teeth were manually attached and after that, the patient's final dentures were inserted and remain in clinical use ever since [67].

When making occlusal splints, the digital method has the primary benefit of producing the devices faster and more accurately than when they are constructed by hand.

This method's primary drawback for occlusal splint is that it wastes about 70% of the material because it can only mill 1-2 splints from a single blank [68].

2.3.3. Additive Manufacturing (AM)

Using three-dimensional models as a basis, items are manufactured layer by layer through AM procedures [69].

3D printing is a term that is commonly used to refer to all additive procedures. AM is a relatively recent method of preventing material waste. Despite being first presented by Charles W. Hull in 1986, dental industry took some time to accept AM, also known as three-dimensional (3D) printing. Nonetheless, in the years that followed, technology developed quickly. After the fused deposition modeling (FDM) technique patent expired in 2009, 3D printers started to gain significant traction in the consumer market. In the end, this dynamic infiltrated the dental industry. Printing units got cheaper and smaller, and their uses shifted. The materials that can be printed on have grown to include ceramics, metal, polymers, and even human tissue. There are three categories of materials used in fast prototyping processes: powder, polymers, and metals[70].

Selective laser melting or sintering has been used to successfully construct a number of additive dentistry processes, such as the manufacturing of crowns, fixed partial dentures, and partial denture frames. In addition to printing metal powders, UV light layering techniques such as stereolithography (SLA), PolyJet, and digital light processing (DLP) can also be used to produce polymers and polymerize polymeric resins. These methods enable the production of molds for the vacuum thermoforming of occlusal devices. For the manufacturing of translucent surgery models, implant drilling guides, and occlusal devices, this is particularly fascinating. Accurate casts can be produced when combined with computed tomography or cone beam computed tomography data[68].

Dental 3D printing follows this hype cycle **(Figure 2.1.)**. In 2014, a Gartner analysis estimated that it will take 10 to 15 years for 3D printing to become widely used. For the dentistry industry, this may be roughly true[70].

Based on machine processes, AM technology has been classified by the American Society for Testing and Materials (ASTM) Committee F42[71,72]. Moreover, AM procedures can be categorized based on the material's state, which can be further split into four groups: (1) liquid, (2) paste/filament, (3) powder, and (4) solid sheet[73].

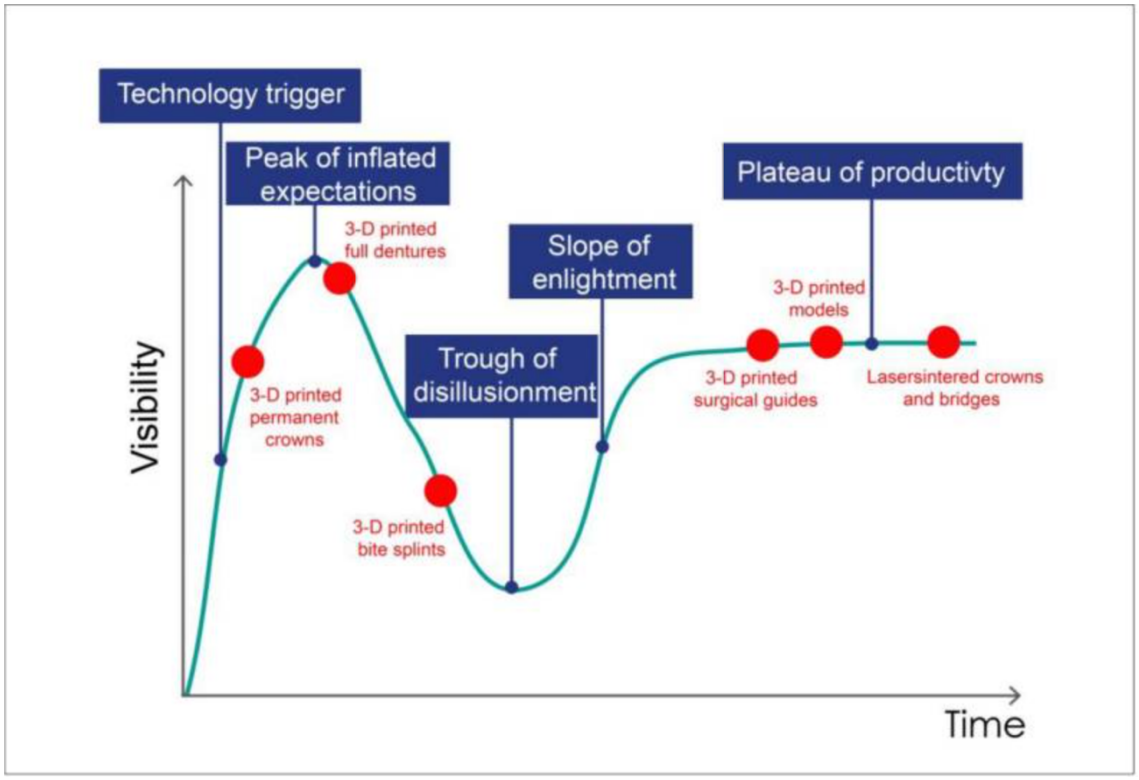


Figure 2.1. Dental 3D printing follows the characteristics of the Gartner hype cycle.

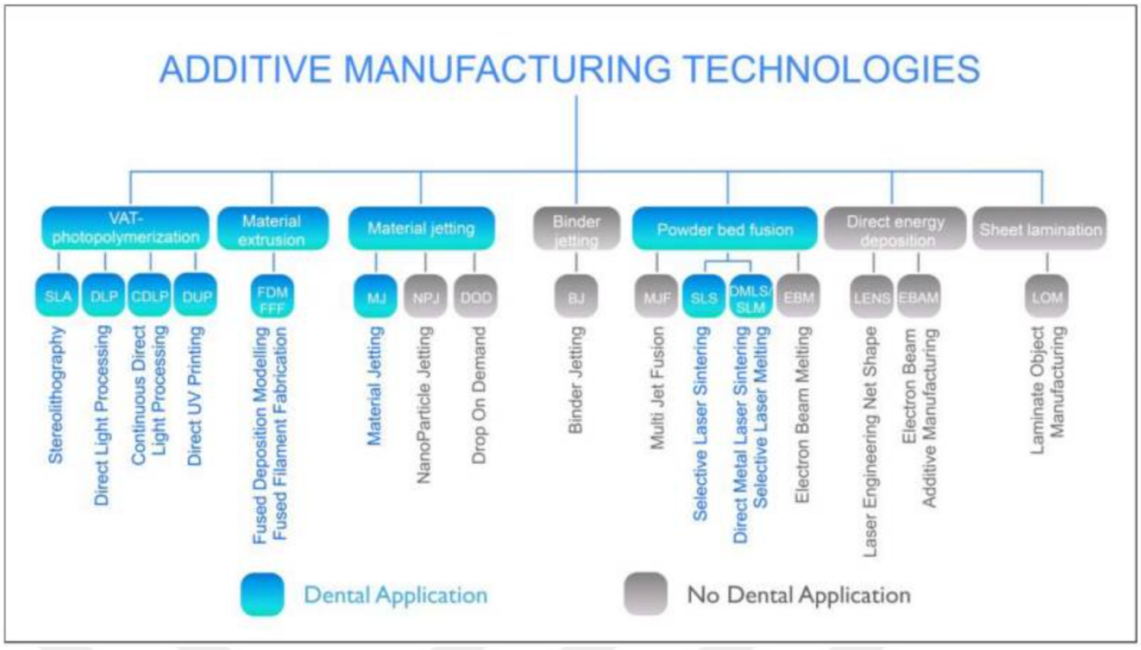


Figure 2.2. Overview of the existing process categories in additive manufacturing. (According to EN ISO 17296-2).

Within additive manufacturing, the following seven technique classes are distinguishable (Figure 2.2) [70,72,74].

2.3.3.1.Vat photopolymerization stereolithography (VPP)

By "laser vat-polymerizing" successive layers of liquid resin, this method produces objects. A vat containing liquid photopolymer is exposed to the UV laser, which selectively polymerizes the resin for every cross-section [75,76]. Because of its excellent chemical bonding between the layers, flawless surface finish, and highest levels of accuracy, it was the first commercially viable technology to be presented to the market [77,78]. Furthermore, the size and quantity of parts produced affect its speed. Since small parts are orientated next to one another on the same build platform, many small part scans can be created simultaneously. It could take one or two minutes to develop a layer, and it could take six to twelve hours to grow up completely. Larger items may require many days to process [41].

- **Stereolithography (SLA)**

Three components make up the stereolithography (SLA) process: an ultraviolet (UV) laser, a model-building platform, and a liquid resin vat. Its foundation is the "top-

down" methodology. As the first layer, as depicted in the CAD model, fully polymerizes and covers the object in a single layer of new resin, the cross section of the object is removed using a blade filled with resin. In order to polymerize the subsequent layer, the structure of the platform has been reduced in the z-direction. Until thing is entirely constructed, these steps are repeated [75,76].

Other devices use a modified "bottom up" SLA method, where the build platform is elevated along the z-direction and dipped from above into the resin vat while being built. With this method, less resin is needed, and the printed product has better control over the thickness of each layer, improved manufacturing accuracy, and fewer porosities [79]. When the creation is finished, the excess resin is drained and reused. The object is post-cured in a UV oven after being sterilized in an alcohol bath to strengthen it and polymerize the unreacted resin groups[80].

- **Digital Light Processing (DLP)**

It is part of vat-polymerization technology since both utilize liquid photopolymers as the building material and light as the polymerizing agent to print accurate, small things with complex shapes [81].

A transparent vat containing the photopolymer is placed above a projector that makes use of a powerful LED source. A digital mirror device (DMD) is a component of the micro-electro-mechanical system (MEMS) that powers the projector. The build platform (x-y axis) is dipped one layer thickness into the resin vat during the printing process. The DMD reflects UV light onto the resin vat's surface, where it produces a layer in accordance with the CAD design[72]. DMDs are composed of separately movable micro-mirrors that regulate the UV light's path of reflection. The construction platform moves higher (z-axis) by one layer thickness when a layer is built. The procedure is repeated until the object is completed. Compared to the SLA method, DLP produces items faster, with smoother surfaces, better precision, and at a lower cost. However, this approach results in parts that are impacted by polymerization shrinkage, necessitating a manual post-processing procedure [82,83].

- **Continuous liquid interface production (CLIP)**

Recently found vat-polymerization method called continuous liquid interface production (CLIP) utilizes oxygen and UV radiation to continually create three-

dimensional objects. Photopolymers are used in the CLIP process, just like in the SLA and DLP procedures. It is made up of an oxygen-permeable window, a UV light, a top-down build platform, a liquid resin vat, and a screen that produces light underneath the window which is made of a thin, amorphous Teflon film that is highly resistant to chemicals and has superior control over UV translucency and oxygen permeability. The extremely thin liquid resin interface layer that is located above the window does not polymerize once light enters it. This thin layer, known as the "dead zone," is made of oxygen-rich resin. The first step in 3D printing procedure is to project continuous series of 2D cross-sectional pictures of a 3D CAD model into the resin using light. The extremely thin liquid resin interface layer that is located above the window does not polymerize once light enters it. This thin layer, known as the "dead zone," is made of oxygen-rich resin. When oxygen levels in the resin above dead zone drop, the resin begins to polymerize, creating a solid three-dimensional object in shape of the projected two-dimensional cross-sections. The 3D item is drawn upward by the construction platform from liquid resin vat as it forms. The procedure is kept up until shape is constructed. CLIP method creates products out of biological material, ceramics, and soft elastics, it can be integrated into a wide range of scientific and technological fields [84].

2.3.3.2. Material Extrusion

- **Fused Deposition Modeling (FDM)**

The process of FDM involves the extrusion of semi-molten thermoplastic material via a nozzle to build things. Metal or plastic wire filaments are taken off of a spool and forced into a heated nozzle that is manipulated in both horizontal (x/y) and vertical (z-axis) directions by a numerically controlled system. Certain FDM printers may also allow multiple print heads, which facilitates the co-printing of support systems for geometrically difficult items and allows for the printing of various colors on the same object. The nozzle tip drops additional layers that fuse together when the build platform is lowered after the initial layer has created. Until the thing is completed, this process is repeated.

This procedure uses reliable and inexpensive products for the environment. Additionally, it can create colorful objects out of different materials without combining them. The exceptional stability of the objects produced by this procedure eliminates the

need for a chemical post-processing step. One drawback of the generated things is their low surface quality, which can be attributed to the thickness of the filament. To smooth the surface, finishing is necessary, and building large, complex structures might take several days [80,85].

2.3.3.3. Material Jetting or Multi-Jet Modeling (MJM)

This method 3D prints an object using inkjet technologies using several nozzles. Multi-Jet Modeling (MJM) generates high-resolution objects with intricate geometry. The printer software combines the cartilages made of thermo-polymer material in this procedure. Following that, as they travel down the x/y axis, the inkjet nozzles jet photopolymer, which is going to be cured by a UV lamp. Before exposing the next layer to more UV radiation, a leveling blade is used to ensure that the layers have the same thickness. The part is then made totally by building a second layer on top of first layer after build platform has been decreased by one layer thickness along the z-axis. Through this method, layer thicknesses less than 20μm can be obtained. As a support structure, a gel-type monomer or wax is injected along with the part material during the 3D printing process to prevent structural collapse. After 3D printing process is complete, this support can be simply removed by heating or water washing out. MJM is affordable, efficient in constructing parts, office-friendly, able to blend two materials with contrasting colors, and produces very accurate things. Nevertheless, the strength of items made with MJM is reduced [85–87].

2.3.3.4. Powder Bed Fusion (PBF)

- **Selective Laser Sintering (SLS)**

Using carbon dioxide (CO_2) laser to fuse or sinter successive layers of powdered material together, SLS is a powder bed fusion process that creates items. Powder is heated slightly below the material's melting point prior to sintering, and a laser is used to trace cross-sectional slices from the digitally created CAD model to selectively fuse powdered materials. The powder is distributed throughout the surface of a construction chamber by a roller once powder particles are lying loosely in a bed. Everytime a layer is completed and adheres to the one before it, the manufacturing piston is lowered by one layer thickness. When the object is finished, these processes are repeated. The sintered powder

does not melt and no support structures are required because the unbound, loose powder particles support the 3D-printed result. Powder residues can be recycled. A post-processing operation, such as high-pressure air sanding or pressured air cleaning, is necessary after the product is constructed. This method uses a range of inexpensive, safe for the environment materials to create fully thick, very accurate objects. Additionally, the finished products are rough and porous [86,88].

2.4. Gastroesophageal Reflux Disease (GERD)

A medical condition known as GERD occurs by reflux of stomach contents, which can lead to uncomfortable symptoms and/or consequences [89]. Around the world, GERD is a highly prevalent digestive disease. It has been connected to significant morbidity and a lower quality of life [90].

2.4.1. Epidemiology and Pathophysiology of GERD

GERD risk factors include smoking, a high body mass index (BMI), getting older, experiencing anxiety or despair, and engaging in less physical exercise at work. The size and timing of meals, especially in relationship between sleep, as well as the acidity of the food can all affect the signs of GERD [91,92]. Compared to male patients, patients with non-erosive reflux disease are more likely to be female and experience GERD symptoms.

Although there may be other contributing variables, the lower esophageal sphincter's (LES) dysfunction is the main cause of gastroesophageal reflux. Both pathologic and physiological variables play a role in pathophysiology of GERD. Transient relaxations of lower esophageal sphincter (TLESRs) are the most frequent cause. TLESRs are discrete, non-swallowing intervals of LES tone inhibition. Even though they are physiological, their primary cause of acid reflux in people with GERD also become more prevalent in the postprandial period. Concerns should also be raised by hiatal hernia, lower LES pressure, delayed stomach emptying, and impaired esophageal clearance[93].

2.4.2. Complications of GERD

Barrett's esophagus and esophagitis are two serious complications of GERD. The severity of esophagitis ranges greatly; severe cases cause the esophagus to narrow,

ulcerate, and undergo significant dental erosions [94]. Gastrointestinal (GI) bleeding is another possible outcome of esophagitis. Anemia, hematemesis, coffee-ground emesis, melena, and, in certain cases, hematochezia are symptoms that can indicate upper gastrointestinal bleeding. Prolonged acid exposure can cause persistent inflammation of the esophagus, which can result in scarring and the formation of peptic strictures. Peptic strictures typically manifest as dysphagia [95].

2.4.3. Management of GERD

Numerous medications and surgical procedures, such as proton pump inhibitors, alginates, antacids, histamine 2 receptor antagonists, and surgical fundoplication, are available to treat GERD. However, because of their expense, potential side effects, and/or problems, these treatments come with a patient burden. Furthermore, even with the best medical care, 20–45% of GERD patients still have persistent symptoms. As a result, individuals with GERD frequently try alternative lifestyle treatments, which medical practitioners advocate as supplementary or alternative methods of managing GERD symptoms. Lifestyle treatment involves making adjustments to sleeping arrangements, clothing fit, exercise routines, and dietary habits[96].

2.4.4. GERD pH

The most sensitive method for determining the features, course, and resolution of all forms of gastroesophageal reflux, including acidic, weakly acidic, and weakly alkaline reflux, is esophageal impedance monitoring. To make use of the impedance pH monitoring results, a precise and well-organized analysis is needed.

2.4.4.1. Esophageal pH detected by impedance during reflux

Based on the esophageal pH as measured by impedance during reflux, there are four subgroups **(Figure 2.3.)** [97,98].

i. acid reflux, a decrease in pH to < 4

ii. superimposed acid reflux (acid re-reflux), reflux while the pH is < 4 during an acid clearing interval (before the esophageal pH has recovered to > 4 after acid reflux)

iii. weakly acid reflux, the pH nadir is > 4 but < 7 during reflux

iv. weakly alkaline reflux, pH remains > 7 or increases to > 7.

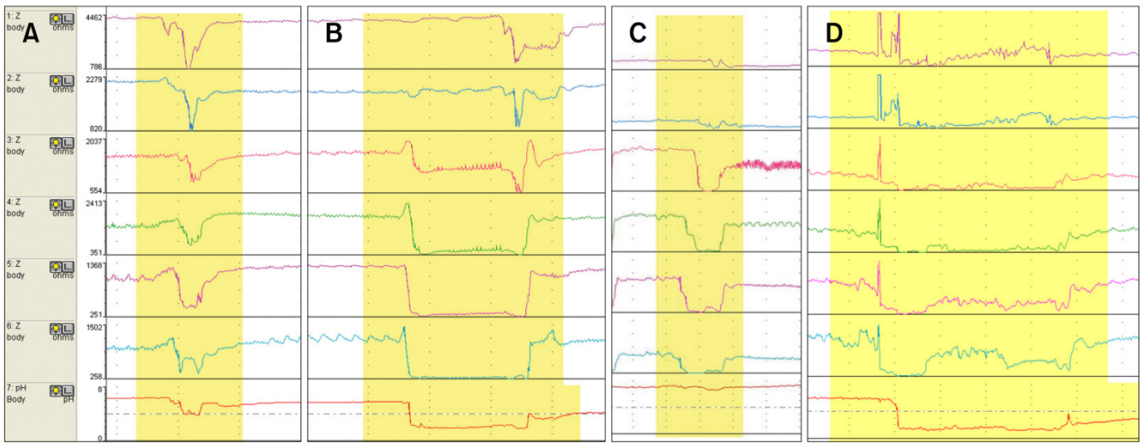


Figure 2.3. Esophageal impedance pH recordings showing examples of the 4 types of reflux. (A) Weakly acidic reflux, (B) acid reflux, (C) weakly alkaline reflux and (D) superimposed acid reflux.

2.4.5. Association between GERD and Bruxism

The mechanisms of SB may be related to various factors, including occlusal interferences, medical treatment, problems with sleep, stress, anxiety, and esophageal acidification. It's interesting to note that GERD and SB have many of these characteristics. For instance, there is evidence associating stomach physiopathology and SB pathogenesis to stress. Acid secretion and stomach motility, which are common signs

of GERD, can be affected by stress or emotions. Another thing that GERD and SB have in common is altered sleep. That is to say, GERD and sleep quality are correlated in both directions: reflux can affect sleep quality, and sleep can affect SB, which arises from sleep arousal and may be linked to airway disruptions and sleep apnea. In addition, symptoms of reflux disease are worse and more common at night [14]. There has been evidence recently suggesting that the etiology of SB may involve esophageal acidification. SB volunteers have higher rates of esophageal reflux episodes while they sleep as compared to non-SB participants. Additionally, the frequency of RMMA episodes lowers after receiving a proton pump inhibitor, a medication that stops the release of gastric acid.

2.5. Assessment Methods of Corrosion and Erosion

2.5.1. Surface Roughness Assessment

Surface roughness is determinant of surface texture that affects the amount and direction of reflected beams from the surface [99], wear of the opposing tooth linked to change in friction coefficient, wear of opposing tooth by modifying stress concentration, and gingival biological response [100–103]. It is important to recognize and address the fact that most restorations as well as appliances are chair-side selectively altered to arrange the shape or occlusal contacts, increasing surface roughness. A number of parameters were added, and the distance between the top and bottom edges of the surface profile was measured, in order to compute roughness[104]. Average height of these peaks and valleys is known as average roughness; it is denoted by the letters Sa when measured with three-dimensional equipment and Ra when measured in two dimensions. Ra value is primarily utilized in dental research, however it does not provide a detailed definition of the surface property [105]. It is a simple calculation for roughness estimation, although it might not be able to tell peaks from valleys. Therefore, another parameter, like average roughness depth Rz, may be used to measure detailed profiles [106,107].

2.5.1.1. Methods of Measuring Surface Roughness

Surface roughness can be measured quantitatively using a variety of profilometers, including contact stylus profilometer, laser profilometer and 3D optical profilometer [104,108]. The most common type was found to be the contact profilometer [109]. In addition to these tools, confocal laser scanning microscopy (CLSM) and atomic force microscopes (AFM) can be used to measure quantitative values and assess the qualitative

qualities of a surface texture. When measuring corroded surfaces, profilometers with and without contact are both useful solutions [110].

2.5.1.1.A. Contact (Stylus) Profilometer

The stylus provides a numerical result by tracing the path on a line [105,111,112]. Force 1/1000 of 1 N is applied by the contact stylus. But the sample may have scratches from the stylus's contact with it. It offers accurate assessments of the marginal differences of restorations in vitro as well as precise measurements of roughness for flat samples housed under corrosive design [109] **(Figure 2.4.)**.

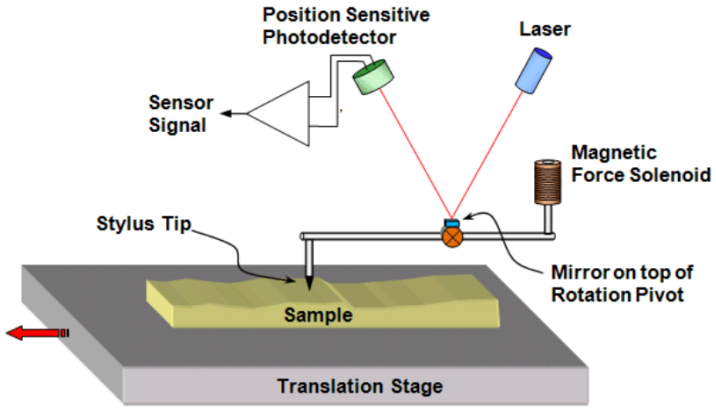


Figure 2.4. Basic elements of the stylus profilometer

2.5.1.1.B. Laser Profilometer

The laser beams have a better resolution than a contact profilometer, therefore it provides more accurate surface texture measurements. However, because of overshot production along the sharp edges of groove floors, this approach may result in some artifact forms [109]. **(Figure 2.5.)**. Additionally, because the beam scatters in liquid, moist surfaces may result in some measuring precision issues[113].

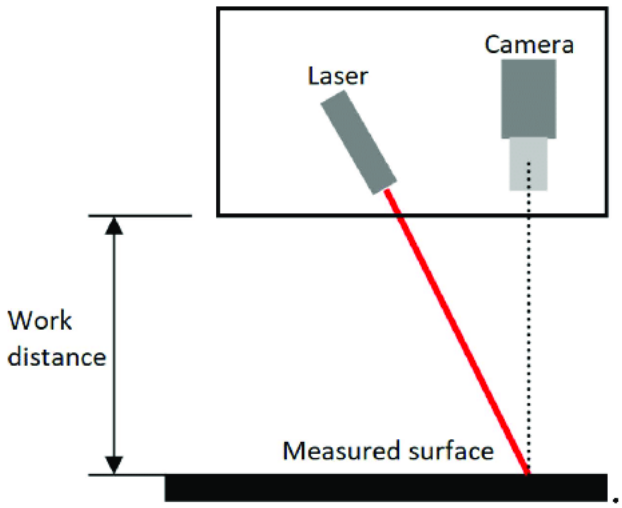


Figure 2.5. Basic elements of the laser profilometer

2.5.1.1.C. 3D Optical Profilometer (White Light Interferometer)

White light interferometry, often known as the 3D optical profilometer, was introduced in 2005 as a non-destructive optical imaging technique for solid material metrology and surface roughness measurement [114,115]. Because the beams in this approach have a low wavelength, light reflects from the sample surface instead of penetrating. These tools are designed to view a material's three-dimensional profile over a comparatively large area. The beam splitter is illuminated using a white light source. While some beams are reflected off a reference mirror, others are aimed directly at the substance. Interference fringes are produced when the beams that are reflected back from the material and mirror almost precisely match in length and converge on the imaging detector on the camera[115]. **(Figure 2.6.)**.

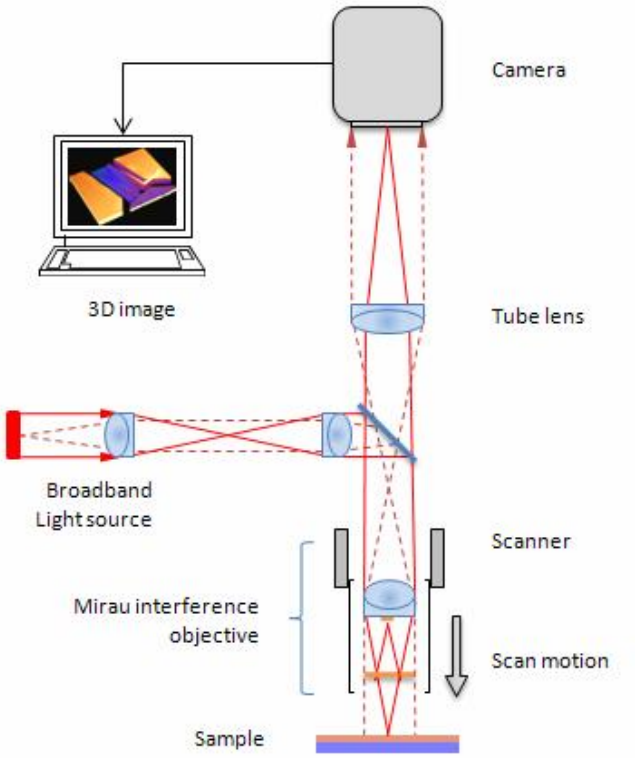


Figure 2.6. Basic elements of the 3D optical profilometer (white light interferometer)

When using the vertical scanning interferometry mode, topography images are recorded [113]. The material is lowered via a vertical range that the user defines, and vertical level variations can be defined with sensitivity down to three nanometers [115]. The material should, at the very least, partially reflect the incoming light during measurements. Light-absorbing materials were unable to provide the camera with enough light to create a contrast that would allow for an accurate measurement. As a result, samples with steep slopes typically do not reflect light back toward the camera lens, making measurements inaccurate [115].

2.5.1.1.D. Atomic Force Microscope

AFM and scanning tunneling microscopes are members of the scanning probe microscope family. AFM is now a typical piece of equipment in almost all R&D departments that deal with the engineering or characterization of material surfaces and interactions[116].

Superior imaging of many materials, such as polymers, ceramics, metals, biomolecules, cells, is primary utilize of AFM. Multiple methods of operation enable the assessment of surface topography, lateral surface structure, changes in elasticity, besides

other aspects. High resolution scanning of a variety of materials, like polymers, ceramics, metals, biomolecules, and cells, is the primary use of AFM.The evaluation of surface topography, lateral surface composition, and variations in elasticity, among other things, is made possible by several operating modes. Subnanometer precision in scanning the sample is made possible by the use of ultra-sharp probes with radii ranging from 4 to 60 nm. The surface topography determines how the cantilever moves. The source that is mounted behind the cantilever generates a diode laser beam. When the stylus and cantilever move, roughness is seen in the beams that strike the receiver. As a result, a surface pattern was created visually (Figure 2.7) [116,117].

The stylus can be used in two modes: tapping, where it taps the surface, or dynamic, where it pulses and moves up and down. When in non-dynamic mode, the tip touches the surface permanently and moves horizontally. When measuring the attraction between molecules in the non-contact mode, the stylus moves in accordance with the strength of van der Waals forces [113].

Although photographs have a greater resolution than a profilometer, the scanned area is limited to 0.25 mm2, and the scanning process takes 60 minutes for this size [113]. The scanning process is step-by-step, which explains the limited area. As a result, surface roughness depends on the scanned area and increases with broader scanning [118].

In general, it is able to quantify the height variations on the original material loss levels brought on by abrasion and erosion [113]. The utilization of both an AFM and a profilometer would increase the accuracy and reliability of the research [118].

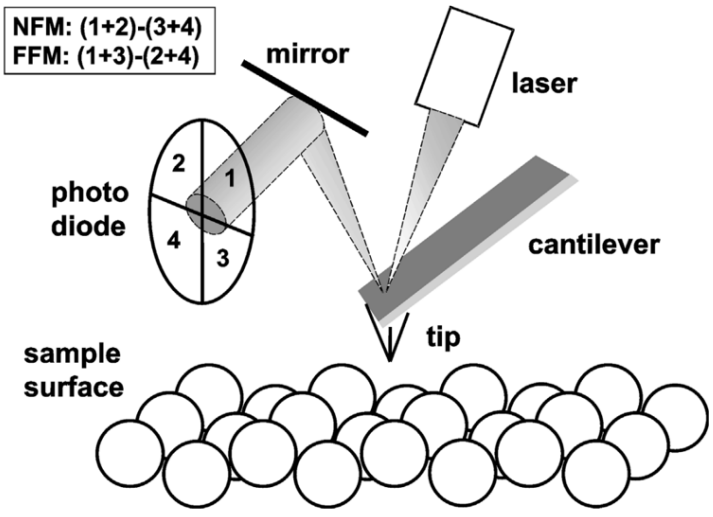


Figure 2.7. Schematic representation of an AFM.

2.5.1.1.E. Confocal Laser Scanning Microscope (CLSM)

This device is used to obtain optical slices, higher resolution pictures, and three-dimensional reconstructions from samples. Furthermore its higher resolution make this method's three-dimensional inspections of degraded surfaces are among its main advantages [113]. Compared to contact profilometers, CLSM is more sensitive and produces smaller roughness. Its short processing time, ability to evaluate topography, height profile simultaneously, ability to produce high-definition, completely focused images that exceed those of 656 scanning electron microscopes are just a few of its benefits.

On the other hand, a limitation in target area measurement is a drawback [119]. The CLSM process can be elucidated as follows: Through the lens, a gas laser light source (such as Ar/Kr or He/Ar) is focused onto a narrow focal plane of a material that is either fluorescently dyed or auto-fluorescent. There are certain wavelengths at which the laser can emit (488 nm in general). Fluorescent light released from the lit region and reflected laser light are combined by the objective lenses and detector. The identical planes that house the illumination and detection centers are the source of the confocal characteristic. The microscope stage can be raised or lowered to collect sample topographical data from various focus planes. Using data obtained as two-dimensional images from successive focal areas, the computer generates a three-dimensional image of the surface [113].

2.5.2. Surface Topography Assessment

Since only a portion of a material is scanned with a profilometer, roughness is not a reliable indicator of the topography of that material [120]. To correctly evaluate surface topography, surface roughness data is often verified using scanning electron microscope (SEM) images[121,122].

By carefully applying an electron shower to the material and then having the electrons bounce off the sample, a SEM image is created that is black and white. The source fires electrons into a vacuum-sealed chamber, and the electrons are directed, focussed, and guided by electromagnets to scan the sample and create a topographic image. After that, the electrons are found using positively charged detectors, which turn the collected data into a black-and-white picture. SEM uses variable pressure chambers to provide three-dimensional assessment of samples or wet cells and tissues. A poor-quality image is produced when electrons are prevented from interacting with the specimen by the vacuum, which keeps them from coming into touch with air molecules

and gas molecules [123,124]. Finally, by applying a conductive coating to the samples to stop the material from overheating, the image quality is enhanced. A sputter coater is used to apply a thin layer of gold to the material. A sputter coater employs argon ions to extract gold atoms off a gold plate and coat the surface with a conductive gold coating. [124]. Given that SEM imaging increases the quality of topography evaluation and allows for more extensive data collecting about texture in greater detail at a higher resolution, the results are more detailed [120,125,126]. Changes in shape and contour that the surface profilometer would miss are seen in SEM photos [121].

With SEM imaging, surface changes after erosive attack can be qualitatively evaluated [127,128]. It is possible to grade the degree of surface modification using personalized scales [113].

2.6. Weartribology

Wear results from mechanical contact, motion of a solid or liquid substance, chemical action, combined chemical and mechanical action. Wear is a loss of material from a surface. Considering some mechanisms which might be significant in industrial processes do not occur in the mouth, this nomenclature may differ slightly from that used in industry. Wear in the mouth is typically the result of several processes operating at once. One important effect that occlusal interactions might have is wear. If unchecked, it may result in impaired masticatory function, a reduction in life quality, and possibly even a worsening of systemic health [129,130]. Wear is a complex phenomenon that results from a series of connected events, and tribologists have defined it using six terms [12].

2.6.1. Abrasive Wear

Particles are the source of abrasive wear that are pushed through area that is being worn. Three-body wear and two-body wear are two distinct types of abrasive wear. In contrast to three-body wear, which occurs when an abrasive agent is situated between two surfaces that move independently, two-body wear transpires when abrasive particles are embedded within one of the surfaces.[131–133]. **(Figure 2.8.)**.

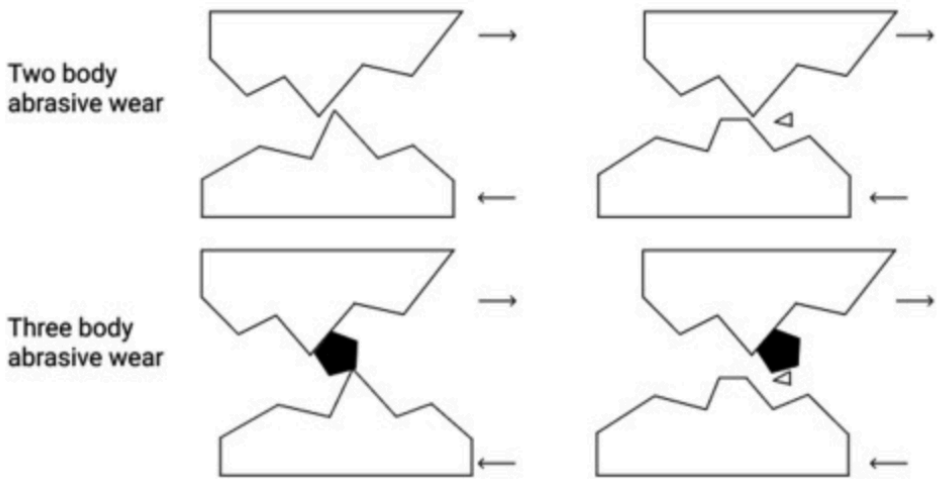


Figure 2.8. Schematic of two-body and three-body wear[134]

2.6.2. Attrition

Mineralized tooth substance's physical loss that is caused by tooth-to-tooth contact [135].

2.6.3. Adhesive wear

Using this method, submicroscopic fragments are separated from contact areas on the restorative material by sliding mandibular movements. When sliding, the restorative material is squeezed, forming a ridge in front of the sliding cusp. Tensile stresses created by the cusp's movement can lead to microcracks in the surface of the restorative material [112].

2.6.4. Corrosive wear

A chemical process known as corrosive wear results in layer which can be scraped off by coming into touch with an antagonist [136,137]. When chemicals from food, drink, bacteria, and saliva come into contact with the surfaces (material or tooth), it can dissolve chemically and produce corrosive wear. Corrosive wear can result from erosion or abrasion [137-140].

2.6.5. Erosion

Dental erosion is gradual chemical breakdown of hard tissue materials brought on by endogenous or exogenous acids in absence of bacterial action [141]. Blumia Nervosa

(BN) [142], rumination, GERD [143], and the consumption of acidic meals[144] have all been identified as causes.

Erosion begins with the superficial demineralization of tooth enamel and progresses to the dissociation of subsurface structures and significant loss of hard tissue. Acids with a pH of 5.5 or less—a crucial threshold—dissolve the crystals of hydroxyapatite found in enamel [145]. Patients with erosion frequently exhibit significant, widespread defects that gradually expose the dentin [146].

Like other dental materials, PMMA-based materials are subjected to a variety of chemical, mechanical variables in the oral environment, which may negatively impact their mechanical, physical, and chemical properties [147–151]. The chemical considerations, dental materials' surface hardness and roughness can change negatively in acidic conditions in particular [152–158].

Erosion is a common cause of dental tissue degradation, resulting in both direct dentin exposure and superficial enamel loss. Acid might originate from an external or internal source [159,160].

2.6.5.1. Extrinsic Erosion

Extrinsic factors can include exposure to gas-chlorinated swimming pools, acidic or carbonated drinks, citric pastilles, a variety of medications, saliva replacements, oral hygiene swabs, and industrial aerosols and fumes from batteries [144,161,162].

2.6.5.2. Intrinsic Erosion

Bulimia nervosa, rumination or voluntary reflux phenomena, xerostomia, malabsorption syndrome, persistent vomiting during pregnancy, subclinical regurgitation due to chronic gastritis linked to drinking, and GERD are risk factors that contribute to intrinsic erosion [161,163].

With a pH of 1-3, gastric juice that contains hydrochloric acid is highly acidic and has a high propensity for erosion, which can accelerate endogenous erosion in enamel [164]. Pepsin is another proteolytic enzyme found in gastric juice. Because pepsin may break down totally demineralized dentin collagen, it creates a risk to the exposed organic matrix when it enters the oral cavity under intrinsic erosive conditions [165,166]. Pepsin causes the dentin's collagens to be removed enzymatically and accelerates the erosion process [167]. As a result, it was proposed that the collagen network's ability to stop erosion would be compromised if the acidic environment eliminated the dentin's mineralized portions since

it would become more susceptible to the pepsin enzyme. Because intrinsic erosion occurs at a lower pH than extrinsic erosion, it damages dental hard tissues more severely [168,169]. Patients with GERD or BN typically have this associated erosion [170].

2.6.6. Fatigue wear

This is the wear and tear on a solid surface brought on by material fatigue fracture. This condition is frequently seen when surfaces roll rather than slide.

2.7. Wear testing devices and wear simulation techniques

The extremely complex wear mechanisms in the mouth can differ depending on the individual. As a result, it appears that a single wear test cannot adequately replicate the various process conditions, which are not comparable and generally accepted. But many wear mechanisms should be taken into consideration during the wear test. The majority of proposed wear tests focus primarily on one component of the many mechanisms; some even suggest that their application is adequate to properly evaluate dental materials' wear resistance. Therefore, it is essential to use laboratory testing to examine the wear factors that happen in clinical settings. Results of a laboratory test on in vitro wear should be comparable to those obtained in vivo. Various test methods may provide prediction of the entire clinical wear. Different wear simulation methods can be identified by taking into consideration three main mechanical approaches [171].

2.7.1. Tooth-brushing machines

The toothbrush/dentifrice abrasion idea is simulated using tooth-brushing equipment. A number of setup elements that are crucial from a clinical standpoint should be considered when designing a trustworthy study. The toothbrush, programmable brushing methods and processes, media, cycles (20,000–35.600 strokes), time (60–100 min), and vertical load (50–300 g) are the components that make up the concept. Important variables in the wear experiments carried out with tooth-brushing machines include dentifrice type, brush design, and brushing behavior [172–174].

2.7.2. Two-body wear machines

Two-body wear simulators were designed to replicate clinical wear. The lack of certain wear factors has resulted in different degrees of success while using these devices. Drawing comparisons and interpreting the wear results are challenging. Examples of

these kinds of machines include pin-on-disk tribometers **(Figure 2.9.)**, grinding wheel designs, two-body wear single-pass sliding, two-body wear rotational counter sample, and two-body machine sliding wear [175].

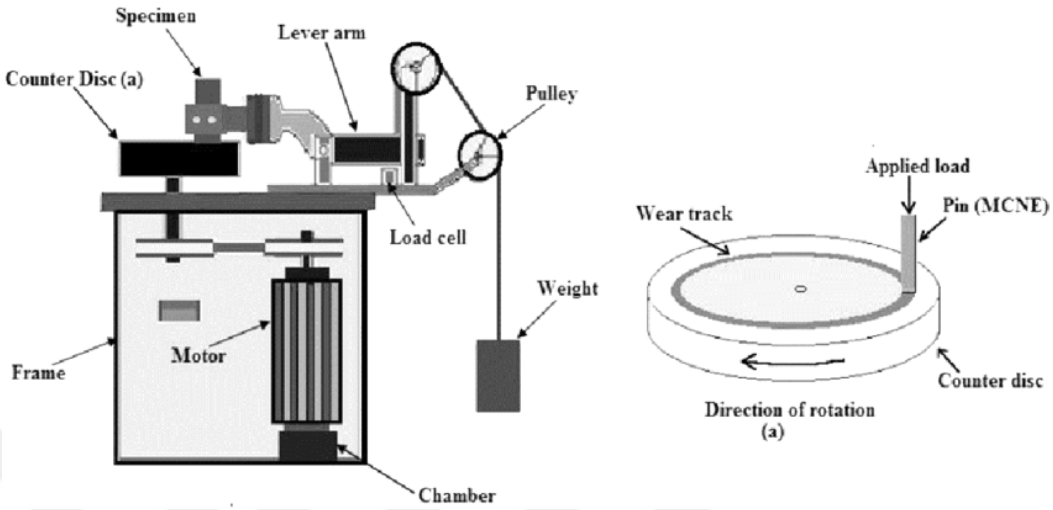


Figure 2.9. Schematic of pin-on-disk tribometer[176]

2.7.3. Three-body wear machines

The three-body wear simulators aim to mimic biological factors and the oral environment in order to classify materials based on how resistant to wear they are with relation to the sources. In vitro three-body wear machines, which can partially simulate chewing motions, are the primary use for chewing simulators. These devices have the potential to employ wear criteria in order to increase clinical success.

Figure 2.10. Chewing Simulator (MOD Chewing Simulator, Esetron; MOD Dental, Ankara, Turkey)

3. MATERIALS AND METHODS

In current study; it was intended comparasion between wear resistance and surface roughness of digitally and conventionally produced PMMA occlusal splint materials exposed to gastric acid and artificial saliva. There were two main subgroups: Gastroesophageal Reflux Disease (GERD) (gastric acid solution) as well as control (artificial saliva solution) subgroups. The samples were categorized into groups based on production methods: Conventional Method (CM) which are heat-cured acrylic (CH) and autopolymerizing resin (CC); Digital Method (DM) which are Subtractive Manufacturing (SM) and Additive Manufacturing (AM) **(Table 3.1.)**. As a result of power analysis, 13 samples were needed for surface roughness and 12 samples were needed for wear behavior. Disc-shaped samples (16 mm diameter and 3 mm thickness) were prepared as 128 in total and 16 in each subgroup. 3 of the 16 samples in each subgroup were reserved for SEM. Respectively, sample production was carried out using the DM in CAM-LAB Dental Prosthesis Laboratory for SM and Dentafab Laboratory for AM. Then, the samples to be obtained by the CM were produced in Yeditepe University Dental Prosthesis Manipulation Laboratory for CC and Simetri Dental Prosthesis Laboratory for CH in accordance with the recommendations of the manufacturer. Firstly, all materials' surface roughness were measured with 3D optic profilometer (Profilm 3D, Filmetrics, USA) then 52 samples were exposed to gastric acid while 52 of them were exposed to artificial saliva.

Before wear test, all samples were scanned with a model scanner (D1000, 3Shape GmbH, Copenhagen, Denmark) to determine amount of material lost before and after the wear process. STL format was utilized to save the scan data. For chewing simulator, samples were fixed 26 mm diameter water pipes with autopolymerizing acrylic (Imicryl, Konya, Turkey) and placed at the apparatus in the experimental setup. Wear process was carried out on the samples by adjusting the parameters. Then, two body wear tests were performed with opposing contact of the metal pin on disc-shaped occlusal splint material.

After that, all measurements were made again. The data obtained for each sample was transferred to program called Geomagic Control (3D Systems Inc., Rock Hill, USA), which transformed data points of surface in order to get three-dimensional image needed for analysis. Statistical analysis was performed from the scan data and results.

3.1.Power Analysis

PASS 15 Power Analysis and Sample Size Sofware, 2017 (NCSS LLC, Kaysville, UTAH, USA) were used to determine the sample size. Sample size of 13 was obtained for each subgroups. The total sample of 104 subjects achieves 80% power to detect differences among the means the alternative of equal means using an F test with a 0.05 significance level.

Table 3.1. Materials Used in the Study and Manufacturers

Material	Code	Name	Main Composition	Manufacturer
Heat-cured acrylic Material	**CH**	**Duracryl**	**PMMA**	**Duradent, Spofa Dental, Czech Republic**
Autopolymerizing Resin Material	**CC**	**ProBase Cold**	**PMMA**	**Ivoclar Vivadent AG, Schaan, Liehchtenstein**
Subtractive Manufacturing Material	**SM**	**Ceramill-A Splint**	**PMMA**	**Amann Girrbach AG, Koblach, Austria**
Additive Manufacturing Material	**AM**	**Keysplint Hard**	**PMMA**	**Keystone Industries GmbH, Singen, Germany**

3.2. Preparation of Samples Used in Research

3.2.1. Conventional Method (CM)

3.2.1.1. Specimen Preparation for Heat-cured Acrylic (CH)

A special flask which has the material of nickel-titanium is designed for the CM. It has four screw points at the corners so that the pressure can be achieved **(Figure 3.1., Figure 3.2. and Figure 3.3)**.

Diameter of materials was determined as 16 mm and thickness was 3 mm. Therefore, dimensions of holes were made according to this size.

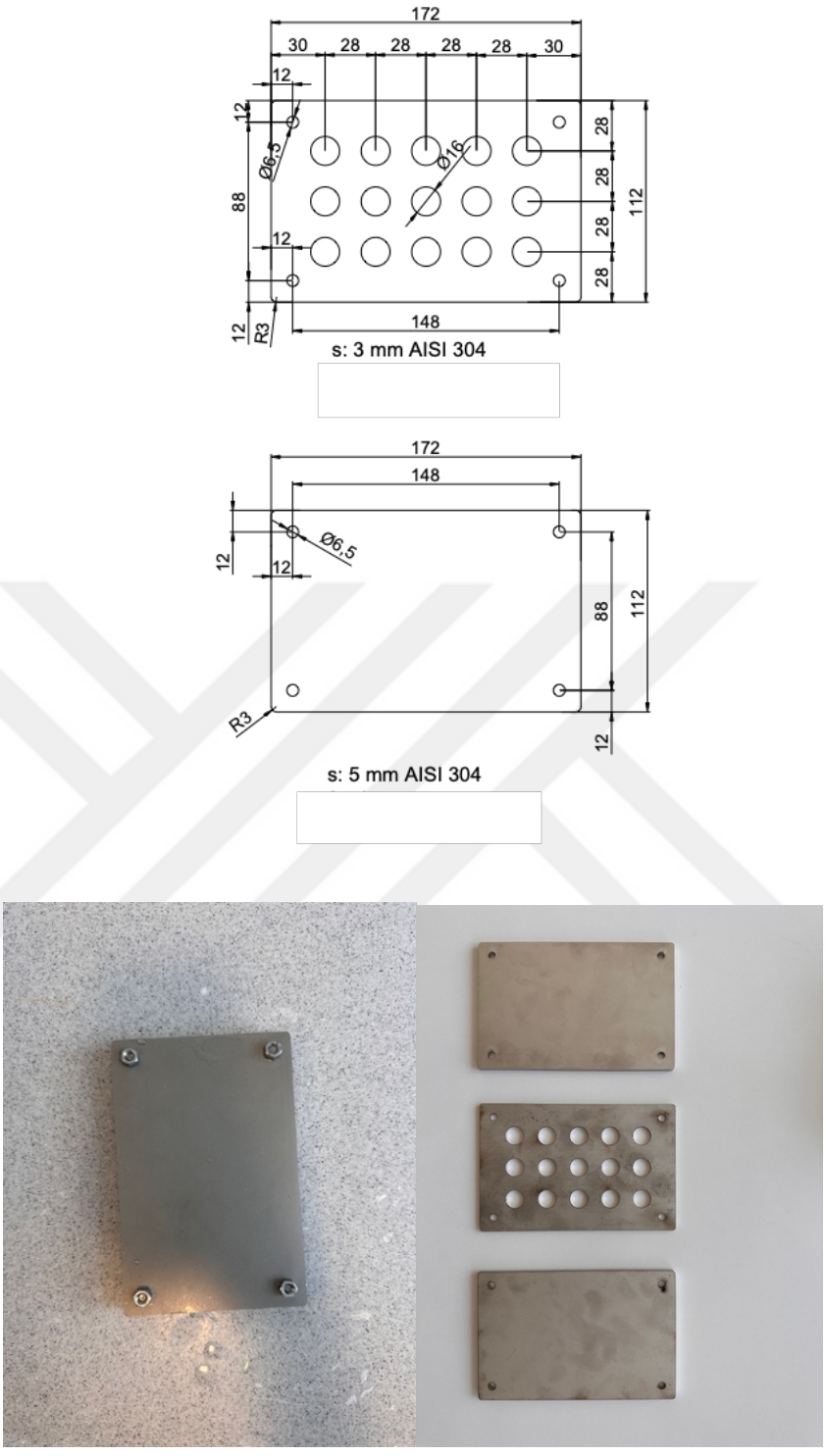
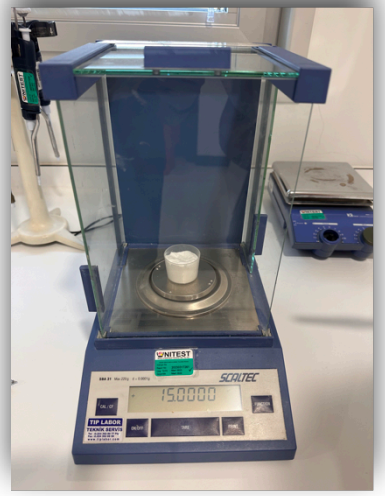


Figure 3.1., Figure 3.2. and Figure 3.3. Flask design for the CM.

Precision scale was used for the acrylic powder and liquid measurement according to manufacturer's instructions **(Figure 3.4. and Figure 3.5.)** with the ratio of 23,4 g polymer (powder): 10 ml monomer (liquid) at room temparature.



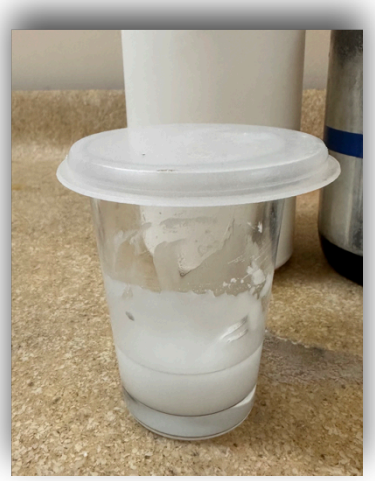


Figure 3.4 and Figure 3.5. The precision scale was used for the acrylic powder and liquid measurement according to manufacturer's instructions.

Lids of flasks were closed and tightened with secrews via appropriate pressure. CH specimens were cured for 25 minutes at 100°C in a water bath. Flasks that had been taken out of polymerization apparatus were allowed to come to room temperature at conclusion of this time. Then, they put into water bath for gradually cooling **(Figure 3.6. and Figure 3.7.).**

Figure 3.6. and Figure 3.7. Water bath at 100°C for 20 min and then allowed to cool slowly in the water bath.

3.2.1.2. Specimen Preparation for Auto-polymerizing Resin Group (CC)

The measurement of the acrylic powder and liquid was done according to manufacturer's instructions with the ratio of 15 g polymer (powder): 10 ml monomer (liquid) at room temparature. The lids of the flasks were closed and tightened with screws via appropriate pressure. Auto-polymerizing resin specimens were placed into a pressure polymerizing apparatus with 6 bar of pressure at 40° C for 15 minutes **(Figure 3.8. and Figure 3.9)**.

Figure 3.8. and Figure 3.9. Pressure-polymerizing apparatus (Silit Sicomatic; Germany)

3.2.2. Digital Method (DM)

3.2.2.1. Specimen Preparation for Subtractive Manufacturing Group (SM)

For SM, MillBox CAD Software (CIMsystem, Italy) was used for designing the materials' proportions as 16 mm diameter and 3 mm thickness cylindric shape **(Figure 3.10. and Figure 3.11.)**.

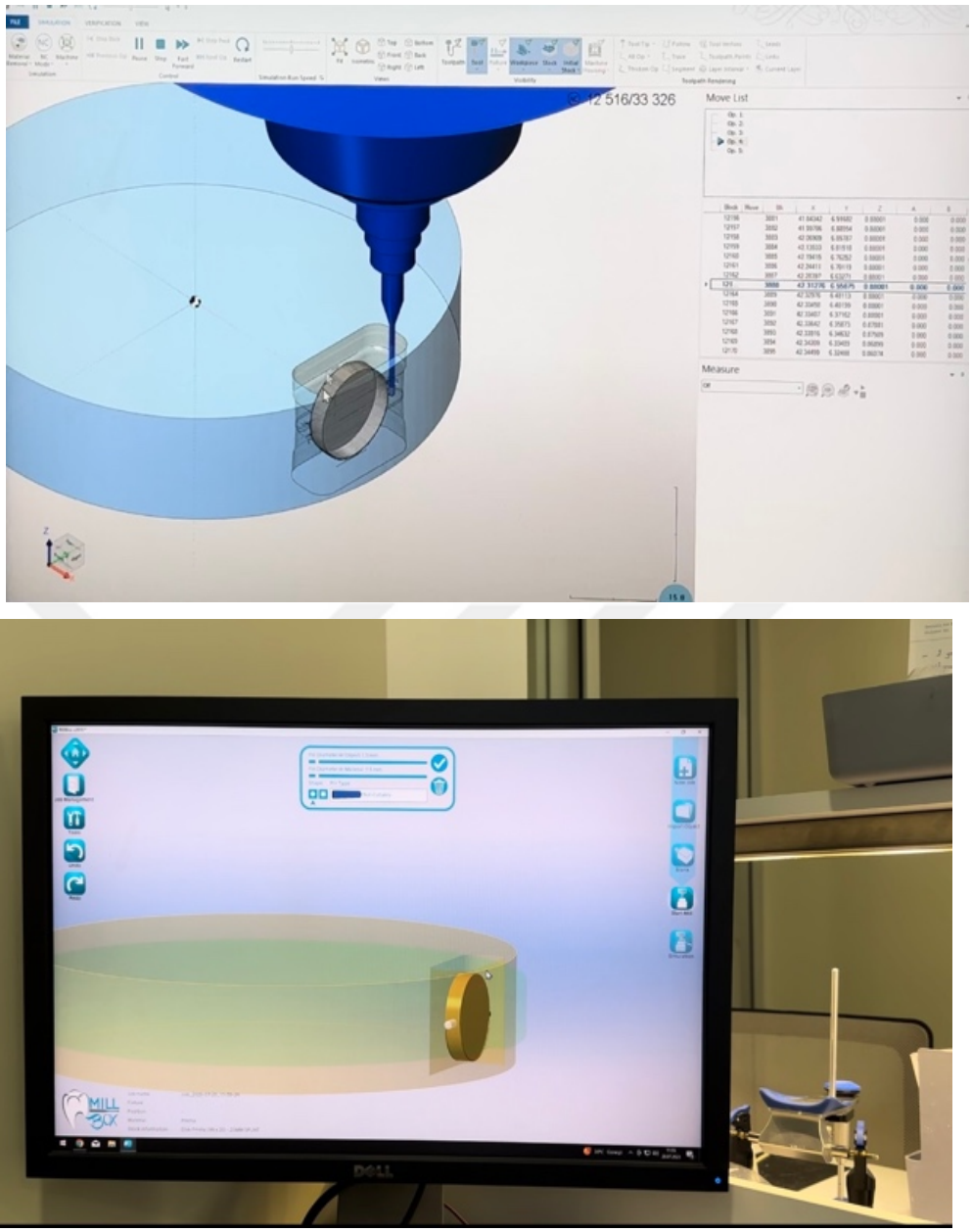


Figure 3.10. and Figure 3.11. design of samples on disk with MillBox CAD Software (CIMsystem, Italy)

Axis speed, bur position and depth of cut settings were selected. Designing data and production commands in STL format were sent to the 5-axis milling machine (Dentifa Nio) **(Figure 3.12. and Figure 3.13)** A low speed, water-cooled diamond bur was used to cut out the samples that were scraped off the block. **(Figure 3.14.)**. After the process 35 specimens were achieved from a block with 98*20 dimensions **(Figure 3.15.)**.

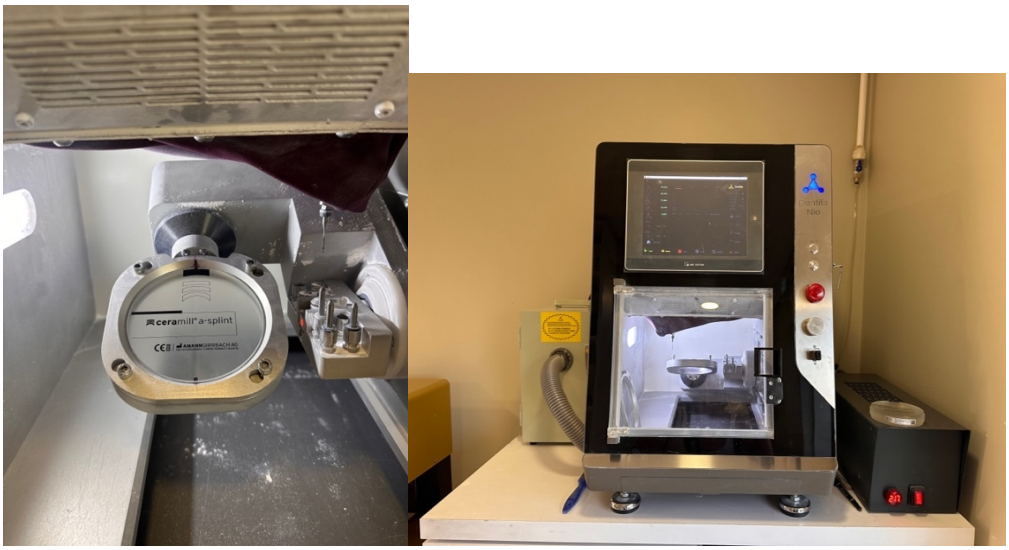


Figure 3.12. and Figure 3.13. Subtractive Manufacturing

Figure 3.14. Milling process with low speed water-cooled diamond saw

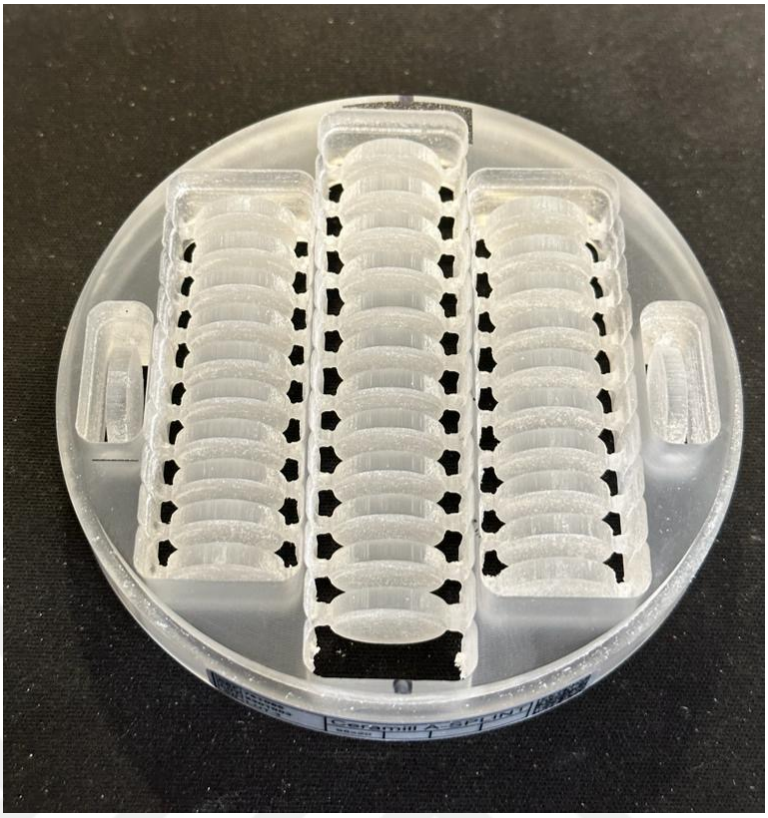


Figure 3.15. Achieving 35 specimens from a block with 98*20 dimensions (Ceramill A splint, Amann Girrbach AG, Koblach, Austria)

3.2.2.2. Specimen Preparation for Additive Manufacturing Group (AM)

In AM technic, Rhino-5 Design Software Program (Rhinoceros, TLM, Inc., Seattle, Washington, America) was used. The specimens' proportions are obtained as 16 mm diameter and 3 mm thickness cylindric shape **(Figure 3.16. and Figure 3.17.)**.

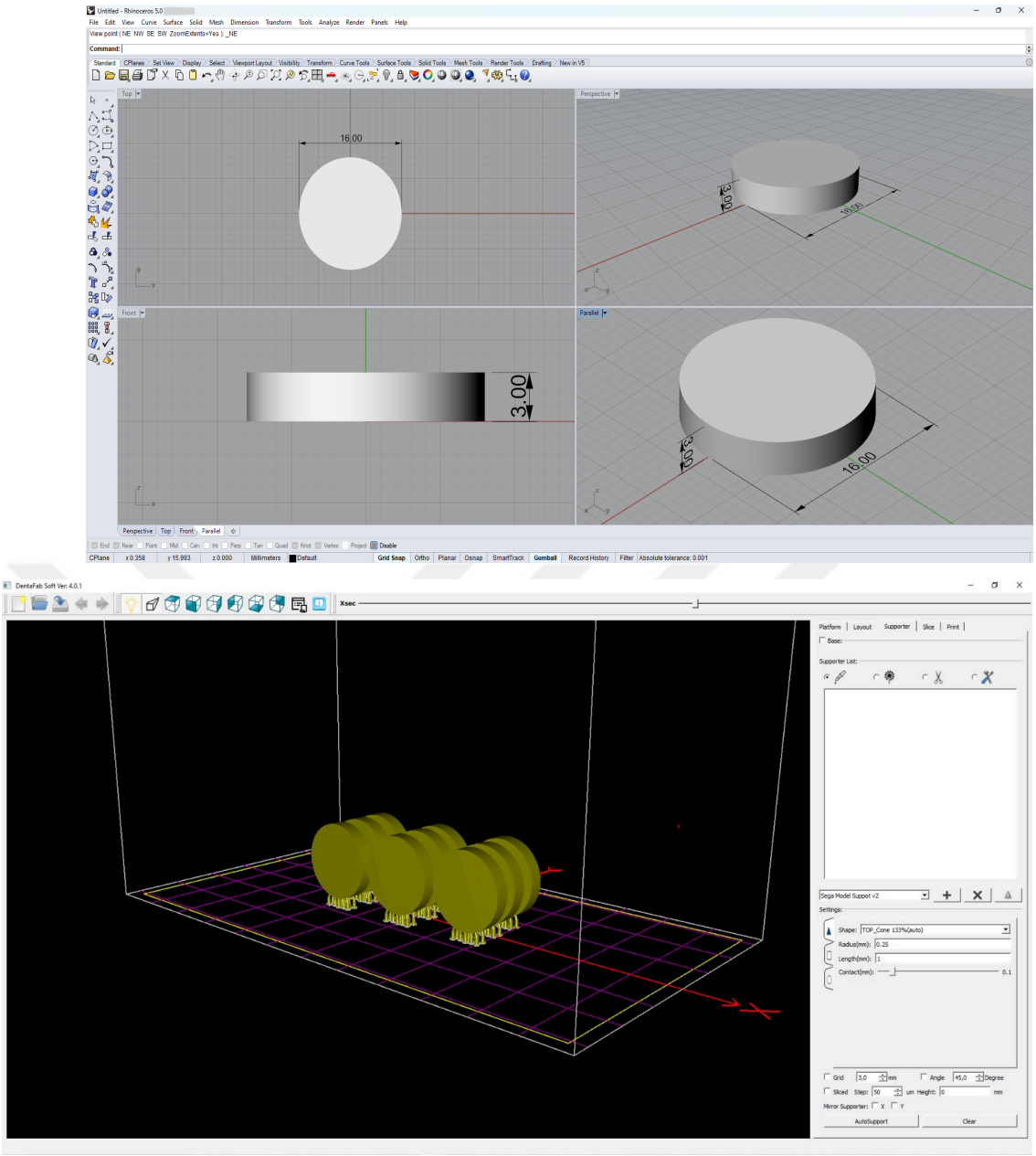


Figure 3.16. and Figure 3.17. Additive Manufacturing Technic CAD Software Program (Rhinoceros, TLM, Inc., Seattle, Washington, America)

Digital Light Processing (DLP) 3D-printed machine DentaFab SEGA 3D Printer (2013, Istanbul, Turkey) **(Figure 3.18. and Table 3.2.)** was used and photopolymerized acrylic liquid resin KeySplint Hard (Keystone Industries GmbH, Singen, Germany) was hardened layer by layer with 90 degree angle **(Figure 3.11.)**.

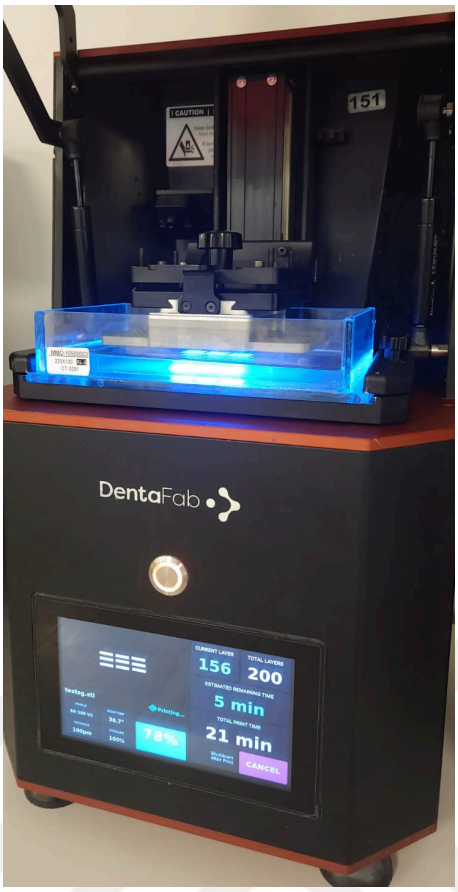


Figure 3.18. Addivitive Manufacturing DentaFab SEGA 3D Printer (2013, Istanbul, Turkey)

Figure 3.19. Additive Manufacturing Resin KeySplint Hard (Keystone Industries GmbH, Singen, Germany)

Table 3.2. Properties of the 3D printed machine

Machine	DentaFab SEGA 3D Printer
Print technology	Digital Light Processing (DLP)
Control screen	7’’ Touch Screen
Print area	120 x 67,5 x 110 mm
Light source	405 nm UV Projector
Print speed	60 mm/h

Following the printing procedure, the specimens were cleaned twice for a total of six minutes using 90% isopropyl alcohol (3 minutes for pre-cleaning and 3 minutes for primary cleaning).

The materials were post-cured to achieve the total polymerization after drying with compressed air. The samples were subjected to final polymerization in the ultraviolet polymerization device Otoflash G171 (2000 flashes, 2 min cooling, 2000 flashes: Hinrichs Dental, Goslar, Germany) **(Figure 3.20.)**. Finally, samples were achieved **(Figure 3.21.)**.

Figure 3.20. Ultraviolet polymerization device Otoflash G171 (NK-Optik GmbH, Baierbrunn, Germany)

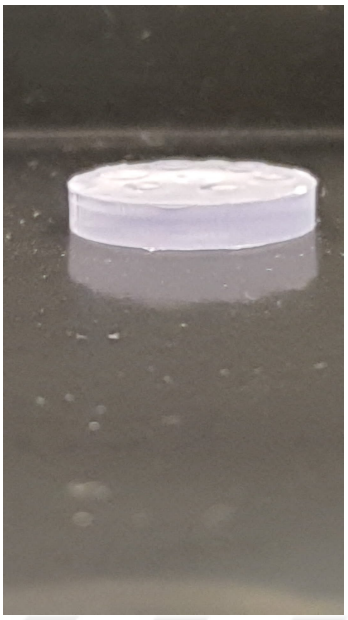


Figure 3.21. 3D printed acrylic sample

3.3. Grinding and Polishing

Every specimen was ground to the desired final size by automatic polishing machine (Ecomet 250, Buehler, Lake Bluff, Illinois) at 400 rpm under water cooling **(Figure 3.22.)**. P600, P800, P1200 waterproof silicon carbide papers (English Abrasives & Chemicals Ltd., Stafford, UK) were used **(Figure 3.23.)**. Silicon carbide paper was replaced every three specimens. Furthermore, polishing was done with goat hairbrush using pumice, then with gyspsum powder and cotton disc. Finally, the polishing process was completed with polishing paste (Ivoclar Vivadent AG, Schaan, Liechtehstein) **(Figure 3.24 and Figure 3.25.)**.

Figure 3.22. Ecomet 250, Buehler, Lake Bluff, Illinois

Figure 3.23. Waterproof silicone carbide sand papers with granulometers up to 600, 800 and 1200 meshes (English Abrasives & Chemicals Ltd., Stafford, UK).

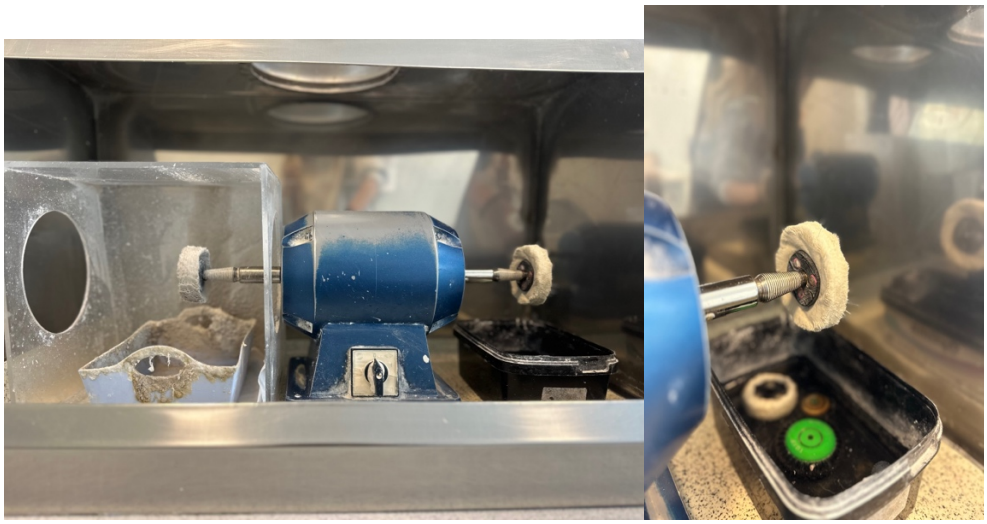


Figure 3.24. and Figure 3.25. Polishing with goat hairbrush using pumice, then with gyspsum powder and cotton disc.

3.4.Ultrasonic Cleaning

The samples were cleaned for ten minutes at room temperature using distilled water in an ultrasonic bath (AS 8772 Ultrasonic Cleaner, General Home Orsay Ltd. Şti., İstanbul, Turkey) **(Figure 3.26. and Figure 3.27.)**.

Figure 3.26. and Figure 3.27. Samples cleaned in distilled water for 10 minutes with ultrasonic cleaner (AS 8772 Ultrasonic Cleaner, General Home Orsay Ltd. Şti., İstanbul, Turkey)

After the grinding and polishing steps all materials were ready for the experiment **(Figure 3.28.)**.

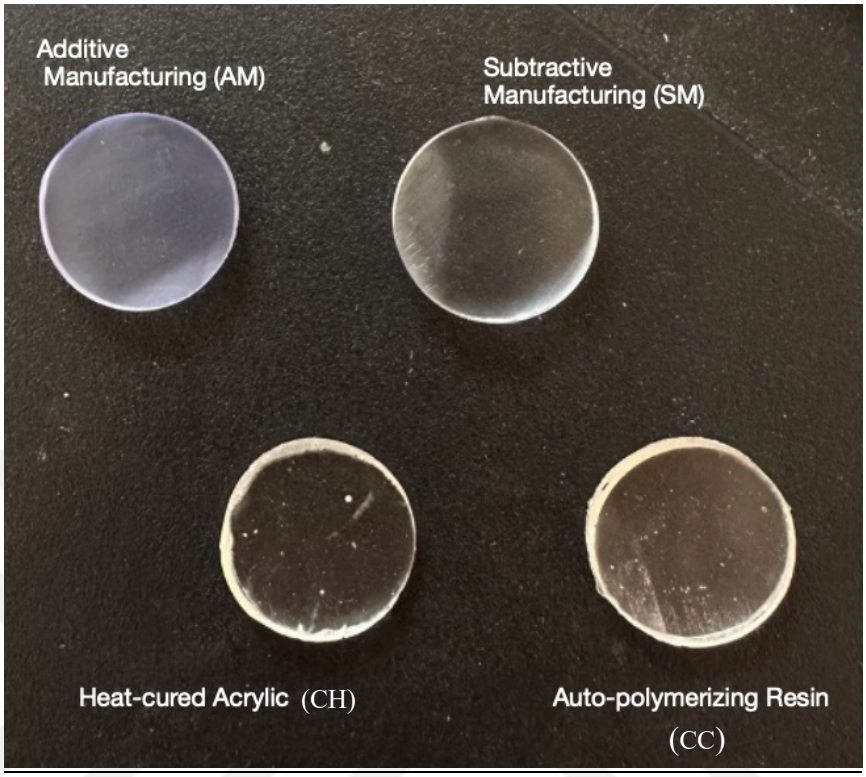


Figure 3.28. Specimens for all groups

By casting autopolymerizing acrylic (Imicryl, Konya, Turkey), all generated disk-shaped samples were put in plastic holders within the chewing simulator's test chambers. **(Figure 3.29.)**.

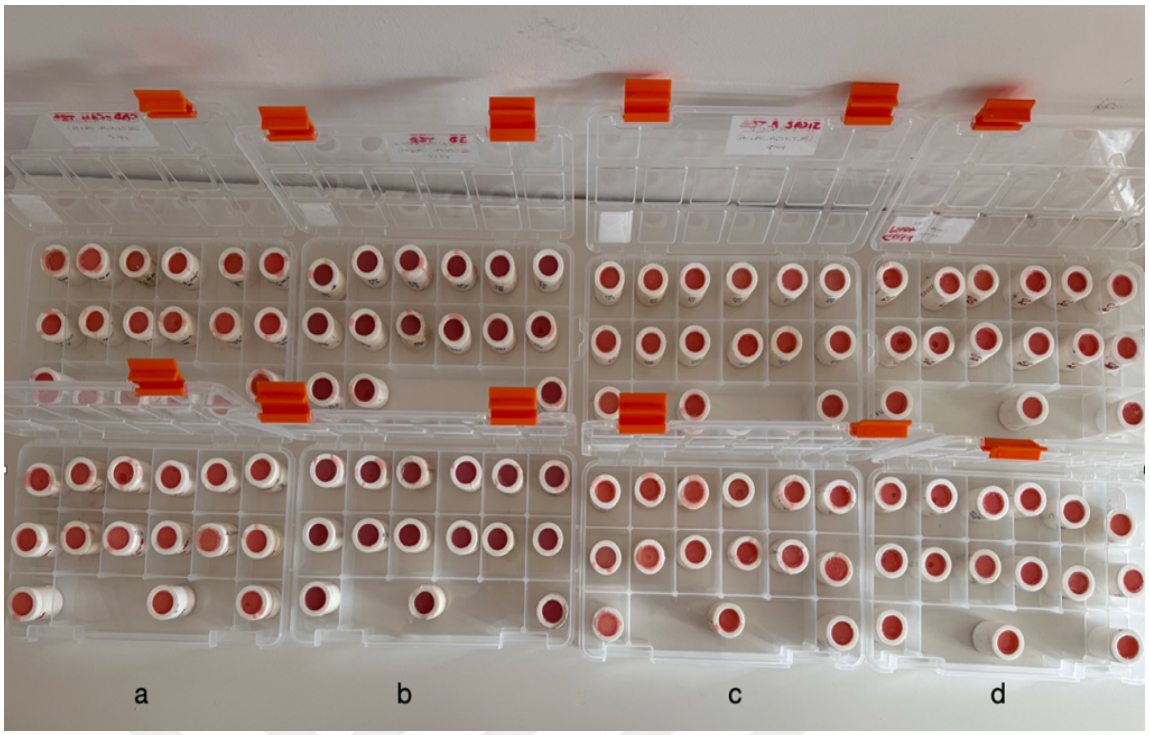


Figure 3.29. (a) Subtractive Manufacturing Group, (b) Additive Manufacturing Group, (c) Heat-cured Acrylic, (d) Auto-polymerizing resin preparation of samples for placement in chewing simulator chambers

On the plastic part around the discs; lines, dots and crosses were created with bur from three different regions at an angle of 120 degrees **(Figure 3.30.).** These points were used for reference purposes as fixed measurement points in order to match the scans to be made before and after the test in the chewing simulation with minimum error.

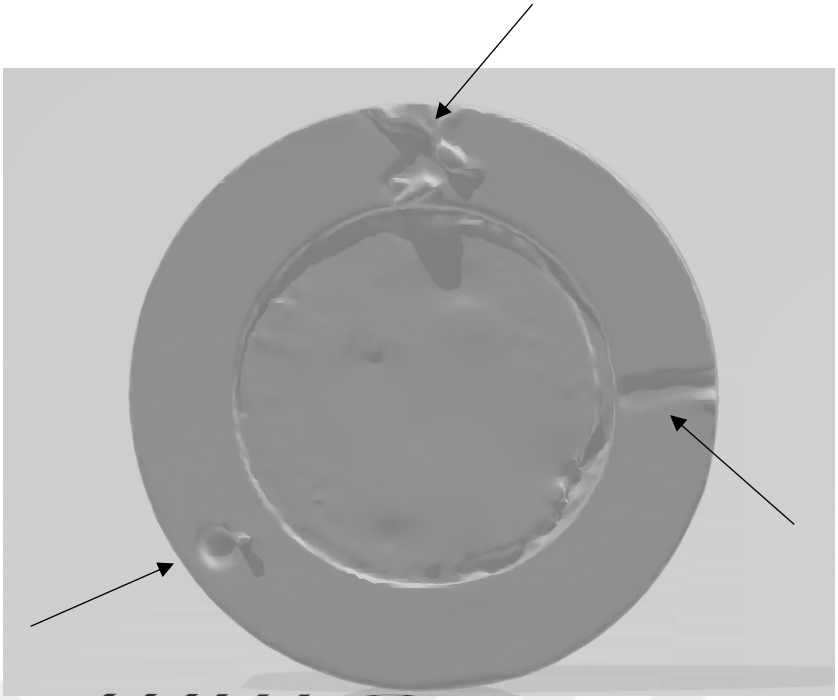


Figure 3.30. On the plastic part around the discs; lines, dots and crosses were created from three different regions at an angle of 120 degrees.

3.5.Preparation of Antagonists Used in the Study

For this study metal pin of the chewing simulator was used.

3.6.Evaluation of the surface roughness (Ra)

For avoiding surface damaging, surface roughness was evaluated with 3D optic profilometer (Profilm 3D, Filmetrics, Inc-A KLA Company, USA) at Çukurova University Science and Technology Application and Research Center (ÇÜMERLAB) **(Figure 3.31.)**.

The initial overall surface roughness average value was measured less than 0.2 micrometers for each sample; but we wanted to assess average roughness of area where wear will occur (apptoximately the middle point of the cylindrical sample) for both T_0 and T_1 times. Because; we only wanted to study surface roughness of wear area; otherwise, existance of artificial saliva and acid would affect the roughness results.

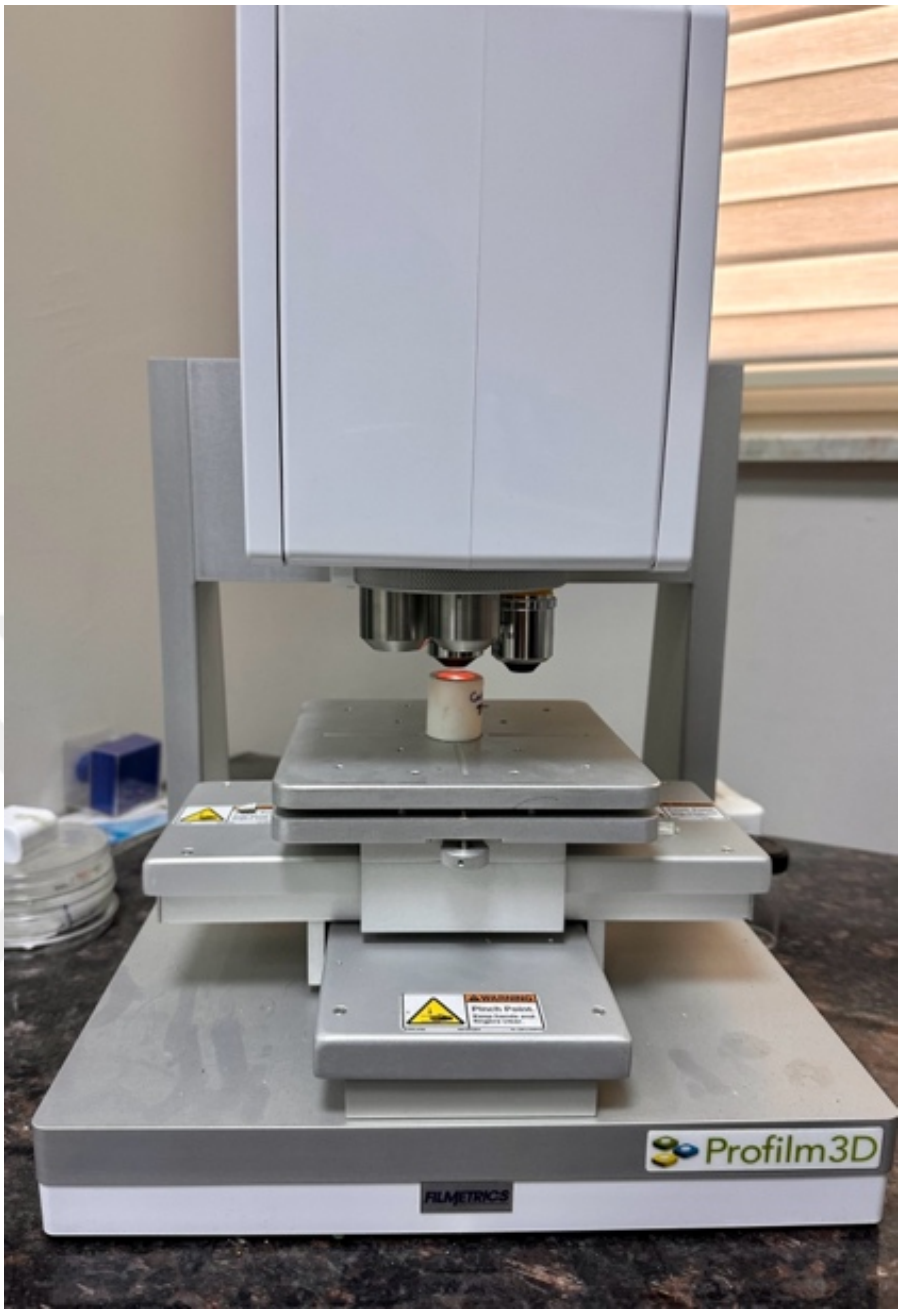


Figure 3.31. Surface roughness machine (Profilm 3D, Filmetrics, Inc-A KLA Company, USA)

The surface roughness was assessed (Profilm 3D, Filmetrics, Inc-A KLA Company, USA) (Figure 3.32.)

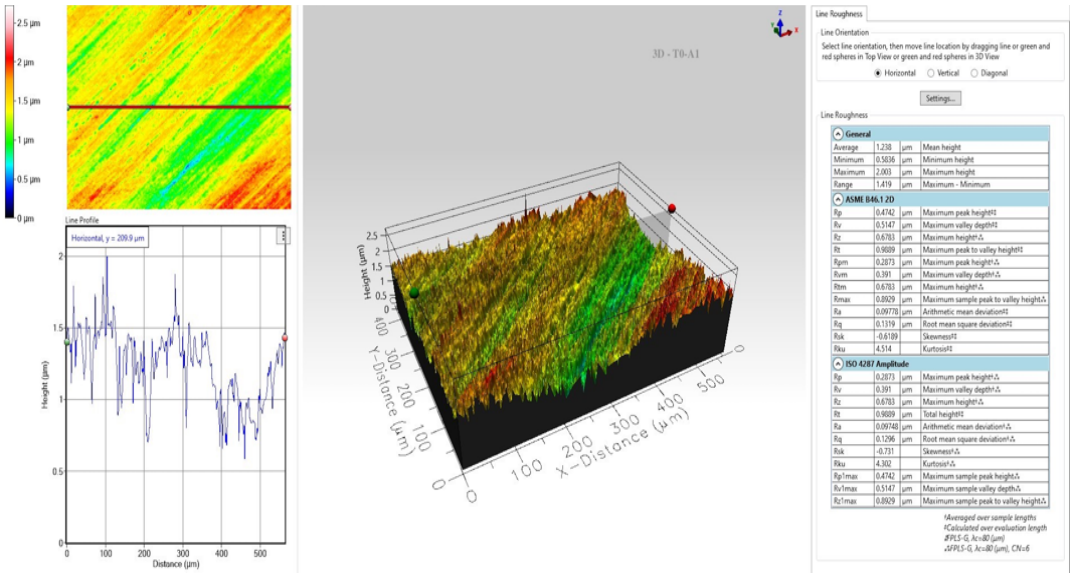


Figure 3.32. Filmetrics profilm 3D Software

3.7.Taking Scanning Electron Microscope Images of Samples

Two samples from every group were investigated with SEM (FEI Quanta FEG 650 SEM (JSM IT-800, JEOL Ltd., Akishima, Tokyo, Japan) before and after the wear process as well as after solution exposure. The SEM device at Çukurova University Science and Technology Application and Research Center (ÇÜMERLAB) was used **(Figure 3.33.).** The SEM images of the specimens showed visual signs of volume loss, including irregularities, pits, valleys, scratches, and inhomogeneities. The specimens were gold sputtered with a gold layer using a sputter coater in a vacuum evaporator prior to SEM analysis.

Figure 3.33. Scanning Electron Microscope **FEI Quanta** FEG **650** SEM (JSM IT-800, JEOL Ltd., Akishima, Tokyo, Japan)

3.8. Solution Preparations

Two different solutions were prepared at Yeditepe University Faculty of Pharmacy as GERD (gastric acid solution) and control (artificial saliva solution) subgroups **(Figure 3.34. and Figure 3.35.)**. GERD subgroup was simulated under the conditions of 4 pH and 37^{O}C. The artificial saliva group which is control subgroup was simulated under the conditions of 7 pH and 37^{O}C. Each solution's formula and pH has been demonstrated **(Table 3.3.)**.

Table 3.3. Solution formulas and pH's table

Solution Name	pH	Temperature	Formula*
GERD (Gastric Acid Solution)	4	37^{o}C	0,5 g NaCl + 6 mol/L HCl + 99,5 mL distilled water
Control Subgroup (Artificial Saliva Solution)	7		1,3 g NaCl + 0,03 g $MgCl_2.6H_2O$ + 0,08 g $CaCl_2.2H_2O$ + 0,3 g KCl + 0,9 g KOH + 472 µL H_3PO_4

Figure 3.34. Gastric acid solution (GERD Subgroup)

Figure 3.35. Artificial saliva solution (Control Solution Subgroup)

Half of each group of a total of 104 materials (52) were kept in gastric acid and half in saliva for 45 hours at room temparature, for simulating 6 months. Plastic boxes were chosen as acid resistant. The bottom of each compartment of the boxes was chosen to be oval in shape so that the solutions could contact the entire surface to be measured. **(Figure 3.37.)**.

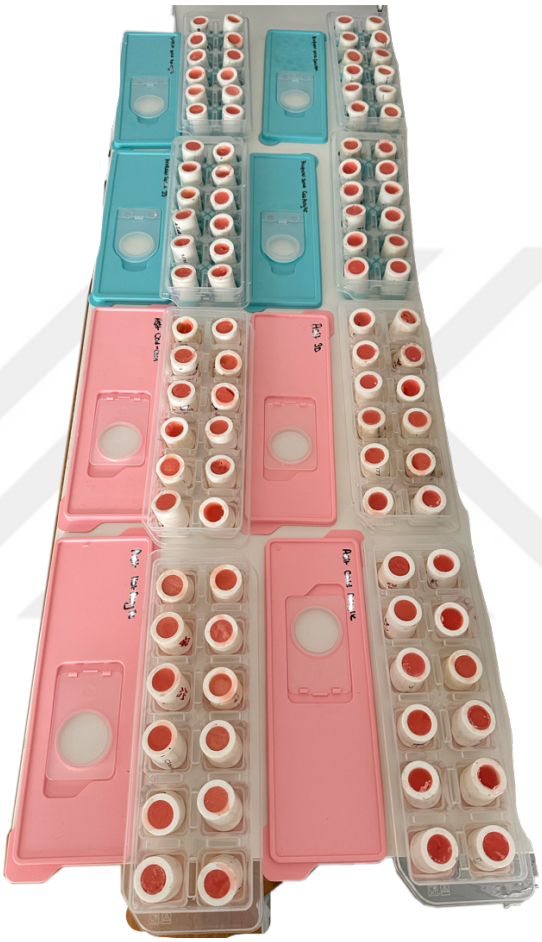


Figure 3.37. Materials were kept in gastric acid and half in saliva for 45 hours at room temparature.

3.9.Scanning Samples with a Scanner

104 disc-shaped samples were scanned with a model scanner (D1000, 3Shape GmbH, Copenhagen, Denmark) to determine amount of material lost before and after the wear process **(Figure 3.38. and Figure 3.39.)**.

The desktop model scanner transfers the image obtained by its 4 cameras with 5 megapixel resolution to computer monitor to which it is connected. It has a simultaneous multiple linear scanning mechanism with a sensitivity of 5 µm with blue LED light. Every experimental group had the identical scanning settings since the scanner was calibrated

thereafter. Scanning data was recorded in STL format and numbered. These STL files are designated as initial reference scans.

Figure 3.38. 3D laser scanner (D1000, 3Shape GmbH, Copenhagen, Denmark)

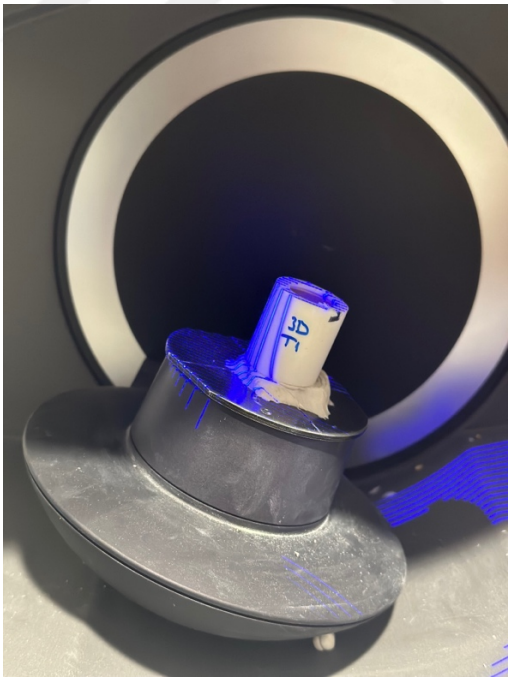


Figure 3.39. Scanning of the samples

3.10. Chewing Simulator and Abrasion Test

In our study, two body wear tests were performed with opposing contact of the metal pin on disc-shaped occlusal splint material. A six-chamber chewing simulator (MOD Dental, Esetron Smart Robot Technologies, Ankara, Turkey) was used. Dual axis chewing simulator is a computer-controlled device that can apply force in two directions (vertical and horizontal). There are six plastic sample holders and a liquid reservoir surrounding these holders. During application, the liquid reserves are filled with distilled water. Samples from all groups were subjected to 100 N loading at 37°C. **(Figure 3.40. and Figure 3.41., Figure 3.42. and Figure 3.43)**.

Figure 3.40. Chewing Simulator (MOD Chewing Simulator, Esetron; MOD Dental, Ankara, Turkey)

Figure 3.41., Figure 3.42. and Figure 3.43. Sample with 100N and distilled water on chewing simulator

During the cycle, the vertical movement amount of the antagonist metal pin was set to 6 mm and horizontal movement was set to 0.7 mm. The number of 20,000 cycle, which corresponds to a clinical wear of 6 months for night-time splint users, was applied to the samples. Also, 37°C intraoral simulation section was selected for this study. The values of the dual-axis chewing simulator used as follows **(Table 3.4. and Figure 3.44.)**:

Table 3.4. Parameters of chewing simulation

Parameter	Characteristics
Sample quantity tested	104
Weight per sample	10 kg
Number of cycles	20,000
Cycle frequency	1,66 Hz
Vertical movement	6 mm
Horizontal movement	0,7 mm
Forward Speed	60 mm/s
Backward speed	60 mm/s
Hot/cold bath temparature	37° C

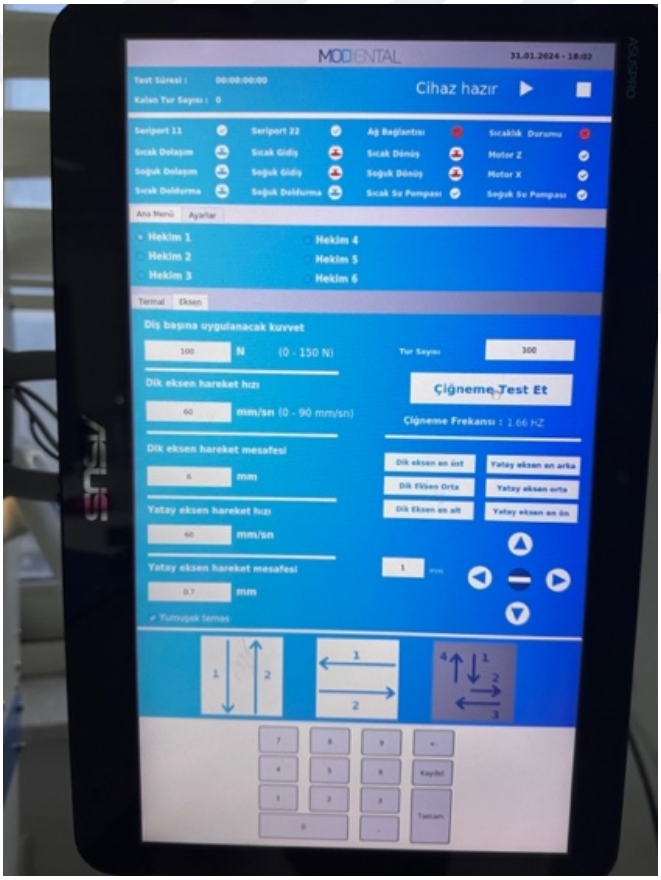


Figure 3.44. Chewing simulator's parameters

3.11. Wear Measurements

After chewing simulator, samples were scanned by using the same scanner. The data obtained for each sample was transferred to the metrology software (Geomagic Control X 2022, 3D Systems, Luxembourg, Luxembourg), which transformed the data points of surface to obtain three-dimensional image required for analysis. The images transferred to the program were cut on a certain plane and brought to equal dimensions in order to display the areas exposed to wear **(Figure 3.45.)**. This measurement gave us the linear value.

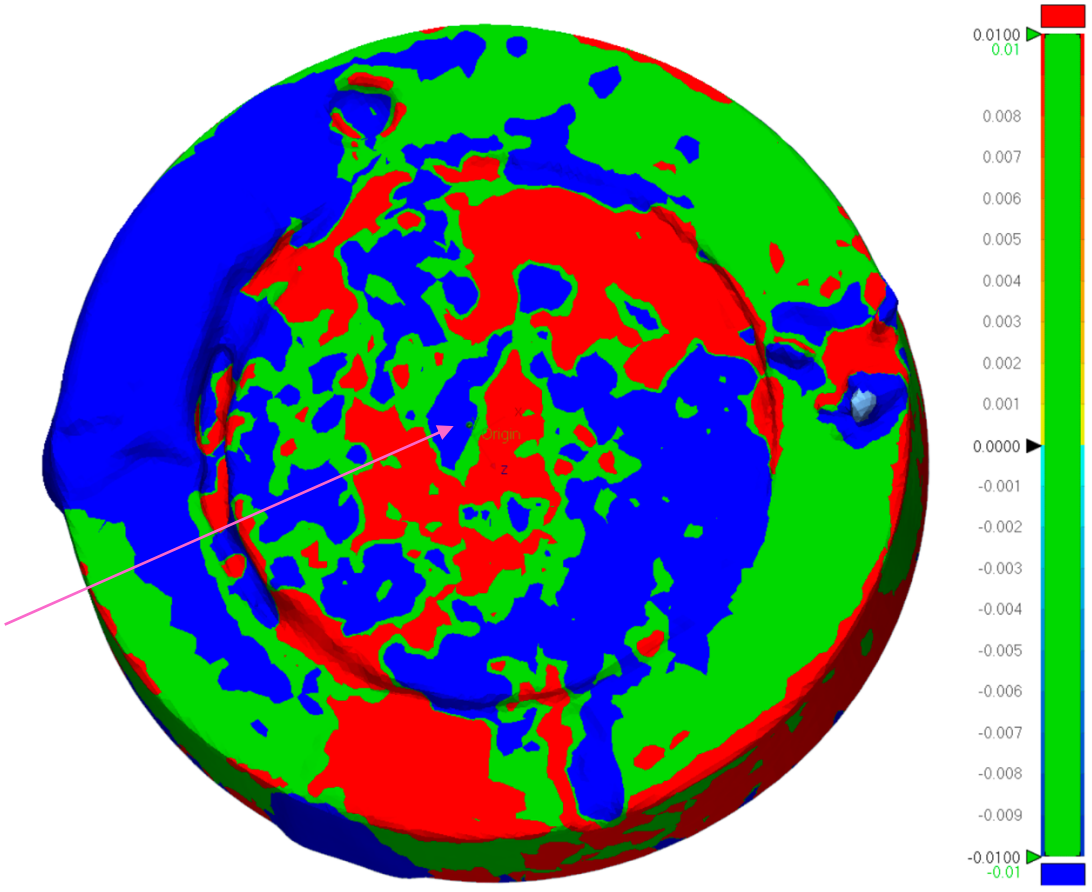
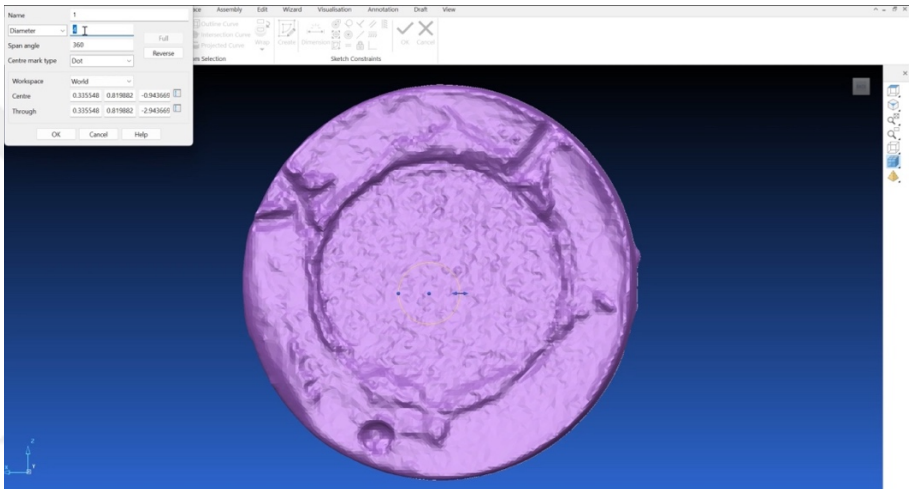


Figure 3.45. Geomagic Control (Geomagic Control X 2022, 3D Systems, Luxembourg, Luxembourg)

Afterwards, we transferred the beginning and end of all STL formats to the software called Autodesk Powershape Ultimate 2019 (San Rafael, California, U.S.) and made volume calculations. The volume calculation difference of the samples was determined by taking cylindrical sections from worn area in middle of the beginning and end of the samples and recorded for statistical values **(Figure 3.46, Figure 3.47. and Figure 3.48., Figure 3.49., Figure 3.50)**. The wear area of all models was controlled with the X, dot and line symbols we placed around the water pipe. After the chewing simulator, it was observed that the wear of all samples was at the exact middle point.



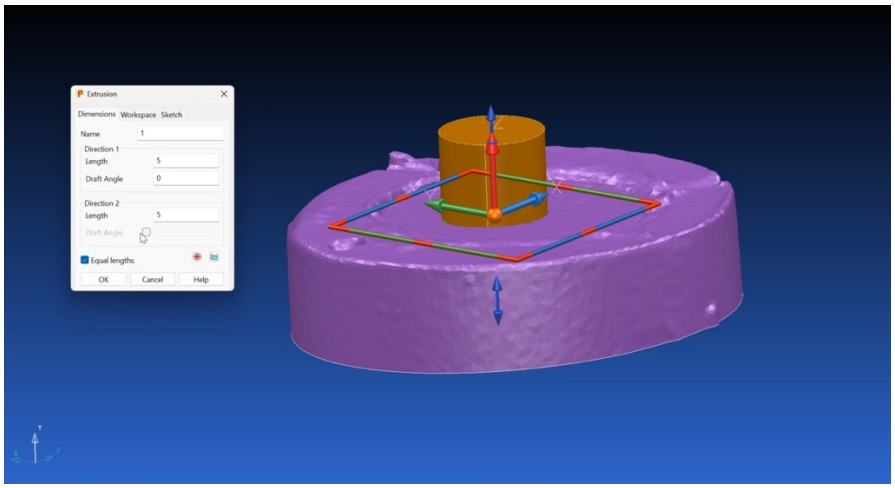
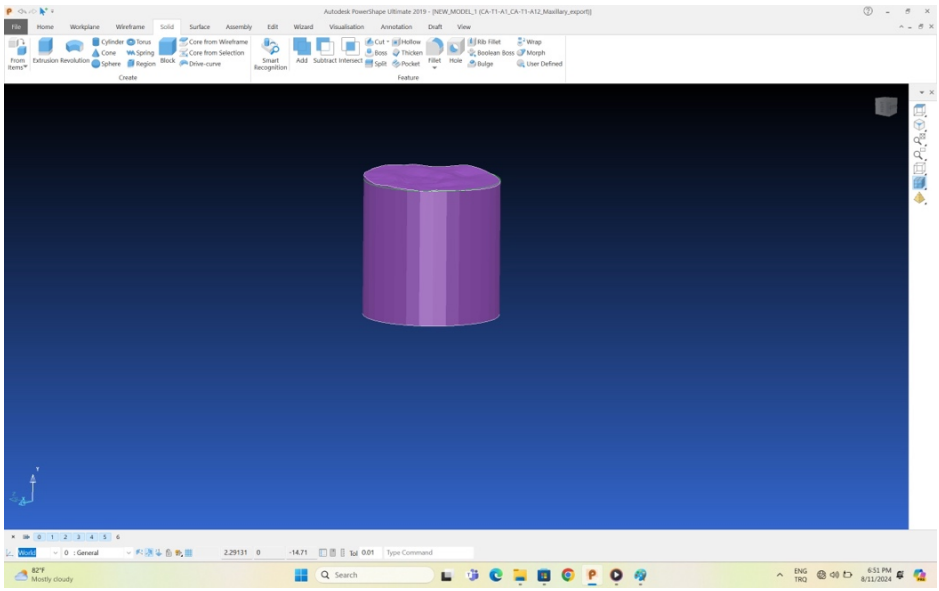
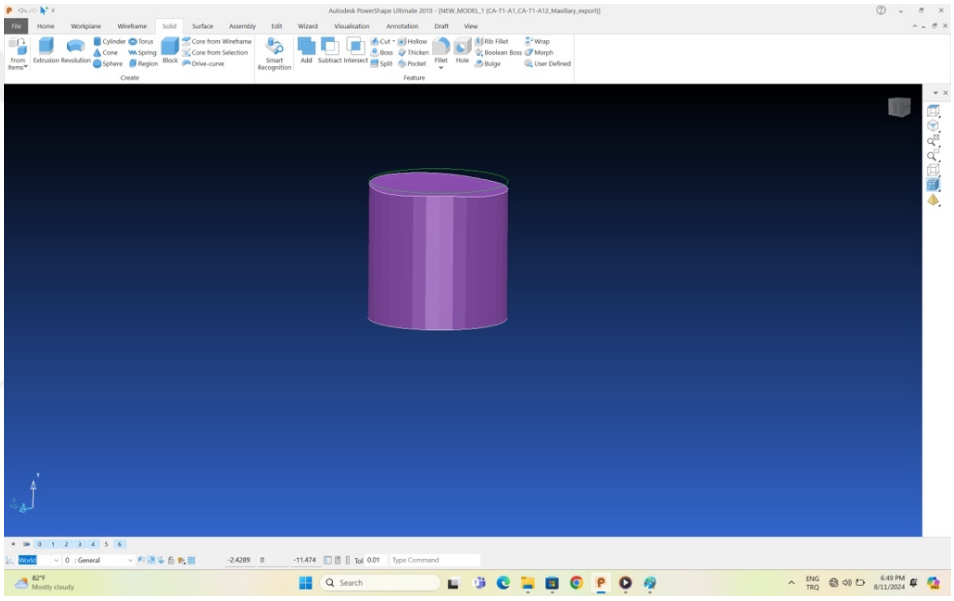


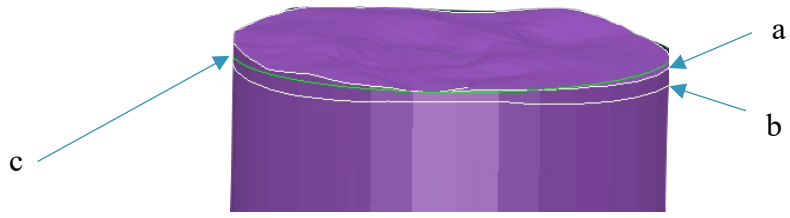
Figure 3.46 and Figure 3.47. Volume calculation with cylindrical sections from the worn area Autodesk Powershape Ultimate 2019 (San Rafael, California, U.S.)

a



b



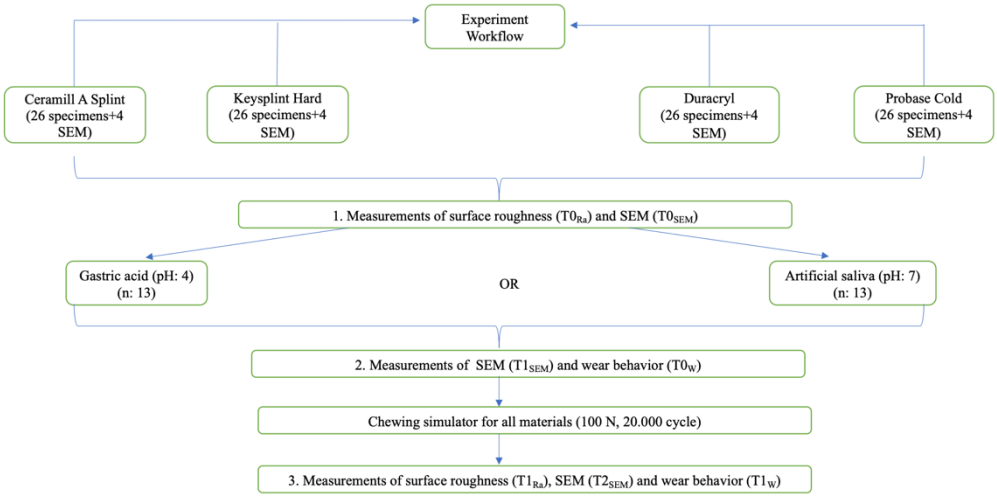


(Figure 3.48., Figure 3.49. and Figure 3.50.) Volume loss calculation before and after the chewing simulator (a: Before chewing simulator image, b: After Chewing simulator image, c: (green line) Before-after Chewing simulator image)

3.12. Measurements

Summary of study design and measurement protocol is shown **(Table 3.5.)**.

Table 3.5. Experiment workflow



Measurements were made for $T0_{SEM}$, $T1_{SEM}$ and $T2_{SEM}$ for scanning electron microscope; $T0_{Ra}$ and $T1_{Ra}$ for surface roughness as well as $T0_{W}$ and $T1_{w}$ for wear behavior.

Baseline (T0) measurements were made before samples were immersed in prepared solutions and demonstrated initial surface roughness and SEM.

T1 measurements were taken after time simulating six months of gastric acid exposure. The exposure time to gastric acid for GERD patients who have reflux for six months equals to 45 hours totally for night time. (for one night it happens approximately 3-15 minutes). Control groups were stored in artificial saliva until the experiment has been done (same time with GERD). Initial wear behavior (T0) and changes in SEM measurements were examined. Then chewing simulator was used, it was checked whether the extra prepared sample broke under increased force. Samples from the entire test group were subjected to the loading of 100N with 20.000 cycle for simulating just nocturnal bruxism. After that, measurements of wear behavior ($T1_{W}$), surface roughness ($T1_{Ra}$) and SEM ($T2_{SEM}$) were examined.

4. RESULTS

4.1. Mechanical Properties

Surface roughness and wear volume were evaluated.

4.1.1. Surface Roughness

Surface roughness (Ra) value was evaluated among four different occlusal splint materials [Duracryl (CH), Probase Cold (CC), Ceramill A (SM), Keysplint Hard (AM)] on baseline (T_0). Surface roughness of samples were taken into the chewing simulator was recorded after being kept in two different solutions (T_1). Ra changes from T_0 to T_1 (ΔRa) were evaluated based on material and solution. Higher Ra value in 'μm' means that the material is rougher.

Prior to the chewing simulation, all group had Ra levels that were clinically acceptable, meaning they did not above the permitted limit value of 0.2 μm for hard surfaces in the oral environment because of the bacterial adhesion [177,178].

Descriptive results on surface roughness of four subgroups before and after the solution exposures and chewing simulator were shown at Table 4.1.

Table 4.1. Descriptive information regarding surface roughness about the groups

Surface Roughness (Ra)		n	Min	Max	x̄	SD	Std. Error of Mean.	Skewness	Kurtosis
T0 Ra	Keysplint Hard (AM) Control Subgroup	13	0.6	2	1.18	0.46	0.13	0.835	-0.387
	Keysplint Hard (AM) GERD Subgroup	13	0.8	2	1.33	0.30	0.08	0.539	1.182
	Ceramill A (SM) Control Subgroup	13	0.5	1.3	0.74	0.21	0.06	1.389	1.288
	Ceramill A (SM) GERD Subgroup	13	0.5	1.5	0.85	0.27	0.08	0.985	1.439
	Duracryl (CH) Control Subgroup	13	0.7	7	3.88	1.88	0.52	-0.423	-0.669
	Duracryl (CH) GERD Subgroup	13	0.6	7	2.68	1.75	0.49	1.196	1.065
	Probase Cold (CC) Control Subgroup	13	0.7	4	1.26	0.86	0.24	1.058	1.019
	Probase Cold (CC) GERD Subgroup	13	0.6	2.5	1.15	0.53	0.15	1.012	1.344
T1 Ra	Keysplint Hard (AM) Control Subgroup	13	0.6	6	2.82	1.33	0.37	0.887	1.462
	Keysplint Hard (AM) GERD Subgroup	13	1.2	4	1.98	0.70	0.20	1.082	1.450
	Ceramill A (SM) Control Subgroup	13	1	4	2.12	0.94	0.26	1.213	1.155
	Ceramill A (SM) GERD Subgroup	13	0.8	2	1.42	0.50	0.14	0.259	-1.018
	Duracryl (CH) Control Subgroup	13	2	10	5.08	2.36	0.65	0.744	-0.212
	Duracryl (CH) GERD Subgroup	13	2	10	5.31	2.18	0.61	0.527	0.533
	Probase Cold (CC) Control Subgroup	13	2	4	2.72	0.84	0.23	0.695	-1.196
	Probase Cold (CC) GERD Subgroup	13	1	3.5	2.04	0.78	0.22	0.324	-0.220
ΔRa	Keysplint Hard (AM) Control Subgroup	13	-0.1	5.1	1.64	1.34	0.37	1.468	0.938
	Keysplint Hard (AM) GERD Subgroup	13	-0.3	2.7	0.65	0.80	0.22	1.428	0.812
	Ceramill A (SM) Control Subgroup	13	0.2	3.5	1.38	1.02	0.28	1.018	0.709
	Ceramill A (SM) GERD Subgroup	13	0.1	1.5	0.57	0.45	0.12	0.933	-0.112
	Duracryl (CH) Control Subgroup	13	-3	5	1.20	2.32	0.64	-0.438	-0.591
	Duracryl (CH) GERD Subgroup	13	-3	8	2.63	2.64	0.73	-0.140	1.325
	Probase Cold (CC) Control Subgroup	13	-2	3	1.45	1.25	0.35	-1.327	1.407
	Probase Cold (CC) GERD Subgroup	13	-0.5	2.3	0.89	0.91	0.25	0.144	-0.956

Descriptive characteristics of the study parameters are seen in Table 1. Since the kurtosis and skewness values of all parameters were between -1.5 and +1.5, their distributions were considered normal and parametric tests were applied in comparisons.

4.1.1.1. Evaluation of initial surface roughness according to different materials and solutions

Table 4.2. Initial surface roughness according to different materials and solutions

	AM	SM	CH	CC	
T0 Ra	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**[1]p**
Control	1.18±0.46	0.74±0.21	3.88±1.88	1.26±0.86	0.001*
GERD	1.33±0.3	0.85±0.27	2.68±1.75	1.15±0.53	0.001*
[2]p	0.323	0.242	0.105	0.685	
[1]Oneway ANOVA Test		[2]Student t test		*p<0.05	

Evaluation of initial surface roughness according to different materials and solutions were shown at Table 4.2. There was a statistically significant difference in initial roughness values between the materials in artificial saliva solution (p:0.001; p<0.05). As a result of the post hoc Tukey HSD test performed to determine significance; Initial surface roughness average of heat-cured acrylic group was found to be significantly higher than AM (p:0.001), SM (p:0.001) and CC (p:0.001) groups (p<0.05). There was no statistically significant difference in the initial surface roughness values among the various materials (p>0.05).

There was a statistically significant difference in initial surface roughness values between the materials in the gastric acid solution (p:0.001; p<0.05). As a result of the post hoc Tukey HSD test performed to determine significance; the initial roughness mean value of the heat-cured acrylic group was found to be significantly higher than AM (p:0.003), CAD-CAM (p:0.001) and autopoymerizing resin (p:0.001) groups (p<0.05). There was no significant difference in terms of initial surface roughness values among other materials (p>0.05).

When AM material is used; there was no significant difference in initial roughness values between artificial saliva and gastric acid (p>0.05).

When SM material is used; there was no statistically significant difference in initial surface roughness values between artificial saliva and gastric acid (p>0.05).

When using CH material; there was no significant difference in initial surface roughness values between artificial saliva and gastric acid (p>0.05).

When CC material is used; there was no significant difference in initial surface roughness values between artificial saliva and gastric acid (p>0.05).

4.1.1.2. Evaluation of final surface roughness according to different materials and solutions

Table 4.3. Final surface roughness according to different materials and solutions

	AM	SM	CH	CC	
T1 Ra	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**[1]p**
Control	2.82±1.33	2.12±0.94	5.08±2.36	2.72±0.84	0.001*
GERD	1.98±0.70	1.42±0.50	5.31±2.18	2.04±0.78	0.001*
[2]p	0.056	0.028*	0.798	0.043*	
[1]Oneway ANOVA Test		[2]Student t test		*p<0.05	

Evaluation of final surface roughness according to different materials and solutions were shown at Table 4.3. There was a statistically significant difference in terms of final surface roughness values between materials in artificial saliva solution (p:0.001; p<0.05). As a result of the post hoc Tukey HSD test performed to determine significance; average final surface roughness of the heat-cured acrylic group was found to be significantly higher than AM (p:0.002), SM (p:0.001) and CC (p:0.001) groups (p<0.05). There was no significant difference in terms of final surface roughness values among other materials (p>0.05).

There was a statistically significant difference in terms of final surface roughness values between materials in gastric acid solution (p:0.001; p<0.05). As a result of the post hoc Tukey HSD test performed to determine significance; average final surface roughness

of CH group was found to be significantly higher than the AM (p:0.001), SM (p:0.001) and CC (p:0.001) groups ($p<0.05$). There was no significant difference in terms of final surface roughness values among other materials ($p>0.05$).

When AM material is used; although final surface roughness of artificial saliva solution was higher than that of gastric acid, this difference was close to significance but not statistically significant ($p>0.05$).

When SM material is used; final surface roughness of artificial saliva solution was statistically significantly higher than that of gastric acid (p:0.028; $p<0.05$).

When using CH material; there was no significant difference in final surface roughness values between artificial saliva and gastric acid ($p>0.05$).

When CC material is used; final surface roughness of the artificial saliva solution was statistically significantly higher than that of gastric acid (p:0.043; $p<0.05$).

4.1.1.3. Evaluation of final surface roughness compared to the initial one for different materials and solutions

Table 4.4. Final surface roughness compared to the initial for different materials and solutions

		AM	SM	CH	CC
Solution	**Ra**	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**
Artificial Saliva	**T0**	1.18±0.46	0.74±0.21	3.88±1.88	1.26±0.86
	T1	2.82±1.33	2.12±0.94	5.08±2.36	2.72±0.84
	p	0.001*	0.001*	0.087	0.001*
Gastric Acid	**T0**	1.33±0.30	0.85±0.27	2.68±1.75	1.15±0.53
	T1	1.98±0.70	1.42±0.50	5.31±2.18	2.04±0.78
	p	0.013*	0.001*	0.004*	0.004*
Paired samples t test		**p<0.05*			

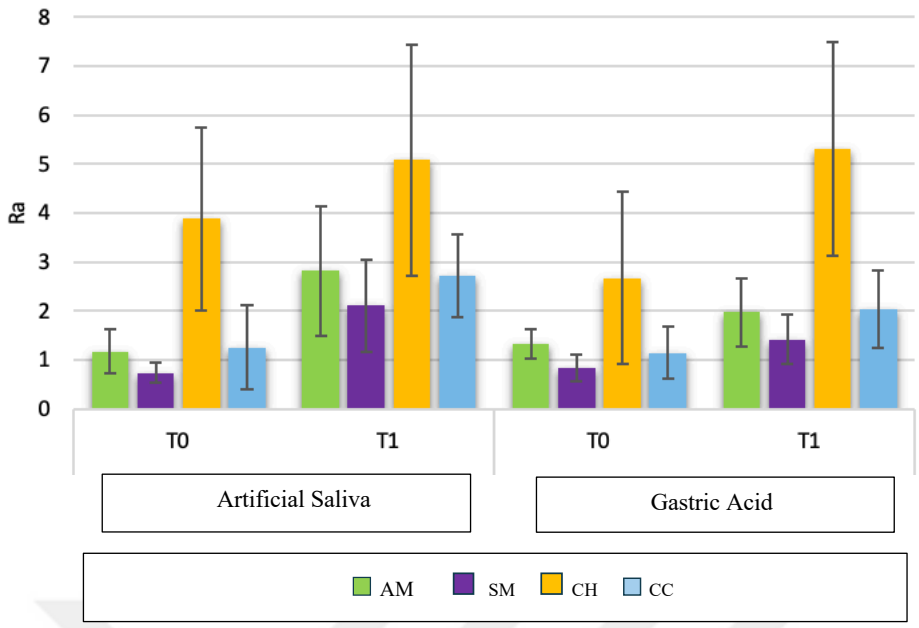


Figure 4.1. Initial and final surface roughness evaluations on different materials and solutions

Evaluation of final surface roughness compared to the initial one for different materials and solutions were shown at Table 4.4. and initial and final surface roughness evaluations on different materials and solutions were shown in Figure 4.1.

In the artificial saliva AM group; the increase in final surface roughness compared to initial surface roughness was statistically significant (p:0.001; $p<0.05$).

In artificial saliva SM group; the increase in final surface roughness compared to initial surface roughness was statistically significant (p:0.001; $p<0.05$).

In the artificial saliva CH group; there was no statistically significant change in the final surface roughness compared to the initial surface roughness (p:0.087; $p>0.05$).

In artificial saliva CC group; the increase in final surface roughness compared to initial surface roughness was statistically significant (p:0.001; $p<0.05$).

In gastric acid AM group; increase in final surface roughness compared to initial surface roughness was statistically significant (p:0.013; $p<0.05$).

In gastric acid SM group; the increase in final surface roughness compared to the initial surface roughness was statistically significant (p:0.001; $p<0.05$).

In gastric acid CH group; the increase in final surface roughness compared to the initial surface roughness was statistically significant (p:0.004; $p<0.05$).

In gastric acid CC group; increase in final surface roughness compared to initial surface roughness was statistically significant (p:0.004; $p<0.05$).

4.1.1.4. Evaluation of the joint effect of material and solution on roughness change

Table 4.5. The joint effect of material and solution on roughness change

Δ Ra	**Type III Sum of Squares**	**df**	**Mean Square**	F	p
Material	13.69	3	4.563	1.985	0.121
Solution	1.408	1	1.408	0.612	0.436
Material*Solution	24.589	3	8.196	3.565	0.017*
Two-way ANOVA Test	**p<0.05*				

Evaluation of joint effect of material and solution on roughness change were shown at Table 4.5. There was no statistically significant difference in terms of surface roughness change between materials (p>0.05).

There was no statistically significant difference in terms of surface roughness change between the solutions (p>0.05).

The joint effect of material and solution on surface roughness change was statistically significant (p:0.017; p<0.05).

Detailed evaluations of mentioned results are seen in Table 4.6.

4.1.1.5. Detailed evaluation of the joint effect of material and solution on roughness change

Table 4.6. The joint effect of material and solution on roughness change in detailed

	AM	SM	CH	CC	
Δ Ra	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**[1]p**
Control	1.64±1.34 [Aa]	1.38±1.02 [Aa]	1.20±2.32 [Aa]	1.45±1.25 [Aa]	0.912
GERD	0.65±0.80 [Ba]	0.57±0.45 [Ba]	2.63±2.64 [Ab]	0.89±0.91 [Aa]	0.002*
[2]p	0.031*	0.015*	0.155	0.202	
Two-way ANOVA Test		**p<0.05*			

Different lowercase letters indicate differences between materials.
Different capital letters indicate differences between solutions.

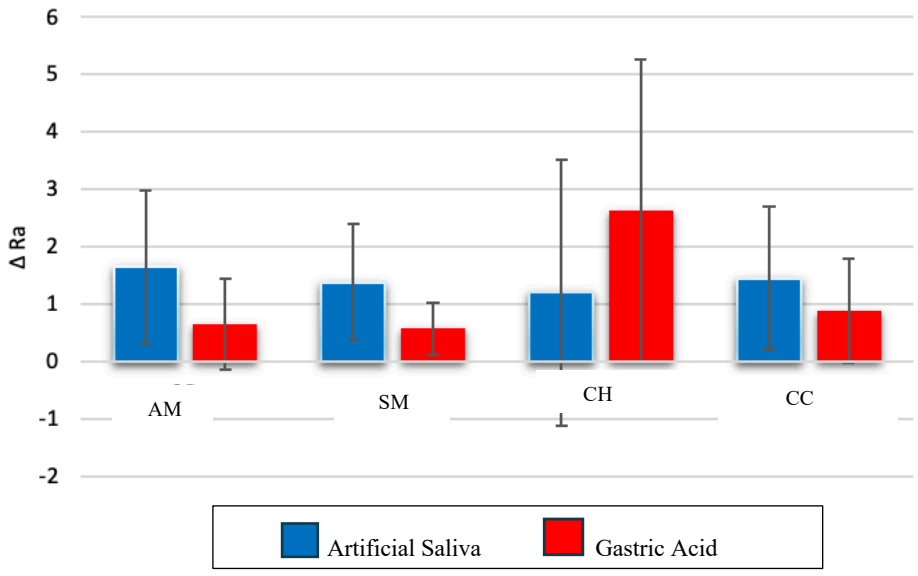


Figure 4.2. Surface roughness change evaluations on different materials and solutions

Detailed evaluation of joint effect of material and solution on roughness change were shown at Table 4.6. and surface roughness change evaluations on different materials and solutions were shown at Figure 4.2.

When effects of solutions on materials are evaluated;

There was no statistically significant difference between materials in artificial saliva solution in terms of final surface roughness changes compared to beginning ($p:0.912$; $p>0.05$).

There was a statistically significant difference between materials in terms of final surface roughness changes compared to beginning in gastric acid solution ($p:0.002$; $p<0.05$). As a result of post hoc Tukey HSD test performed to determine significance; amount of surface roughness increase in CH group was found to be significantly higher than the AM ($p:0.006$), SM ($p:0.004$) and CC ($p:0.020$) groups ($p<0.05$). There was no significant difference in terms of surface roughness changes among other materials ($p>0.05$).

When the effects of solutions on materials are evaluated;

When AM material is used; the amount of surface roughness increase at end of the artificial saliva solution compared to beginning was statistically significantly higher than that of gastric acid ($p:0.031$; $p<0.05$).

When SM material is used; the amount of increase in surface roughness at the end of the artificial saliva solution compared to beginning was statistically significantly higher than that of gastric acid ($p:0.015$; $p<0.05$).

When using CH material; there was no statistically significant difference between artificial saliva and gastric acid in terms of the increase in surface roughness at the end compared to the beginning ($p>0.05$).

When CC material is used; there was no statistically significant difference between artificial saliva and gastric acid in terms of increase in surface roughness at the end compared to the beginning ($p>0.05$).

4.1.2. Descriptive information regarding volumetric and linear change about the groups

Table 4.7. Volumetric and linear change about the groups

	Group	N	Min	Max	x̄	SD	SE	Skewness	Kurtosis
ΔV (mm^3)	Keysplint Hard (AM) Control Subgroup	13	-2.85	0.45	-0.721	0.94	0.26	-1.050	0.984
Volume	Keysplint Hard (AM) GERD Subgroup	13	-2.44	0.2	-1.032	0.83	0.23	-0.380	-0.876
	Ceramill A (SM) Control Subgroup	13	-1	0.37	-0.110	0.35	0.10	-1.142	1.217
	Ceramill A (SM) GERD Subgroup	13	-1.78	0.6	-0.201	0.57	0.16	-1.132	1.071
	Duracryl (CH) Control Subgroup	13	-3.2	2.8	-0.162	1.76	0.49	-0.077	-0.121
	Duracryl (CH) GERD Subgroup	13	-2.13	1.32	-0.391	1.15	0.32	-0.114	-1.297
	Probase Cold (CC) Control Subgroup	13	-2.79	2.2	-0.328	1.68	0.47	0.087	-1.423
	Probase Cold (CC) GERD Subgroup	13	-6.18	0.9	-2.277	2.72	0.75	-0.299	-1.889
ΔV (μm)	Keysplint Hard (AM) Control Subgroup	13	-1.21	2.12	-0.124	0.78	0.22	1.482	1.183
Linear	Keysplint Hard (AM) GERD Subgroup	13	-0.21	1.22	0.174	0.49	0.14	1.499	1.020
	Ceramill A (SM) Control Subgroup	13	-1.87	0.61	-0.266	0.63	0.17	-1.468	1.369
	Ceramill A (SM) GERD Subgroup	13	-0.65	0.87	-0.031	0.46	0.13	0.539	-0.309
	Duracryl (CH) Control Subgroup	13	-1.71	0.4	-0.261	0.54	0.15	-1.285	1.070
	Duracryl (CH) GERD Subgroup	13	-0.82	0.13	-0.199	0.25	0.07	-1.404	1.194
	Probase Cold (CC) Control Subgroup	13	-1.25	0.5	-0.063	0.44	0.12	-1.409	1.290
	Probase Cold (CC) GERD Subgroup	13	-1.11	1.29	-0.067	0.62	0.17	0.769	1.067

Table 4.7 displays the descriptive features of the study parameters. Since the kurtosis and skewness values for all volumetric and linear changes were between -1.5 and +1.5, their distributions were considered normal and parametric tests were applied in comparisons.

4.1.2.1. Evaluation of the joint effect of material and solution on volumetric change

Table 4.8. Joint effect of material and solution on volumetric change

ΔV (mm³)	**Type III Sum of Squares**	**df**	**Mean Square**	F	p
Material	22.363	3	7.454	3.577	0.017*
Solution	10.817	1	10.817	5.191	0.025*
Material*Solution	14.903	3	4.968	2.384	0.074
Two-way ANOVA Test	**p<0.05*				

Evaluation of the joint effect of material and solution on volumetric change were shown at Table 4.8. There was a statistically significant difference between materials in terms of volumetric change (p:0.017; p<0.05).

There was a statistically significant difference in terms of volumetric change between solutions (p:0.025; p<0.05).

The joint effect of material and solution on volumetric change was not statistically significant (p:0.074; p>0.05).

Detailed evaluations of the mentioned results are seen in Table 4.9.

4.1.2.2. Detailed evaluation of the joint effect of material and solution on volumetric change

Table 4.9. Joint effect of material and solution on volumetric change in detail

	AM	SM	CH	CC	
ΔV (mm³)	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**[1]p**
Control	-0.721±0.94^{Aa}	-0.11±0.35^{Aa}	-0.162±1.76^{Aa}	-0.328±1.68^{Aa}	.635
GERD	-1.032±0.83^{Aab}	-0.201±0.57^{Aa}	-0.391±1.15^{Aa}	-2.277±2.72^{Bb}	0.006*
[2]p	0.380	0.630	0.698	0.040*	
Two-way ANOVA Test		*p<0.05			

Different lowercase letters indicate differences between materials.
Different capital letters indicate differences between solutions.

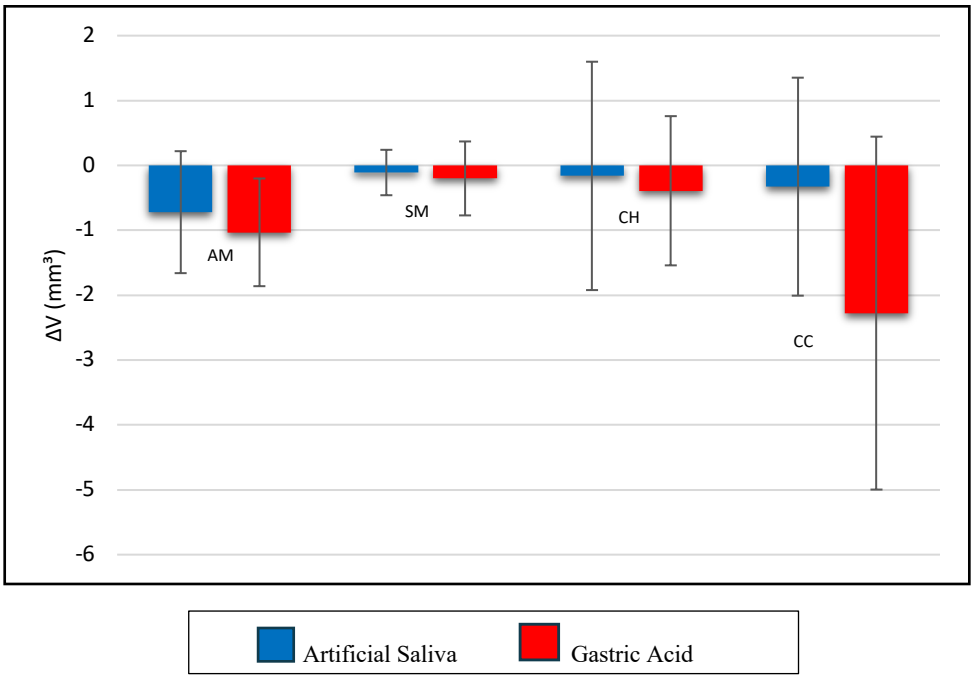


Figure 4.3. Volumetric change in different solutions

Detailed evaluation of the joint effect of material and solution on volumetric change were shown at Table 4.9. and volumetric change in different solutions were shown at Figure 4.3. When the effects of solutions on materials are evaluated;

There was no statistically significant difference in the amount of volumetric change between the materials in the artificial saliva solution (p:0.635; $p>0.05$).

There was a statistically significant difference in terms of volumetric change between materials in gastric acid solution (p:0.006; $p<0.05$). As a result of post hoc Tukey HSD test performed to determine significance; the volumetric loss amount of the autopolymerizing resin group was found to be significantly higher than the SM (p:0.007) and CH (p:0.018) groups ($p<0.05$). There was no significant difference in terms of volumetric change among other materials ($p>0.05$).

When the effects of solutions on materials are evaluated;

When AM material is used; there was no statistically significant difference in terms of volumetric loss between artificial saliva and gastric acid ($p>0.05$).

When SM material is used; there was no statistically significant difference in terms of volumetric loss between artificial saliva and gastric acid ($p>0.05$).

When using CH material; there was no statistically significant difference in terms of volumetric loss between artificial saliva and gastric acid ($p>0.05$).

When CC material is used; the volume loss of gastric acid solution was statistically significantly higher than that of artificial saliva solution (p:0.040; $p<0.05$).

4.1.2.3. Evaluation of the joint effect of material and solution on linear change about the groups

Table 4.10. Joint effect of material and solution on linear change about the groups

ΔV (μm)	**Type III Sum of Squares**	**df**	**Mean Square**	**F**	**p**
Material	0.936	3	0.312	1.047	0.375
Solution	0.567	1	0.567	1.904	0.171
Material*Solution	0.392	3	0.131	0.438	0.726
Two-way ANOVA Test					

Table 4.10. displays the evaluation of the material and solution's combined impact on the groups' linear change. Regarding the linear change between the materials, there was no statistically significant difference (p:0.375; p>0.05). There was no statistically significant difference in terms of linear change between the solutions (p:0.171; p>0.05).

The joint effect of material and solution on linear change was also not statistically significant (p:0.726; p>0.05).

Detailed evaluations of the mentioned results are seen in Table 4.11.

4.1.2.4. Detailed evaluation of the joint effect of material and solution on linear change

Table 4.11. The joint effect of material and solution on linear change in detail

	AM	SM	CH	CC	
ΔV (µm)	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**x̄ ±SD**	**[1]p**
Control	-0.124±0.78	-0.266±0.63	-0.261±0.54	-0.063±0.44	0.784
GERD	0.174±0.49	-0.031±0.46	-0.199±0.25	-0.067±0.62	0.261
[2]p	0.256	0.287	0.711	0.988	
Two-way ANOVA Test					

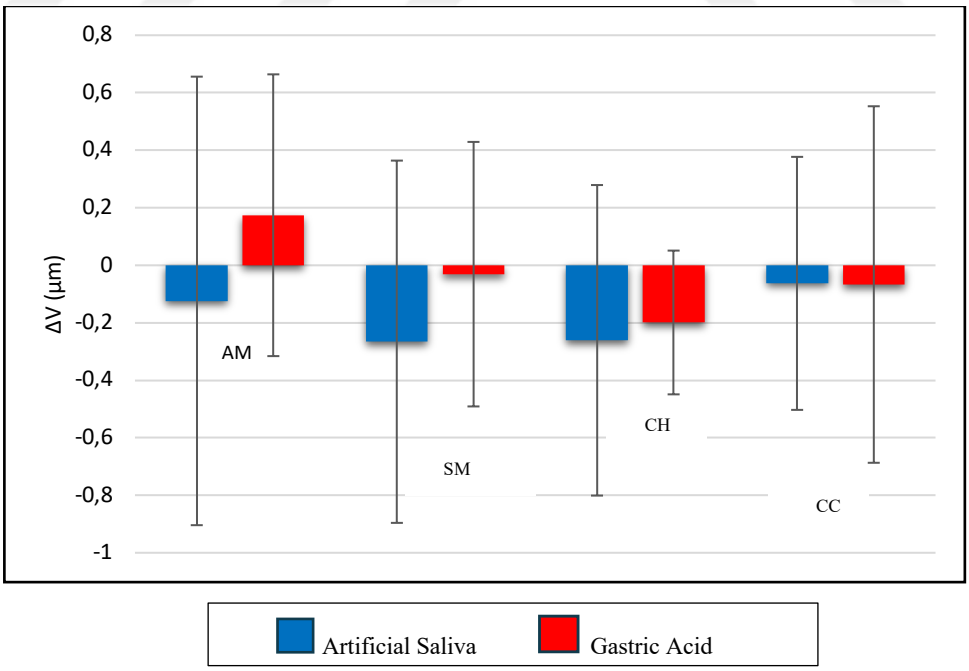


Figure 4.4. Linear change in different solutions

Table 4.11. displays a comprehensive analysis of the combined impact of material and solution on linear change and linear change were displayed in several solutions in Figure 4.4.

The amount of linear change between the materials in the artificial saliva solution did not differ statistically significantly (p:0.784; p>0.05).

There was no statistically significant difference in the amount of linear change between the materials in gastric acid solution (p:0.261; p>0.05).

When AM material is used; there was no statistically significant difference in terms of linear loss amounts between artificial saliva and gastric acid (p>0.05).

When SM material is used; there was no statistically significant difference in terms of linear loss amounts between artificial saliva and gastric acid (p>0.05).

When using CH material; there was no statistically significant difference in terms of linear loss amounts between artificial saliva and gastric acid (p>0.05).

When CC material is used; there was no statistically significant difference in terms of linear loss amounts between artificial saliva and gastric acid (p>0.05).

4.1.3. Correlation of surface roughness change (ΔRa) and volumetric change (ΔV (mm³)) in different materials and solutions

Table 4.12. Correlation of ΔRa and ΔV (mm³) in different materials and solutions

ΔRa- ΔV (mm³)		
Group	**r**	**p**
Saliva 3D	0.208	0.496
GERD 3D	0.197	0.520
Saliva CAD-CAM	0.288	0.340
GERD CAD-CAM	-0.071	0.817
Saliva heat-cured acrylic	-0.031	0.920
GERD heat-cured acrylic	-0.236	0.438
Saliva autopolymerizing resin	0.294	0.330
GERD autopolymerizing resin	0.095	0.759
Pearson correlation analysis		

Upon examining each group independently, no statistically significant correlation was seen between the change in surface roughness and the volumetric change in any group ($p>0.05$) (Table 4.12.).

4.1.4. Correlation of linear change (ΔV (µm)) and volumetric change (ΔV (mm³)) in different materials and solutions

Table 4.13. Correlation of linear change (ΔV (µm)) and volumetric change (ΔV (mm³)) in different materials and solutions

ΔV (µm) - ΔV (mm³)		
Group	**r**	**p**
Saliva 3D	0.188	0.538
GERD 3D	0.492	0.088
Saliva CAD-CAM	0.414	0.160
GERD CAD-CAM	0.498	0.084
Saliva heat-cured acrylic	0.147	0.632
GERD heat-cured acrylic	0.201	0.511
Saliva autopolymerizing resin	0.616	0.025*
GERD autopolymerizing resin	0.737	0.004*
Pearson correlation analysis	**p<0.05*	

In the saliva autopolymerizing resin group; there was a positive, good level (61.6%) and statistically significant relationship between volumetric change and linear change (p:0.025; p<0.05).

In the GERD autopolymerizing resin group; there was a positive, good level (73.7%) and statistically significant relationship between volumetric change and linear change (p:0.004; p<0.05).

There was no statistically significant relationship between volumetric change and amount of linear change in the other groups (p>0.05) (Table 4.13.).

4.2. Scanning Electron Microscope (SEM) Evaluations

Before solution exposure ($T0_{SEM}$), after solution exposure ($T1_{SEM}$) and after chewing simulator ($T2_{SEM}$) SEM images of materials for two subgroups (GERD and control) were shown in figures from Figure 4.5. to Figure 4.8.

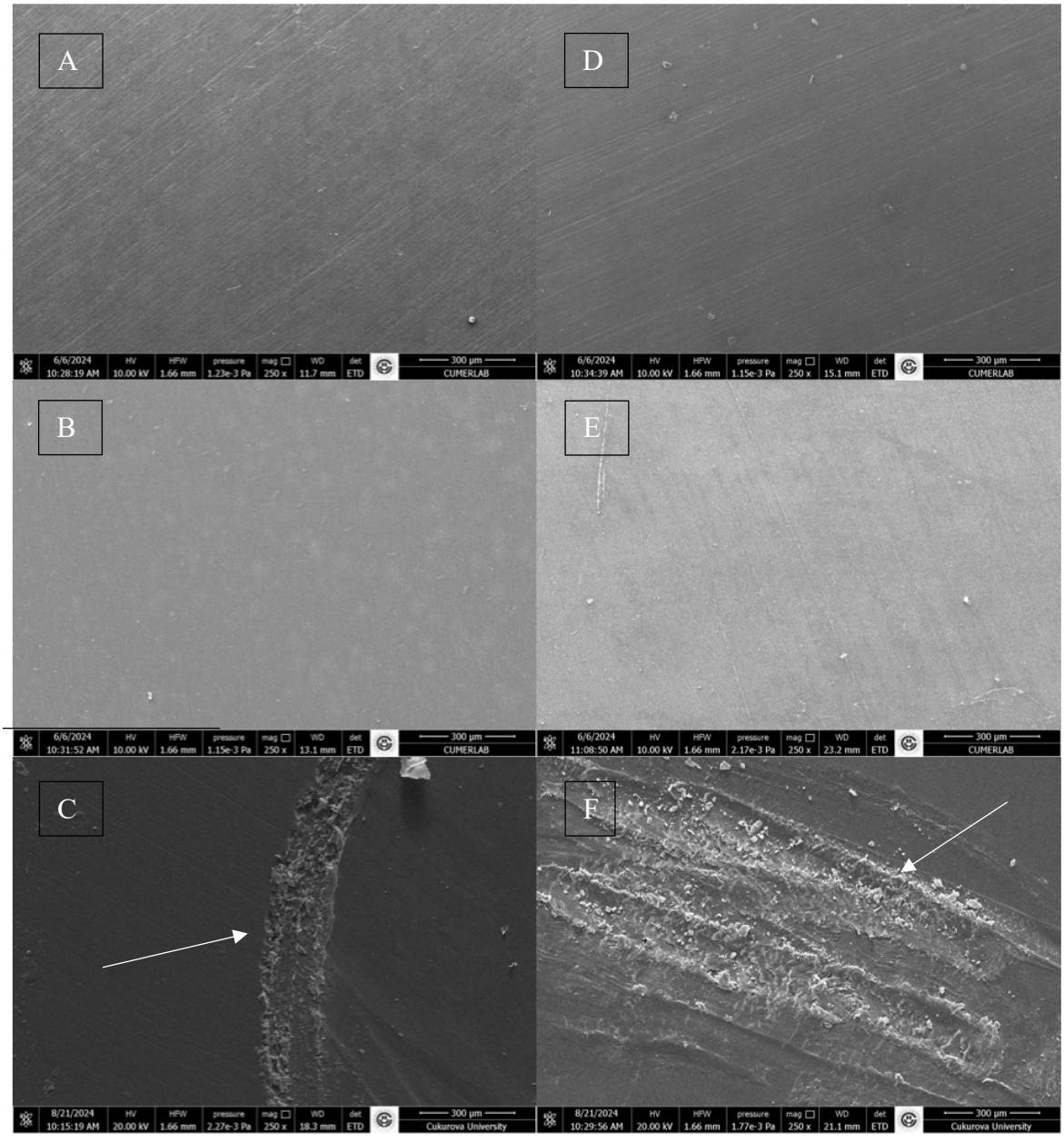


Figure 4.5. SEM images of AM materials with x250 original magnification.
A: Before acid exposure, B: After acid exposure, C: After chewing simulator of GERD group;
D: Before saliva exposure, E: After saliva exposure, F: After chewing simulator of control group.

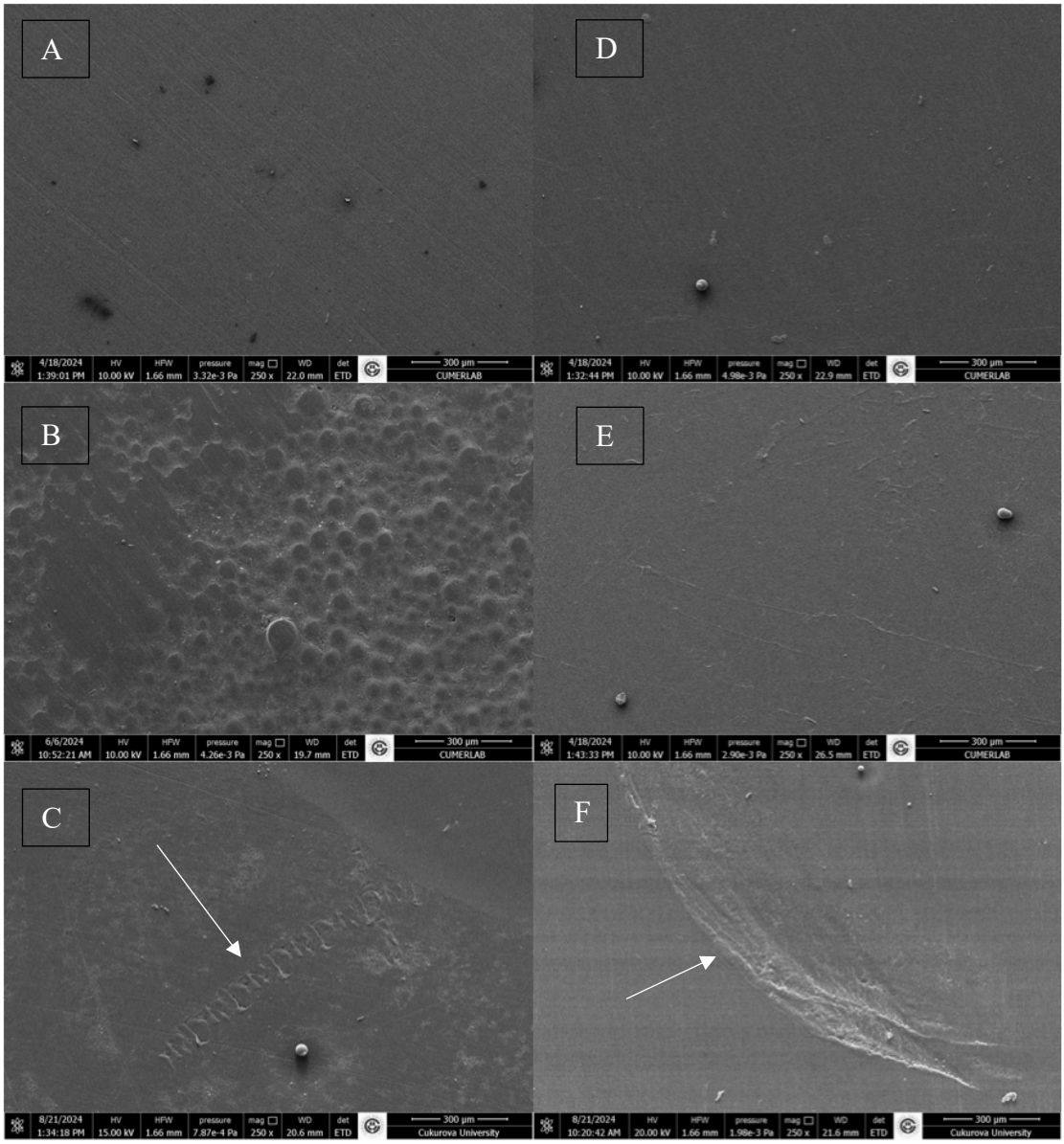


Figure 4.6. SEM images of SM materials with x250 original magnification.
A: Before acid exposure, B: After acid exposure, C: After chewing simulator of GERD group;
D: Before saliva exposure, E: After saliva exposure, F: After chewing simulator of control group.

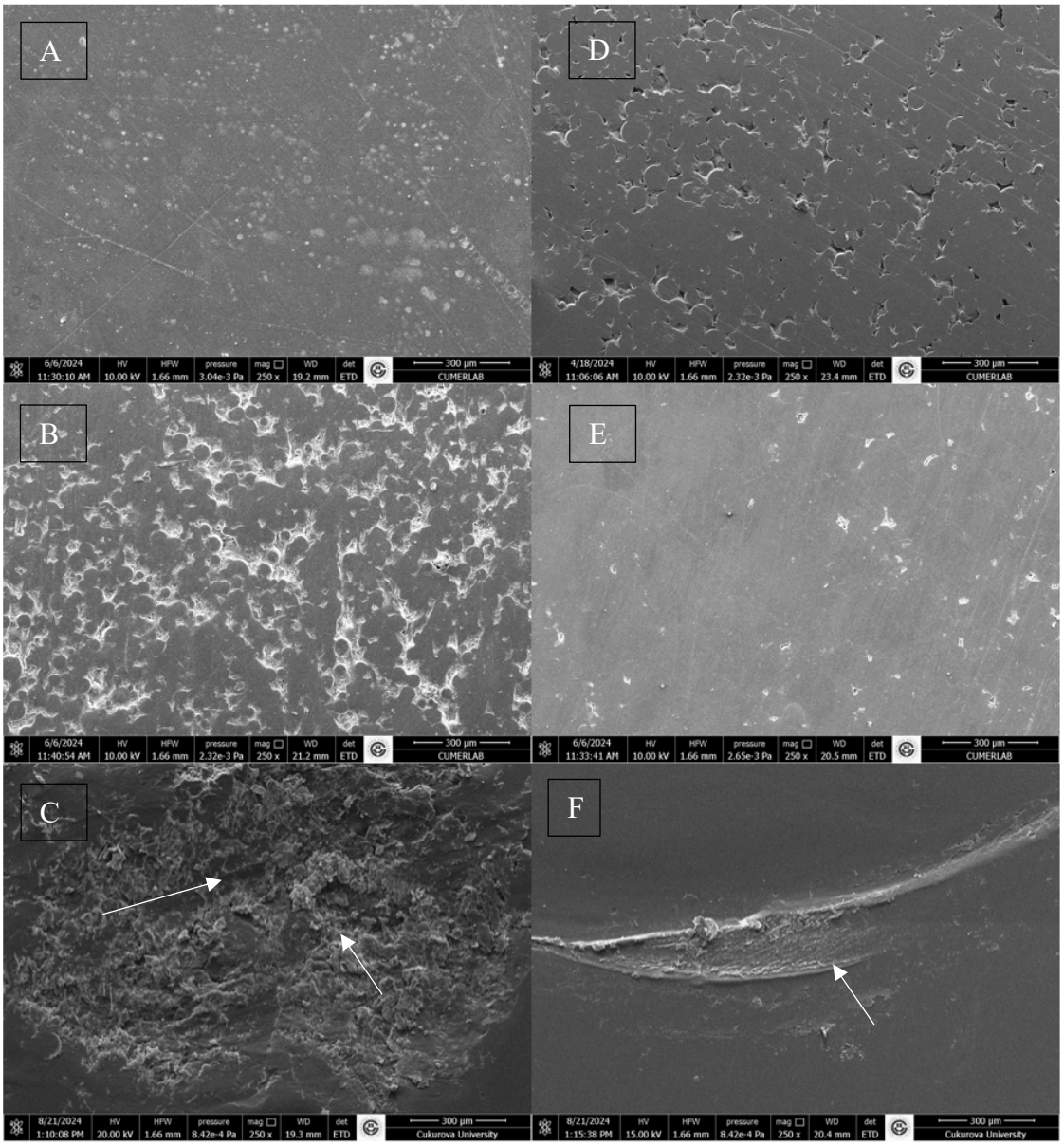


Figure 4.7. SEM images of CH materials with x250 original magnification.
A: Before acid exposure, B: After acid exposure, C: After chewing simulator of GERD group;
D: Before saliva exposure, E: After saliva exposure, F: After chewing simulator of control group.

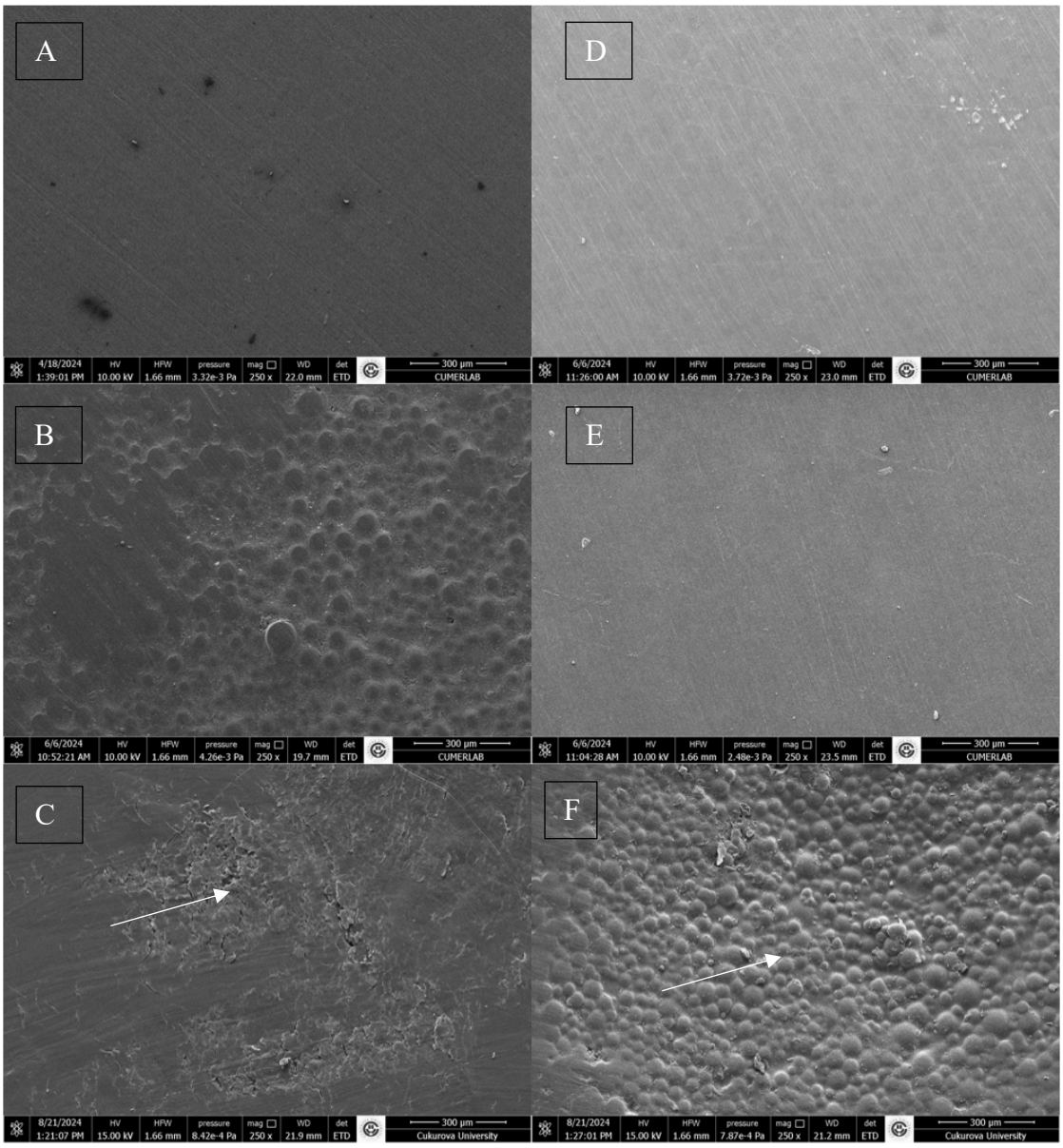


Figure 4.8. SEM images of CC materials with x250 original magnification.
A: Before acid exposure, B: After acid exposure, C: After chewing simulator of GERD group;
D: Before saliva exposure, E: After saliva exposure, F: After chewing simulator of control group.

4.3. 3D Optical Profilometer Evaluations

Before (initial) and after (after chewing simulator) 3D optical profilometer images of four subgroups were shown at Figure 4.9. and Figure 4.10.

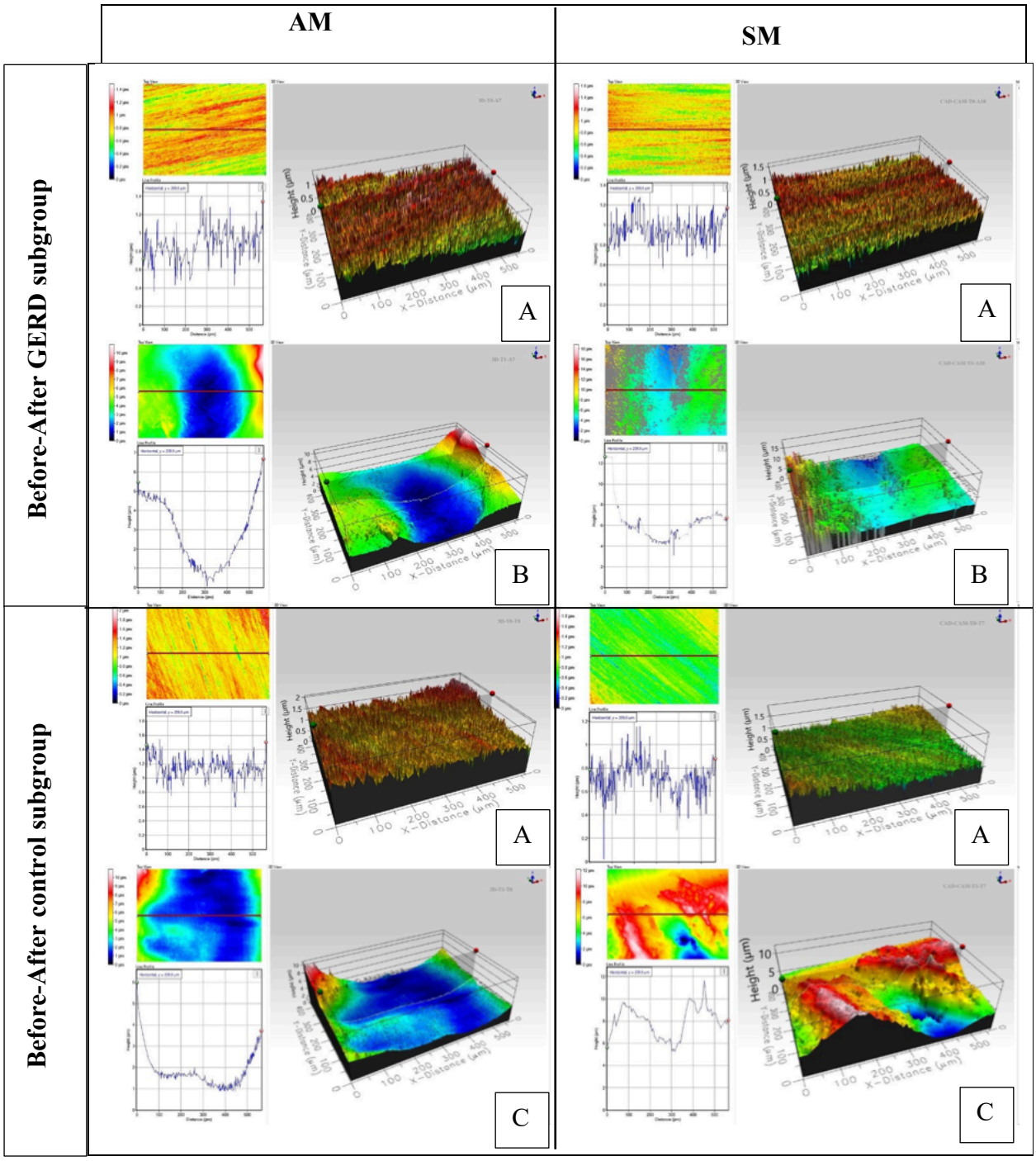


Figure 4.9. 3D topographic images of AM and SM groups taken by 3D optical profilometer. A: before solution exposure. B: after gastric acid exposure and chewing simulator application. C: after artificial saliva exposure and chewing simulator application.

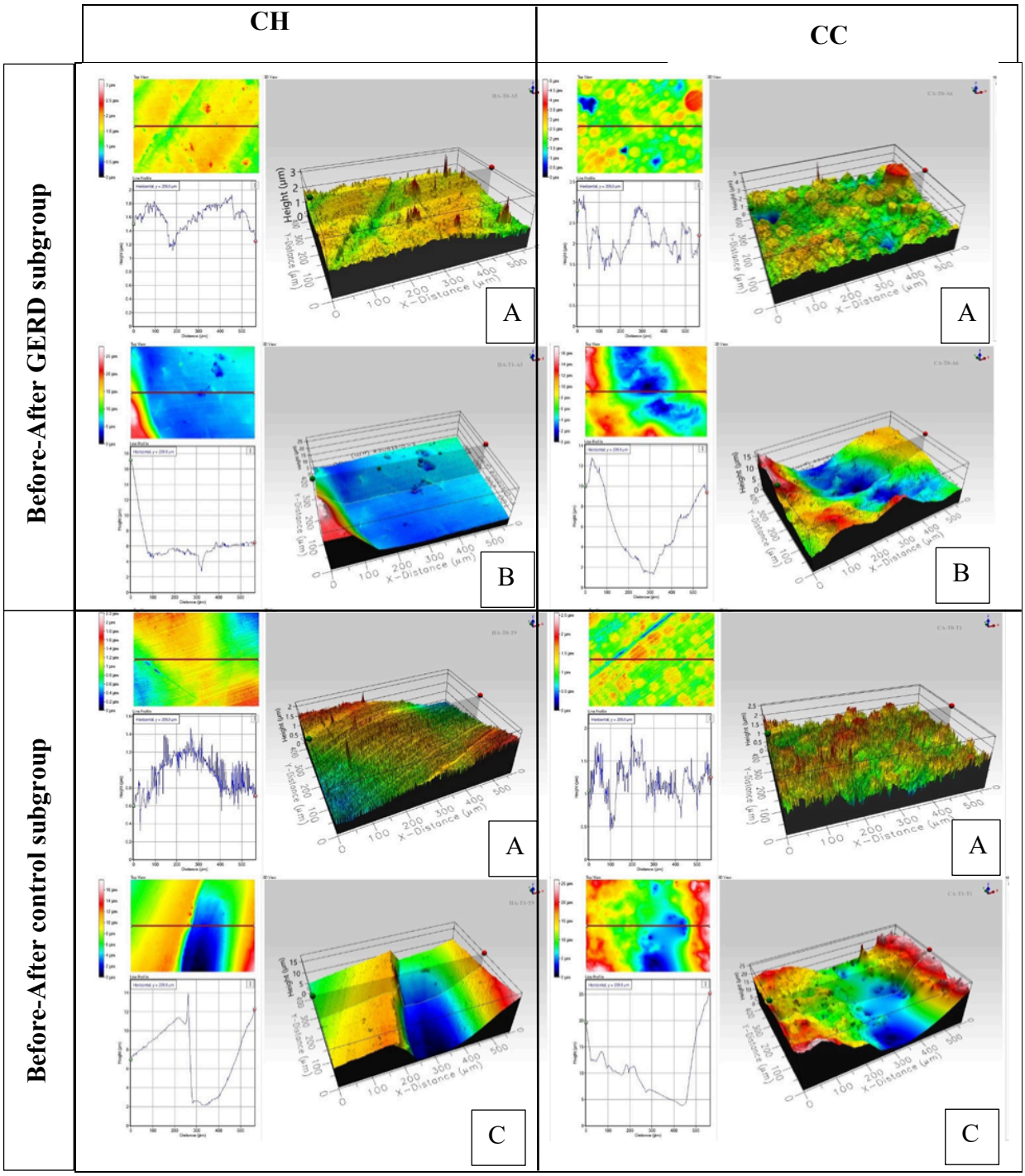


Figure 4.10. 3D topographic images of CH and CC groups taken by 3D optical profilometer. A: before solution exposure. B: after gastric acid exposure and chewing simulator application. C: after artificial saliva exposure and chewing simulator application.

The study's findings were statistically analyzed using the IBM SPSS Statistics 22 program. Shapiro Wilks, Kolmogorov-Smirnov, kurtosis, and skewness values were used to evaluate the study data in order to determine whether the parameters were appropriate for a normal distribution. The outcomes demonstrated that a normal distribution is in fact displayed by the parameters. Post hoc two-way ANOVA and Tukey HSD tests were used to evaluate the combined effect of material and solution on surface roughness change (ΔRa), linear change (ΔV (μm)), and volumetric change (ΔV (mm^3)). In order to compare materials in roughness evaluations at T0 and T1, Oneway ANOVA (post hoc Tukey Test) was utilized, and Student's t test was used to compare solutions. Pearson correlation analysis was used to evaluate correlations between the parameters. Significance level was set at $p<0.05$.

5. DISCUSSION

Patient-related parameters including parafunctional habits, thickness of the masseter muscle, dental status, age, and gender, as well as material-related aspects like microstructural, physical, and mechanical qualities, are crucial for determining the causes and controlling factors of wear [179–183].

Bruxism which is the parafunctional behavior may be a factor in excessive wear. Comparing non-bruxists to bruxists, Lambrechts et al.[184] reported that the molar enamel attrition rate in the former group is as low as 20–40 μm annually. According to Xhonga [185], wear rates in bruxist patients might reach 220-243 μm. When comparing the biting strength of some bruxist patients to non-bruxists, Gibbs et al.[186] claimed that it might be up to six times stronger. Parafunctional chewing forces can reach range from 450-650 N [187], with a mean of 380 N [188]. A young, healthy person's maximal biting force is 654 N for men as well as 464 N for women, according to Pizolato et al[189]. The occlusal force employed during mastication and swallowing is approximately 40% of the maximum biting force, according to Gibbs et al.[190]. This kind of process is more likely to be produced by a parafunction than a function, as the period and force of the bite in bruxism are significantly greater than throughout functional activity.

There are two unique circadian manifestations of bruxism: nocturnal bruxism, which occurs during sleep, and diurnal bruxism, which occurs during wakefulness [1]. Among the possible adverse effects of sleep bruxism that have been documented in the literature include headaches upon waking, complaints of masticatory muscle and TMD [2,3].

Occlusal splint therapy, occlusal adjustment, medication, acupuncture, surgery, physical therapy, massage therapy, biofeedback, cognitive-behavioral therapy, and stress management are some of the suggested treatment methods for TMD[4]. Regarded as the gold standard for managing temporomandibular joint and muscle issues, occlusal splints are successful in 70–90% of instances[4,191,192]. Occlusal splints are used not only as a method for treatment of TMD, but also to manage bruxism. It is explained that in individuals with parafunctional habits, the way in which occlusal splints work is to cause muscle relaxation by interfering with the neuromuscular reflex contraction cycle[193]. They also help to realign the condyles and jaws into a centric connection, disperse the occlusal loads placed on the teeth, and shield them from wear and movement brought on by clenching and grinding[56].

Splints, whether hard or soft, are used for protection against wear and fracture of various restorations as well as the patients' teeth. Compared to their soft counterparts, there appear to be some benefits associated with the hard acrylic occlusal splints. Hard splints can be adjusted to the occlusal surface and provide a more stable and retentive fit overall. They are easier to clean and, if necessary, repair since they have reduced porosity. The material selection is crucial, particularly when therapy is intended to last for an extended length of time, even if the abilities and knowledge of the physician are paramount in developing and creating the ideal splint for each unique patient. When protecting the restorations and minimizing tooth wear, hard splints are advised[194]. For this reason, we used hard occlusal splints in our study.

Wear resistance of occlusal splint materials have clinical signficance. Wear facets can occur in materials with insufficient wear resistance, which compromises occlusal contact stability and shortens the appliance's lifespan [133,195].

Occlusal splints are exposed to load up to 978 N and can be worn for an extended period (6–12 months) for management of bruxism[179,196]. Therefore, in this study, wear and surface roughness of materials after 6 months of splint use for SB were evaluated as an in vitro study.

GERD is a prevalent upper gastrointestinal tract disease. Heartburn is the most typical symptom of the disease, occurring once a week or more. The condition can affect anywhere from a small percentage of adults to over 30% of them, depending on the location of the world [197].

Research has indicated a positive association between GERD and sleep bruxism[2,14].

Li et al.[15] evaluated connection between length of GERD and intensity of bruxism symptoms. The findings show a significant correlation between the development of GERD and an elevated risk of all forms of bruxism. Moreover, there is a connection between the duration of GERD and the degree of bruxism symptoms. Moreover, a portion of the relationship is mediated by anxiety, depression, and insufficient sleep[15].

Proton pump inhibitors, the initial step for treatment for GERD, were found to slightly decrease episodes of sleep bruxism in 12 Japanese patients in a recent randomized cross-over investigation[198].

Studies carried out on adults in good physical condition indicate that oesophageal acidification elevates occurrences of bruxism during sleep, as well as arousals and swallowing during sleep[16].

Furthermore, during awake hours, experimental oesophageal acidification raises masseter muscle activity, particularly involuntary muscular activity. This suggests that there might be a connection between GERD and awake bruxism[199].

People are affected by reflux differently depending on the time of day. While some patients have reflux only at night or sporadically throughout the day, others describe reflux occurring continuously during the day[145]. GERD during the night is linked to more severe symptoms than GERD during the day, such as a reduced quality of life, decreased productivity at work, extraesophageal symptoms, excessive daytime sleepiness, sleep disturbances, and esophageal problems such as stricture, Barret's esophagus, or esophageal adenocarcinoma[200].

Although up to 59% of people report having heartburn monthly[201,202], up to 20% report having symptoms every week[202], and 18% take prescription medications to treat their symptoms, estimates of the prevalence of GERD range from 6% to 10%[201,203]. Especially, GERD symptoms occured when patients lay down to sleep at night (69%)[200]. Moreover, According to reports, 23% of Turkish adults have GERD, a rate comparable to that of Western nations[204]. The worldwide prevalence of bruxism (sleep and awake) is 22.22% [205]. 21% of people worldwide suffer from sleep bruxism.

According to Mengatto et al.[14], association between SB and GERD was evaluated and results of the study show that participants with GERD had higher prevalence of SB (73.7%) than did those in good health (23.1%). When bruxism and GERD are evaluated together, treatment of the condition and material selection become even more important.

According to Miyawaki et al.[206], participants with SB experienced more GERD episodes, and 60% of SB episodes happened during reflux. When proton-pump inhibitor medication was administered, there was a decrease in SB episodes when compared to placebo and control groups[198,206].

There is no study in the literature yet on the suitable splint material that should be used in GERD patients. The findings of current study contribute to knowledge about the wear and surface roughness of splint materials, suggesting that it is possible to make a splint selection specifically tailored to needs of patients with GERD.

5.1. Material and Method

The wear of occlusal splints is replicated and measured using a variety of methods, including different scanning strategies, load, precision, manufacturing processes, and settings. These variables may have significant effects on the accuracy of the volumetric

loss calculations, final wear measurements, and, in certain situations, mechanical characteristics of the materials are represented incorrectly. There is lack of standardization about the studies[207]. We had difficulties especially regarding cycle number and pH. We tried to contribute to the literature on this subject.

5.1.1. Material Selection

In our study, four **PMMA** based occlusal splint materials were selected as Additive Manufacturing (AM), Keysplint Hard (Keystone Industries GmbH, Singen, Germany); Subtractive Manufacturing (SM), Ceramill A Splint (Amann Girrbach AG, Koblach, Austria); Autopolymerizing Resin (CC), ProBase Cold (Ivoclar Vivadent AG, Schaan, Liehchtenstein); Heat-cured acrylic (CH), Duracryl (Duradent, Spofa Dental, Czech Republic).

PMMA resins have long been advised in dentistry because of their advantageous marginal adaptability, strong biocompatibility, low cost, and ease of handling[208,209]. Therefore, in our study we used PMMA.

Reyes-Sevilla et al[210]., examined how various splint materials performed against resin composite materials in terms of wear. These are conventional polyamide resin (ThermoSens) and polymethylmethacrylate (PMMA). Utilizing a profilometer, the wear rate of seven different materials was evaluated after 200,000 cycles. Comparing pPMMA (printed PMMA) and polyamide resin-based splints to cPMMA (conventional PMMA) or mPMMA (milled PMMA), both of them show less wear[210].

According to Gibreel et al.[211], mechanical characteristics of five PMMA and thermoplastic polycarbonate CAD-CAM milled splint materials (ProArt CAD Splint, Therapon Transpa, Temp Premium Flexible Transpa, Cast, and Aqua), in addition to comparing them with PMMA conventional heat-polymerized (Paladon 65) and autopolymerizing resins (Palapress) used in the construction of conventional splints were assesed. CAD-CAM PC based splint materials demonstrate higher fracture toughness and less water sorption when measured against PMMA. While their mechanical properties are generally not superior to conventional heat-polymerized resins, some exceed autopolymerizing acrylic resins in flexural strength and surface microhardness. These advantages indicate their potential for clinical use in splint fabrication.

Benli et al.'s[4] study the two-body wear simulation of PC and PMMA were used to examine variations in surface roughness and wear volume loss of several materials against human maxillary teeth (60000 cycles at 49 N with 5–55 °C thermocycling). The

study's overall findings demonstrated a relationship between the groups' increased Ra values and volume loss following chewing simulation, with two groups' volume loss being equal. In terms of wear and roughness, PC has been determined to be superior to resin materials and PMMA in previous studies. These findings, which show that the PC and PMMA's wear reduction in volume was similar, slightly deviate from those of the current investigation. Based on this result, we chose to use PMMA material, which has been used for many years and has proven biocompatibility.

Benli et al.[4], stated that through a simulated chewing test, surface roughness (Ra) versus enamel antagonism and wear behavior of several milling materials were assessed. There were seventy-five examples in all, composed of polyethyleneterephthalate (PETG), polycarbonate (PC), polymethyl methacrylate (PMMA), and ethylene vinyl acetate (EVA). The results showed that PEEK had the lowest values for both Ra and volume loss, while EVA showed the highest amounts of both. There was no significant difference between PC and PMMA in terms of volume loss. As PEEK produced noticeably better outcomes. However, the primary disadvantage of the PEEK may be its higher cost compared to PMMA splint materials. We preferred to use PMMA because it is a material that is frequently used in the clinic and is more easily obtained financially.

Osiewicz et al.[194] reported on the two-body wear of opposing splint pieces, such as the PMMA-based ProBase heat-cured and cold cured acrylic, whether dentin, enamel, glass-ceramic, or one of four resin composites were used. In comparison to CC, there was more wear on enamel, dentin, and different resin composites when in contact with HC, with dentin showing the most wear. Since we cannot standardize enamel or dentin, we used pins to evaluate the effect of solutions on PMMA.

5.1.2. Shape of Sample

According to a systematic review that published in 2023[212], bar, disc, or rectangular specimen were used in the majority of the studies, while tooth shaped[213,214] and occlusal splint-shaped ones[215,216] were only used in four of them.

Gibreel et al.[56], preferred bar shaped materials for wear (h: 2mm, l:10mm, w:15mm) and for vickers hardness (h: 4mm, l: 10mm, w:10mm). Domanic et al.[8], formed disc shaped materials with h: 3mm, d: 16mm. Osewicz et al.[194], chose the material shape as disc shaped (h: 3mm, d: 10mm). Reyes-Sevilla et al.[210], selected the materials as disc shape (h: 2mm and d: 49mm). Benli et al.[4], preferred disc shaped materials (h: 2mm, d:10 mm). Grymak et al.[217] preferred to create their material in a disk shape with h: 12 mm and

d: 45 mm. Lutz et al.[214] chose the materials' shape as crown-like coping occlusal device. Huettig et al. [195] chose the shape as disc-shaped with h: 3,7mm and d:19,5mm.

Since there was no standardization regarding shape of materials, we obtained disc-shaped materials with a diameter of 16 mm and a thickness of 3 mm to fit the water pipes with an outer diameter of 26 mm, an inner diameter of 17 mm and a height of 30 mm, which we achieved in accordance with the chamber of the chewing simulator.

5.1.3. Surface Treatments Before Testing

A prior study[218] has demonstrated the significance of polishing dental device in order to preserve a smooth surface, reduce bacterial adherence over time, and improve feeling within the oral cavity[218]. The significance of polishing techniques for occlusal device materials has also been confirmed by a recent study[219]. Prior to wear testing, polishing techniques were followed in seven in vitro[4,8,195,210,214,220,221] and two in vivo[222,223] experiments. After testing, other research that did not adhere to polishing guidelines showed noticeably more wear[133,224]. Consequently, it is suggested that the material undergo polishing processes prior to testing.

Benli et al. [4] used 2500 and 4000 grain size grit on a rotary machine under wet conditions. Domanic et al. [8] prefered 220, 500, 800, 1200 waterproof silicon carbide grinding papers and specimens were polished using a universal polishing paste after being finished with a rag wheel and fine pumice slurry. According to Lutz et al. [214] three polishing processes were used on all specimens, reducing the particle size from coarse (95 mm) to medium (50 mm) to fine (5 mm). Using polish, a goat hair brush, and a high-luster buff, a high gloss was obtained. Moreover, Reyes-Sevilla et al. [210] did not do any polishing prior to testing. Wesemann et al. [221] the top surface was polished using discs with grit sizes of 30 μm (P500), 15 μm (P1,200), and 8.5 μm (P2,500), in that order. Gibreel et al. [56] preferred wet polishing with using 800 and 1200 grit silicon carbide abrasive papers. In accordance with Grymak et al. [217], all of the specimens' wear surfaces were successively polished using a polishing machine (TegraPol-21, Struers, Denmark) to a final grit of P2400 carbide paper. This was followed by a final polishing with dental pumice and a high-shine buffing with de nture polishing paste (Ivoclar polishing paste, Liechtenstein).

Polishing is required to simulate clinical application. There is no standardization for polishing procedure for occlusal splint materials. In our study we prefered P600, P800, P1200 waterproof silicon carbide papers (English Abrasives & Chemicals Ltd., Stafford,

UK). Polishing was done with goat hairbrush using pumice, then with gyspsum powder and cotton disc. Polishing process was completed with polishing paste (Ivoclar Vivadent AG, Schaan, Liechtehstein).

5.1.4. Storage prior testing

Samples from nine investigations underwent mechanical testing without any prior water storage[4,195,213,215–217,225–227], whereas samples from the other studies were kept in water at different temperatures and intervals[8,10,56,194,210,211,220,228,229]. Only two research used artificial saliva as their test media[195,217]; the tests in the other investigations were conducted using distilled water or dry medium.

According to Gibreet et al. [56], their in vitro research's goal was to examine wear and surface hardness of nine components that are utilized in occlusal splint manufacturing processes such as subtractive milling, traditional manufacturing, and 3D printing. Additionally, evaluate samples used in rigid and flexible 3D-printed occlusal splints in terms of wear and surface hardness. Materials were held in both dry and wet conditions for thirty days at 37°C in order to evaluate the Vickers surface hardness (VHN), whereas specimens were kept in water at 37°C for twenty-four hours prior to testing. It was shown that surface hardness tended to significantly decrease with water storage. In other words, exposing the acrylic material to liquid before testing to remove residual monomer may cause the hardness of the material to decrease, which may increase wear or water may cause changes in the bonding of the material, which may affect sorption and solubility. On the other hand, Domanic et al.[8] and Kurt et al[220]., stored materials in distilled water prior testing at 25°C for 2 weeks. Osiewicz et al.[194], Reyes-Sevilla et al.[210] stored the materials in distilled water at 20°C for 2 months. But this is against clinical applicability. We cannot leave a patient with bruxism without a splint for two months.

In our study, after the materials were obtained and polished, they were cleaned in an ultrasonic bath with distilled water at room temperature for 10 minutes to remove residual monomer.

5.1.5. Evaluated Parameters

According to literature, hardness was the most often examined parameter in the analyzed research[8–10,211,225–233]. Vickers hardness[10,56,211,227–229,231], Shore-D hardness[225,226], Knoop hardness[230], Martens hardness[10,232,233], and Brinell hardness[9] tests were among the various methods used to measure hardness. "Wear," which was assessed

in 13 research, was the second most often evaluated mechanical characteristic among the included studies[4,8,56,194,195,210,213–217,232,234]. Flexural strength[9,10,211,227–229,231,232,235], e-modulus[211,216,225,227,232,233], flexural modulus[228,229,231,235], fracture toughness[211,228,229,231], and polishability[195,225] were among the other measurements assessed in the study.

The study's aim is to compare the mechanical properties such as; wear resistance and surface roughness of various splint materials that produced from digitally and conventionally exposed with gastric acid and control group as artificial saliva. These parameters were examined to evaluate the status of materials against gastric acid worn and chewing cycle in GERD patients and to evaluate effect of surface roughness on bacterial retention.

5.1.6. Aging

According to Kurt et al. [220], for every component and cycles, there was no discernible difference among wet and dry conditions due to the lack of influence of water on the volume loss in the various material subgroups. According to Lutz et al. [68] occlusal devices are typically worn at night, when eating or drinking hot or cold food does not produce variations in temperature. As a result, the study was carried out at isothermally 37°C.

According to a systematic review from Benli et al.[212] it was noted that over half of the studies that were part of the review were conducted in vitro at 20 °C since the subjects were not subjected to temperature variations in response to changes in food and drink consumption while wearing the occlusal splint[9,10,56,194,210,216,225–233,235].

Samples of nine studies were not aged before the tests[4,195,213,215–217,225–227], while the other samples were aged at different temperatures and at different time intervals ranging from 24 h to two months. The most frequently used temperature value among the applied aging procedures is 37 °C[9,10,56,211,231–233,235,236].

There are three thermal options in the chewing simulator we use. The first is the thermal cycle and simulates 5-55°C. The second is the intraoral simulation and simulates 37°C. The third one is the dry test. We chose not to simulate at 5-55°C because the occlusal splint would be used during sleep and nothing cold or hot could be taken into the mouth during this time. We chose to use 37°C, which was determined by the simulator to simulate the oral environment. All chambers were filled with distilled water to simulate saliva.

5.1.7. pH of GERD selection for this study

One of the main causes of dental erosion is GERD. The flow of stomach contents into the esophagus is known as gastroesophageal reflux disease (GERD), while GER is the term used to describe symptoms or complications of GER. The most commonly recognized standard for diagnosing gastric reflux disease (GERD) is the occurrence of heartburn twice a week or more[145].

Ayazi et al.[237] suggested a pH cutoff of 5.0 when the patient is in a supine position and 5.5 when the patient is upright.

According to Barron et al.[145] in vitro experimental erosion has been shown to occur at an oral pH of less than 3.7. Saliva can lessen the chance of demineralization, therefore any decrease in salivary flow rate or buffering ability needs to be taken into consideration.

Alnasser et el. [238] compared the surface roughness of various CAD-CAM restorative dental components alters after being exposed to an acidic pH. For 45 and 91 hours, test materials were subjected to HCl (pH=2).

The surface topography and optical characteristics of monolithic zirconia were assessed by Sulaiman et al. [169] following immersion in simulated gatric acid. For 24 and 96 hours, respectively, they employed gastric juice to simulate two and more years of clinical exposure.

According to Tinastepe et al. [239] changes in the surface roughness and hardness of denture base acrylic resins after exposure to gastric acid was evaluated. 1.2 pH was used in this study. Every specimen's Vickers surface hardness values (VHVs) were noted at baseline (T0), 24 hours (T24), and 96 hours (T96) following immersion.

There is not enough consensus on how to simulate gastric acid in vitro and how long it would take to reproduce the clinical situation[239]. The half-life of the gastric acid attack experienced during sleep is 2-4 minutes. The patient's REM sleep usually begins 4 hours after going to bed and may vary depending on the patient's sleep cycle. According to the literature, it was said that gastric acid attack can be happened 3 times in the sleep time between 1-5 minutes. The patient may feel as if this is happening all night long.

Considering that the sleep time is 8 hours;

Minimum Scenario for GERD

1*30=30 minutes (monthly)

30*6=180 minutes (for six months) ≈ 3 hours

3*3=9 hours (Considering that there were 3 attacks)

Maximum Scenario for GERD

5*30=150 minutes (monthly)

150*6=900 minutes (for six months) ≈ 15 hours

15*3=45 hours (Considering that there were 3 attacks)

The duration of acid reflux attacks experienced by a patient with reflux during sleep at night varies depending on the person's eating habits, sleeping position, medications taken, and the person's general health condition. In this case, reflux attacks may last longer. In our own study, we exposed the GERD group to gastric acid at pH 4 for 45 hours to simulate 6 months, while we exposed the control group to artificial saliva at pH 7 for 45 hours under the same time and conditions to standardize the study.

5.1.8. Cyle number on Chewing Simulator

According to the literature, approximately 240,000-250,000 cycles in the chewing simulator correspond to one year of use in the clinical environment[240]. But this cycle corresponds to 24 hours. The splint used for patients who grind their teeth at night should be based on sleep time. For this reason, we calculated the duration of teeth clenching during sleep (approximately 8 hours) in order to provide an accurate simulation in our study[241]. Because patients do not grind their teeth all night long, they grind their teeth only during a short period of sleep. Therefore, the whole night should not be taken as a basis.

Sleep bruxism appears to occur as a reaction to micro-arousals during sleep: three to ten seconds-long episodes of increased heart rate and muscle tone in healthy subjects, eight to fifteen times each hour[21].

Minimum Scenario for Bruxism:

- Minimum number of bruxism cycles per hour: 8
- Sleep duration per day: 8 hours
- Number of days in a month: 30 days
- Period: 6 months

Calculation: Minimum bruxism cycles: 8 (cycles/hour) * 8 (hours/day) * 30 (days/month) * 6 (months)

Minimum bruxism cycles: 11,520

Maximum Scenario for Bruxism:

- Maximum number of bruxism cycles per hour: 15
- Sleep duration per day: 8 hours
- Number of days in a month: 30 days
- Period: 6 months

Calculation: Maximum bruxism cycles: 15 (cycles/hour) * 8 (hours/day) * 30 (days/month) * 6 (months)
Maximum bruxism cycles: 21,600

A patient who grinds their teeth at night could experience the following range over a 6-month period in a chewing simulator:

- 11,520 cycles: the minimum value.
- 21,600 cycles: the maximum value.

These mathematical calculations offer a broad range for the scenarios that are presented. The person's actual circumstances could be anywhere in this range.

For this reason, in our study, we only calculated the value of teeth clenching at night. Accordingly, we used 20,000 cycles in the chewing simulator (MOD Dental, Esetron Smart Robot Technologies, Ankara, Turkey) as an average value.

According to Grymak et al.'s systematic review[207], one year of clinical wear is equivalent to 250,000 masticatory cycles[231], based on the clinical correlation with the artificial oral cavity; however, this is not equivalent to the same number of bruxing sensations. This study mentioned that an average of 8.1 cycles per hour (8.1 × 24 hours × 365 days = 70,956 cycles per year) of tooth-to-tooth contact resulting from clenching, bruxism, and parafunctional pressures is responsible for the majority of tooth wear. If we evaluate this as patients who grind their teeth all day, the result for calculating night bruxism is 70.965/3 (for 8 hours sleep time) = 23.655. This is a result close to the value we calculated. However, since we directly calculate the contact of the teeth at night, it may be a more accurate result. To get a consensus on the number of cycles that would normally represent the tooth-to-tooth contact throughout a year for sleep bruxism, more research is therefore required.

5.1.9. Applied Force Magnitude

An in-vitro wear test study must include loading force. Physiologic chewing forces in the human mouth range from 10 to 120 N [179,242]. A load of 49–50 N has been employed in many chewing simulation studies and is generally thought to represent the mean value of physiological occlusal forces [4,8,68,207,210,220]. On the other hand some studies has been applied lower force (5N) than the physiologic chewing forces [195,227]. Some studies has been applied higher force than the physiologic chewing forces [9,216,235]. Parafunctional chewing forces can reach a mean of 380 N, therefore increased forces should be preferred in the simulator in an artificial environment. The maximum load capacity of the chewing simulator (MOD Dental, Esetron Smart Robot Technologies, Ankara, Turkey) using in this study is 100 N with 1,66 Hz, 6 mm vertical, 0,7 mm horizontal movement [188].

According to Gibreel et al. [194,220] 15.000 cycle, 1.5 Hz and 20 N were used for two-body wear[56]. On the other hand, 200.000 cycle, 1 Hz with 15 N were applied to the materials via both Osiewicz et al. and Reyes-Sevilla et al.[210]. 60.000 cycle, 0.8 Hz with 49N values were used for 3 months of simulations by Benli et al.[4]. Kurt et al. as well as Domanic et al.[8] simulated 10.000, 20.000, 30.000 cycle with 1,6 Hz and 50 N. Notably, given the established number of 800–1400 chewing cycles per day, 10,000 chewing cycles is nearly the same as the most that daytime splint users can experience in a week, and nocturnal users can encounter in two weeks [220]. But even when this idea is advanced, 12 hours of sleep is taken as a basis. However according to current systematic review, adults and older adults sleep 7-8 hours a day[241]. That's why we made our calculations based on 8 hours of sleep.

Moreover, Schmeiser et al. [213] applied 120.000 cycle, 1.3 Hz and 50 N in their study. Patzelt et al. [215] used 1.2 million cycles, 1.3 Hz and 50 N, but in this study they did not mention how many years they did the 1.2 million cycles. 120.000 cycle, 2 Hz and 50 N were applied by Rosenritt et al.[232]. The 20.000 and 120.000 cycle with 50 N in Lutz et al.'s study corresponds to using the occlusal device at night for a duration of 1 and 6 months, respectively. The calculation was based on a daily maximum of 1400 mastication cycles. This value is obtained from the 5-year simulation being 1.2 million cycles; but there are 2 conditions. First, even if we calculate in this way, 12 hours of sleep per day is too much for the general average today. The main problem is that a patient using an occlusal splint does not clench their teeth throughout the night. Sleep bruxism occurs as a response to micro-arousals, which are three to ten second attacks of high heart rate and

muscle tone that occur eight to fifteen times per hour in healthy individuals. For this reason, we proceeded in our study based on this calculation[21]. The results for material volume loss are not consistent with the findings of Huettig et al.'s study[195], which indicated that wear resistance in additive, subtractive, and conventionally made specimens was similar. In contrast, just 5 N and 5000 cycles were used in the present study, which may not sufficient parameters to simulate the clinical situation.

5.1.10. Measurement of the wear parameters

Comparing 3D sequential images utilizing contact and non-contact profilometers, laser scanners, and micro or cone beam CT scanners is the most effective way to assess wear[243]. Additionally, because 3D scanning minimizes chairside time and allows for considerably faster and more accurate data acquisition than impressions, it is appropriate for both in vitro and in vivo experiments[244–247]. Therefore, after the samples were exposed to the solutions and before being placed in the chewing simulator, 3D scanning was performed with a model scanner (D1000, 3Shape GmbH, Copenhagen, Denmark) and saved in STL format. The same process was repeated after the materials were placed in the chewing simulator.

Volumetric loss is the most accurate approach to measure wear parameters since it can be used to clinical wear and is a function of area and depth region. It can be easily estimated with the assumption that the antagonist parameter will remain largely constant because of its simple force, number of chewing cycles, and constant glide route distance[244,245,248]. While two studies calculated the vertical loss[133,195], five investigations used the volumetric loss[4,214,221,224,249] method and found that volumetric loss is more accurate than vertical loss.

The indenter moves differently in different studies; certain researchers prefer circular movements[195,250], while others include linear vertical and horizontal movements[8,220,251]. Linear movements improve the repeatability and comparability of the results, especially for long-term simulations. However, a circular movement pattern—which reflects physiological movement during mastication—was preferable in this study [252].

Current study, both linear and volumetric wear values were evaluated for observing the difference separately and different results were obtained. Since the chewing simulator imitates chewing movements (horizontal and vertical movements), we thought that evaluating volumetric wear was more appropriate.

5.1.11. Measurement of the surface roughness

Clinical modifications like polishing have an impact on surface roughness because they provide smooth surfaces that wear less, prevent long-term bacterial adherence and have an additional advantage of extending the life of the restoration and oral appliances. Therefore, polishing is advised to get appropriate clinical performance and to prevent occlusal splint surfaces getting worn. Prior to the chewing simulation, all group had Ra levels that were clinically acceptable, meaning they did not above the permitted limit value of 0.2 μm for hard surfaces in the oral environment [4,253,254].

In this study, 3D optical profilometer (Profilm 3D, Filmetrics®, USA) was used for surface roughness evaluation. The average overall surface roughness of the material was initially less than 0.2 μm for each material. However, since non-contact profilometer was used in this study, we wanted to calculate the roughness of the deepest part of the direct wear area both initially and ultimately. For this reason, the values in the surface roughness initial tables being greater than 0.2 μm should not mislead researchers. Unlike other studies, we applied this machine, because it can measure the roughness of the deepest point of the material.

5.2. Findings

First null hypothesis is that no statistical differences in terms of surface roughness change among materials produced by different techniques in gastric acid environment was partially rejected. The amount of roughness change of heat-cured acrylic was found to be significantly higher than the other materials in gastric acid environment. There was no significant difference between the other materials in terms of roughness change. Since there is no study yet examining the effect of gastric acid on occlusal splints, studies on PMMA-based denture materials with similar chemical content can be taken as a basis.

For example, Papathanasiou et al.[255] assessed the effect of commonly used solutions which are Coca-cola (pH: 2-3), coffee (pH: 4), red wine (pH: 3) and distilled water (pH: 6-7) on color stability, gloss, and surface roughness of removable partial dental prostheses polymers (PEEK polymer, a polyamide, an acetal resin and a heat-cured PMMA acrylic resin). It is found that PMMA's surface roughness value was the lowest value similar to PEEK. However, the specimes did not exceed the critical value before and after immersion in any solution. In our study, CH was roughened more in gastric acid. This may be due to the different content of gastric acid compared to other solutions like cola or red wine as well as the use of PMMA with different content.

Mickeviciute et al.'s[256] study the effects of staining solutions (cola and coffee) on the color stability and roughness of different provisional resin material (two PMMA and one bis-acryl composite resins) was evaluated. In this study, both coffee and cola increased the surface roughness of the materials, and cola also had more adverse effect. Similar to our study, acid affects the roughness of the materials; but unlike our study, the solution type, materials and methodologies are different.

In Tinastepe et al.'s[239] study, effect of gastric acid (pH: 1.2) on hardness and roughness change of denture base acrylic resin obtained from 3 different brands (compression-molded, injection-molded, and CAD-CAM milled) with different processing methods was evaluated at 2 time periods (24 and 96 hours). Milled specimens immersed in astric acid for both periods showed lowest changes in surface hardness and surface roughness values. In our study, CAD-CAM was the least rough, too; however, in order to compare the values accurately, a chewing simulator and similar materials are required.

Huettig et al.[195] evaluated surface roughness and wear resistance of three resins for conventional (C) (cold cured), subtractive (S) and additive manufacturing (A) of occlusal splints with application of 5000 cyle and 50 N in the chewing simulator. All

specimen showed a comparable surface roughness (Ra) of C= (0.062±0.01 μm); S= (0.048±0.009 μm); A= (0.06±0.007 μm), statistically significantly difference. Heat-cured acrylic was not evaluated in this study. The roughness of cold-cured acrylic was higher than other materials. Unlike our study, artificial saliva was evaluated; if it was examined in acid, it may give different results. The low number of cycles and applied force in study and fact that measuring device was contact profilometry may be other reasons for the different results.

Second null hypothesis is that no statistical differences in terms of volume change among materials produced by different techniques in gastric acid environment was partially rejected., The amount of volume loss of CC was found to be significantly higher than the other materials in gastric acid environment.

In Lutz et al.[214]evaluated PMMA and resin materials' volume loss for 3DP (additive manufacturing), CAM (milled) and cast (injection molding/conventional) PMMA. 20,000 cycles were calculated as 1 month for a nighttime splint user and materials tested in distilled water. In other words, the patients were calculated as if they were clenching their teeth for the entire 12 hours of sleep. The greatest wear was found in 3DP (1.2±0.3), followed by milling (0.7±0.1) and conventional (injection molding) (0.4±0.1) in this study. In our study, milling materials are the group that is least worn by both saliva and gastric acid. Storage prior testing at 37°C for 24 hours, application of 50 N load and the presence of human enamel in the antagonist as well as difference of the materials may explain the difference in volume loss between these studies.

According to Gibreel et al.[56], wear and surface hardness of occlusal splint materials produced by conventional manufacturing (heat-cured and autopolymerizing resin), subtractive milling (PMMA and Polycarbonate), and 3D printing (flexible and rigid) were evaluated. Heat-cured (27.5±2.4 μm), milled (30.5±2.8 μm), and autopolymerizing resin (36.7±6.3 μm) showed the least amount of vertical wear, with no statistically significant differences. Components used for milling splints made of polycarbonate and flexible 3D printed materials exhibited the most wear. Regardless of the production method, PMMA-based materials had the most stable surface hardness and wear resistance. The fact that we use all our materials as PMMA and rigid occlusal splint can be associated with the results in this study. The reasons for the difference in our results may be due to the evaluation of vertical loss instead of volumetric loss, the storage of the samples in water for 24 hours before the experiment, and the application of the chewing simulator at 20 N and 15,000 cycles, unlike our study.

Grymak et al.[217], investigated the volumetric and vertical loss of occlusal splints manufactured by conventional (heat-cure), subtractive (CAD/CAM) and additive (3D-printing) methods. PMMA materials that were CAD-CAM milled had the highest wear resistance closely followed the heat-cured meterial (both in volumetric and vertical wear). 3D printed materials were significantly lower in wear resistance. In this study, a similar result was found for volumetric loss as in our study (for both acid and saliva). In our study, the least wear was found in milled materials, while the most wear was in 3D materials; but there was no statistically significant difference between them.

On the other hand, following 200,000 loading cycles at 20 N and 50 N, respectively, conventional (injection molding), milled, and additively made occlusal splints showed comparable wear resistance in the study by Wesemann et al.[221]. The applied force of 20 N and 50 N did not cause a statistically significant change between the materials. Milled materials were found the highest hardness followed by the conventional one while additive manufacturing showed reduce surface hardness. The results found in this study may be related to the application of 5-55°C thermocycling or the presence of enamel in the antagonist, which is different from current study. The fact that the chewing cycle was 10 times higher than in our study and the applied load was lower than our study may also be related to the different results. Additionally, keeping the materials at 37°C for 24h before the experiment may be a reason for the difference between the results.

Schmeiser et al.[213] Examine the differences between three-dimensional printing (3DP) (flexible splint) and CAM occlusal splint materials' two-body wear throughout a six-month simulation period using enamel antagonist at 120.000 cycles and 50 N. Although the volumetric loss of milled materials was found to be higher than 3DP materials, no statistical difference was observed. The cause of this outcome may be related to the flexible elastic deformation. Unlike our study, the number of cycles was overestimated, low force was applied, no evaluation was made in an acidic environment, and different materials were used.

Prpic et al.[9] investigated the flexural strength and surface hardness of different materials (conventional (PMMA autopolymerizing resin), milled (PMMA and polyamide) and 3D printing) for creating occlusal devices. In the current research, PMMA (three traditional autopolymerizing materials and one CAD-CAM occlusal device material) produced the most consistent results for surface hardness, with acrylic resin for additive manufacturing coming in second. Explanation of this may be related to the

layering of 3D materials or the dense structure of the CAD-CAM disk. In our study, the reason for the less wear of milled materials may be associated with the harder material, the reason for the more wear of autopolymerizing resin in our study may be the presence of the chewing simulator and the acidic environment. However, since there was no acidic environment in this study and the same experimental environment was not applied, only comments can be made on the general properties of the material.

The third null hypothesis is that no statistical differences in terms of roughness change in occlusal splint materials for gastric acid and artificial saliva environment after chewing simulator was partially rejected. When AM and SM materials were used; the amount of increase in the roughness change of the saliva solution was statistically higher than that of gastric acid. When CH and CC materials were used; there was no statistically significant difference between saliva and gastric acid in terms of the amount of increase in the roughness change. It may be thought that the acrylic layer on the surface was removed with the acid and the chewing simulator as well as reached the new and solid layer underneath, so the acid had less effect on the roughness in the wear area for AM and SM materials. Also, when the initial and final surface roughness were evaluated in different solutions, the only material that was roughened more by gastric acid was CH, while saliva roughened the others more. This can be explained by CH taking in water and swelling while being eroded. Although this brand is one of the most routinely used materials in our country, it is thought that different results can be obtained if a product from another brand is used.

In Tinastepe et al.'s[239] study, the effect of gastric acid (pH: 1.2) on the hardness and roughness change of denture base acrylic resin (compression-molded, injection-molded, and CAD-CAM milled) was evaluated at 24 and 96 hours. In this study, chewing simulator was not used. Therefore, in T24 and T96, the effect of acid on the roughness of the materials was found to be greater than saliva. This circumstance might be connected to the material's structural characteristics. And in all cases, as in our study, the materials produced by milling were found to be less rough. The pH was chosen as 1.2 in this study, while it was chosen as 4 in ours. The difference between the values can be attributed to the use of PMMA with different chemical content, the application of the chewing simulator, and the duration of waiting in acidic environment and different pH.

Nezir et al.[257] assessed how several monochromatic universal composite resins' discolouration, surface roughness (SR), and microhardness were affected by detox solution (pH: 4,04). The correlation between roughness and hardness was not evaluated.

In all monochromatic universal composite resin restorative materials studied, an increase in surface roughness values was noted following immersion in the detox solution, although no statistically significant finding in this aspect when compared to the baseline. Although different materials were used and no chewing simulator was used, it was observed that the acidic environment had an effect on the material.

Zakir et al.[258] assessed surface roughness of two dental ceramics (IPS Emax ceram & Noritake®) before and after they are exposed to mouthwash, Coca-Cola, and a simulated vomit solution (pH=3,8) made of distilled water. Greatest variation in surface roughness was induced by a vomit simulation solution, which was followed by Coca-Cola and chlorhexidine mouthwash. After being submerged in distilled water, no alterations were seen. Although dental ceramics are known for their durable chemical properties compared to acrylic, more roughness has been observed in gastric acid than in other solutions. The difference in the results is thought to be due to the difference in materials, the application of the chewing simulator and the evaluation of the deepest point of the wear for roughness.

The forth null hypothesis is that no statistical differences among occlusal splint materials in terms of volume change in gastric acid and artificial saliva environment after chewing simulator was partially rejected. When AM, SM and CH materials were used; there was no statistically significant difference in terms of volumetric loss amounts between saliva and gastric acid; but in CC materials, the volumetric loss amount in gastric acid was statistically significantly higher than in saliva solution.

Sagsoz et al.[259] investigated the effect of organic acids, heptane and ethanol (the foodsimulating liquids) on CAD/CAM restorative materials (3M ESPE LAVA Ultimate, VITA Enamic, IPS e.max CAD, VITA Suprinity). In contrast to other solutions, weak acids do not significantly alter the structure of resin ceramics, but they do weaken it in glass-matrix ceramics. In this study, it was stated that acid has an effect on different types and harder materials. Unlike our study, different solutions and different materials were used.

Ranjitkar et al.[260] assess the characteristics of wear between specimens in which enamel was opposed to dentine at loads (32 N, 62 N, 100 N) simulating attrition and at pH values (pH: 1.2, 3.0, 6.1) simulating different erosive environments. It was shown that when mechanical action and gastric reflux were simulated, enamel worn faster than dentine because of its greater inorganic content. Even resistant materials such as enamel and dentin are eroded by acid. In this case, it is inevitable that the solutions will affect the

acrylic material. Therefore, gastroenterologists and dentists should be aware of the situation and work together.

The fifth null hypothesis is that no statistical differences in correlation between surface roughness and wear resistance. When the groups were examined separately, no statistically significant relationship was found between the change in surface roughness and the amount of volumetric change in any group. Hence, the fifth null hypothesis was accepted. This may be related to our measurement of the roughness of the deepest part of the wear.

In the study of Benli et al.[4], roughness and wear values of EVA, PMMA, PC, PEEK and PETG materials produced by milling were examined in a chewing simulator. 60,000 cycles and 49 N load were applied in the study. In terms of both roughness and wear values, EVA showed the highest roughness, PEEK showed the lowest, and PC and PMMA showed similar values. The study's overall findings demonstrated a relationship between the groups' increased Ra values and volume loss following the chewing simulation. The different results may be due to the application of different cycles and forces compared to our study, the use of different materials, and the exclusion of additive and conventional manufacturing from the experiment.

Limitation

Long-term in-vivo research is needed to evaluate the results on wear resistance.

On the other hand, water was added to the liquid reservoir of the chewing simulator because acid or saliva would damage the device. The materials were exposed to the solutions before the simulator.

In order to better simulate the oral environment, there should be enamel or other restorative materials instead of steel ball in the opposing arch.

Also, for materials produced on a 3D-printer, studies may be done at different angles, devices, post-processing processes and layer thicknesses.

Another limitation is that lack of standardization from the polishing stage to the measurements, and for this purpose, literature on different topics is used to proceed.

6. CONCLUSION

The results of this in-vitro study led to the following conclusions:

1. As there is a relationship between GERD and bruxism, the splint material used may be affected by the gastric acid environment. Dentists should be careful when choosing materials for patients with GERD.
2. SM specimens immersed in gastric acid and artificial saliva solutions showed the lowest changes both in volume loss and surface roughness values.
3. CC materials have been found to worn more when exposed to gastric acid and extra care should be taken when using this material for a GERD patient.
4. When CH materials were evaluated in terms of surface roughness, they became rougher in the 6-months period and this situation requires more frequent recall.
5. When surface roughness and wear are evaluated, AM materials can be preferred as an alternative in bruxist patients with reflux, but it should also be taken into consideration that numerically they have high wear.
6. Since this is the first study on this subject, studies that include more material and different antagonists are needed.

7. REFERENCES

1. Manfredini D, Ahlberg J, Lobbezoo F. Bruxism definition: Past, present, and future - What should a prosthodontist know? *J Prosthet Dent*. 2022;128(5):905-912. doi:10.1016/j.prosdent.2021.01.026
2. Kuang B, Li D, Lobbezoo F, et al. Associations between sleep bruxism and other sleep-related disorders in adults: a systematic review. *Sleep Med*. 2022;89:31-47. doi:10.1016/j.sleep.2021.11.008
3. Minakuchi H, Fujisawa M, Abe Y, et al. Managements of sleep bruxism in adult: A systematic review. *Jpn Dent Sci Rev*. 2022;58:124-136. doi:10.1016/j.jdsr.2022.02.004
4. Benli M, Eker Gümüş B, Kahraman Y, et al. Surface roughness and wear behavior of occlusal splint materials made of contemporary and high-performance polymers. *Odontology*. 2020;108(2):240-250. doi:10.1007/s10266-019-00463-1
5. Abad-Coronel C, Ruano Espinosa C, Ordóñez Palacios S, Paltán CA, Fajardo JI. Comparative Analysis between Conventional Acrylic, CAD/CAM Milled, and 3D CAD/CAM Printed Occlusal Splints. *Materials (Basel)*. 2023;16(18). doi:10.3390/ma16186269
6. Al-Ani Z, Gray RJ, Davies SJ, Sloan P, Glenny A. Stabilization Splint Therapy for the Treatment of Temporomandibular Myofascial Pain: A Systematic Review. *J Dent Educ*. 2005;69(11):1242-1250. doi:10.1002/j.0022-0337.2005.69.11.tb04023.x
7. Greene CS, Menchel HF. The Use of Oral Appliances in the Management of Temporomandibular Disorders. *Oral Maxillofac Surg Clin North Am*. 2018;30(3):265-277. doi:10.1016/j.coms.2018.04.003
8. Yildiz Domanic K, Aslan YU, Ozkan Y. Two-body wear of occlusal splint materials against different antagonists. *BMC Oral Health*. 2020;20(1):174. doi:10.1186/s12903-020-01165-9
9. Prpic V, Slacanin I, Schauperl Z, Catic A, Dulcic N, Cimic S. A study of the flexural strength and surface hardness of different materials and technologies for occlusal device fabrication. *J Prosthet Dent*. 2019;121(6):955-959. doi:10.1016/j.prosdent.2018.09.022

10. Berli C, Thieringer FM, Sharma N, et al. Comparing the mechanical properties of pressed, milled, and 3D-printed resins for occlusal devices. *J Prosthet Dent*. 2020;124(6):780-786. doi:10.1016/j.prosdent.2019.10.024
11. ISO/TS 14569-2:2001 - Dental materials — Guidance on testing of wear — Part 2: Wear by two- and/or three body contact. https://www.iso.org/standard/24300.html.
12. Lambrechts P, Debels E, Van Landuyt K, Peumans M, Van Meerbeek B. How to simulate wear? Overview of existing methods. *Dent Mater*. 2006;22(8):693-701. doi:10.1016/j.dental.2006.02.004
13. Heintze SD. How to qualify and validate wear simulation devices and methods. *Dent Mater*. 2006;22(8):712-734. doi:10.1016/j.dental.2006.02.002
14. Mengatto CM, Dalberto C da S, Scheeren B, Barros SGS de. Association between sleep bruxism and gastroesophageal reflux disease. *J Prosthet Dent*. 2013;110(5):349-355. doi:10.1016/j.prosdent.2013.05.002
15. Li Y, Yu F, Niu L, Long Y, Tay FR, Chen J. Association between bruxism and symptomatic gastroesophageal reflux disease: A case-control study. *J Dent*. 2018;77:51-58. doi:10.1016/j.jdent.2018.07.005
16. Ohmure H, Sakoguchi Y, Nagayama K, Numata M, Tsubouchi H, Miyawaki S. Influence of experimental oesophageal acidification on masseter muscle activity, cervicofacial behaviour and autonomic nervous activity in wakefulness. *J Oral Rehabil*. 2014;41(6):423-431. doi:10.1111/joor.12159
17. Cebrián-Carretero JL, López-Arcas-Calleja JM. Gastroesophageal reflux diagnosed by occlusal splint tintion. *Med Oral Patol Oral Cir Bucal*. 2006;11(1):E26-8.
18. Aydın N, Topçu F, Karaoğlanoğlu S, Oktay E, Erdemir U. Effect of finishing and polishing systems on the surface roughness and color change of composite resins. *J Clin Exp Dent*. 2021:e446-e454. doi:10.4317/jced.58011
19. Jain V, A. Platt J, Moore K, M. Spohr A, A. Borges G. Color stability, gloss, and surface roughness of indirect composite resins. *J Oral Sci*. 2013;55(1):9-15. doi:10.2334/josnusd.55.9
20. Hickl V, Strasser T, Schmid A, Rosentritt M. Effects of storage and toothbrush simulation on color, gloss, and roughness of CAD/CAM, hand-cast, thermoforming, and 3D-printed splint materials. *Clin Oral Investig*. 2022;26(5):4183-4194. doi:10.1007/s00784-022-04391-3

21. Beddis H, Pemberton M, Davies S. Sleep bruxism: an overview for clinicians. *Br Dent J*. 2018;225(6):497-501. doi:10.1038/sj.bdj.2018.757
22. Mesko ME, Hutton B, Skupien JA, Sarkis-Onofre R, Moher D, Pereira-Cenci T. Therapies for bruxism: a systematic review and network meta-analysis (protocol). *Syst Rev*. 2017;6(1):4. doi:10.1186/s13643-016-0397-z
23. Oliveira JMD de, Pauletto P, Massignan C, et al. Prevalence of awake Bruxism: A systematic review. *J Dent*. 2023;138:104715. doi:10.1016/j.jdent.2023.104715
24. Gauer RL, Semidey MJ. Diagnosis and treatment of temporomandibular disorders. *Am Fam Physician*. 2015;91(6):378-386.
25. Al-Moraissi EA, Farea R, Qasem KA, Al-Wadeai MS, Al-Sabahi ME, Al-Iryani GM. Effectiveness of occlusal splint therapy in the management of temporomandibular disorders: network meta-analysis of randomized controlled trials. *Int J Oral Maxillofac Surg*. 2020;49(8):1042-1056. doi:10.1016/j.ijom.2020.01.004
26. Nassif M, Haddad C, Habli L, Zoghby A. Materials and manufacturing techniques for occlusal splints: A literature review. *J Oral Rehabil*. 2023;50(11):1348-1354. doi:10.1111/joor.13550
27. Ré JP, Chossegros C, El Zoghby A, Carlier JF, Orthlieb JD. Gouttières occlusales. Mise au point. *Rev Stomatol Chir Maxillofac*. 2009;110(3):145-149. doi:10.1016/j.stomax.2009.03.008
28. Manrriquez SL, Robles K, Pareek K, Besharati A, Enciso R. Reduction of headache intensity and frequency with maxillary stabilization splint therapy in patients with temporomandibular disorders-headache comorbidity: a systematic review and meta-analysis. *J Dent Anesth Pain Med*. 2021;21(3):183-205. doi:10.17245/jdapm.2021.21.3.183
29. Christidis N, Doepel M, Ekberg E, Ernberg M, Le Bell Y, Nilner M. Effectiveness of a Prefabricated Occlusal Appliance in Patients with Temporomandibular Joint Pain: A Randomized Controlled Multicenter Study. *J Oral Facial Pain Headache*. 2014;28(2):128-137. doi:10.11607/ofph.1216
30. NILSSON H, VALLON D, EKBERG EC. Long-term efficacy of resilient appliance therapy in TMD pain patients: a randomised, controlled trial. *J Oral Rehabil*. 2011;38(10):713-721. doi:10.1111/j.1365-2842.2011.02210.x

31. Guo YN, Cui SJ, Zhou YH, Wang XD. An Overview of Anterior Repositioning Splint Therapy for Disc Displacement-related Temporomandibular Disorders. *Curr Med Sci*. 2021;41(3):626-634. doi:10.1007/s11596-021-2381-7
32. Klasser GD, Greene CS. Oral appliances in the management of temporomandibular disorders. *Oral Surgery, Oral Medicine, Oral Pathology, Oral Radiology, and Endodontology*. 2009;107(2):212-223. doi:10.1016/j.tripleo.2008.10.007
33. Wright EF, Schiffman EL. Treatment alternatives for patients with masticatory myofascial pain. *J Am Dent Assoc*. 1995;126(7):1030-1039. doi:10.14219/jada.archive.1995.0281
34. Redtenbacher Jos. Ueber die Zerlegungsprodukte des Glyceryloxydes durch trockene Destillation. *Justus Liebigs Ann Chem*. 1843;47(2):113-148. doi:10.1002/jlac.18430470202
35. Zafar MS. Prosthodontic Applications of Polymethyl Methacrylate (PMMA): An Update. *Polymers (Basel)*. 2020;12(10):2299. doi:10.3390/polym12102299
36. ISO. ISO 20795–1:2013 Dentistry-Base Polymers-Part 1: Denture Base Polymers. Available online: https://www.iso.org/standard/62277.html.
37. Ali U, Karim KJBtA, Buang NA. A Review of the Properties and Applications of Poly (Methyl Methacrylate) (PMMA). *Polymer Reviews*. 2015;55(4):678-705. doi:10.1080/15583724.2015.1031377
38. Rama Krishna Alla RSKRVAK. Conventional and Contemporary polymers for the fabrication of denture prosthesis: part I – Overview, composition and properties. *nternational Journal of Applied Dental Sciences*. 2015;1(4):82-89.
39. Bartoloni JA, Murchison DF, Wofford DT, Sarkar NK. Degree of conversion in denture base materials for varied polymerization techniques 1. *J Oral Rehabil*. 2000;27(6):488-493. doi:10.1046/j.1365-2842.2000.00536.x
40. Anusavice K.J. SC, RHR. *Phillips' Science of Dental Materials*.; 2012.
41. van Noort R. The future of dental devices is digital. *Dent Mater*. 2012;28(1):3-12. doi:10.1016/j.dental.2011.10.014
42. Park YJ, Chae KH, Rawls HR. Development of a new photoinitiation system for dental light-cure composite resins. *Dent Mater*. 1999;15(2):120-127. doi:10.1016/s0109-5641(99)00021-4
43. Shintani H, Inoue T, Yamaki M. Analysis of camphorquinone in visible light-cured composite resins. *Dent Mater*. 1985;1(4):124-126. doi:10.1016/s0109-5641(85)80002-6

44. Hassan M, Asghar M, Din SU, Zafar MS. Thermoset polymethacrylate-based materials for dental applications. In: *Materials for Biomedical Engineering*. Elsevier; 2019:273-308. doi:10.1016/B978-0-12-816874-5.00008-6
45. Ogle RE, Sorensen SE, Lewis EA. A new visible light-cured resin system applied to removable prosthodontics. *J Prosthet Dent*. 1986;56(4):497-506. doi:10.1016/0022-3913(86)90397-5
46. Memon MS, Yunus N, Razak AA. Some mechanical properties of a highly cross-linked, microwave-polymerized, injection-molded denture base polymer. *Int J Prosthodont*. 2001;14(3):214-218.
47. Cianci C, Pappalettera G, Renna G, et al. Mechanical Behavior of PET-G Tooth Aligners Under Cyclic Loading. *Front Mater*. 2020;7. doi:10.3389/fmats.2020.00104
48. Ahn HW, Kim KA, Kim SH. A new type of clear orthodontic retainer incorporating multi-layer hybrid materials. *Korean J Orthod*. 2015;45(5):268-272. doi:10.4041/kjod.2015.45.5.268
49. Carvalho V, Soares P, Verissimo C, Pessoa R, Versluis A, Soares C. Mouthguard Biomechanics for Protecting Dental Implants from Impact: Experimental and Finite Element Impact Analysis. *Int J Oral Maxillofac Implants*. 2018;33(2):335-343. doi:10.11607/jomi.5803
50. Sousa AM, Pinho AC, Messias A, Piedade AP. Present Status in Polymeric Mouthguards. A Future Area for Additive Manufacturing? *Polymers (Basel)*. 2020;12(7):1490. doi:10.3390/polym12071490
51. Mizuhashi F, Koide K, Watarai Y. Fabrication of vacuum-formed and pressure-formed mouthguards using polyolefin sheet. *Int J Burns Trauma*. 2020;10(6):345-351.
52. Luo C, Liu Y, Peng B, et al. PEEK for Oral Applications: Recent Advances in Mechanical and Adhesive Properties. *Polymers (Basel)*. 2023;15(2):386. doi:10.3390/polym15020386
53. Liu Y, Fang M, Zhao R, et al. Clinical Applications of Polyetheretherketone in Removable Dental Prostheses: Accuracy, Characteristics, and Performance. *Polymers (Basel)*. 2022;14(21):4615. doi:10.3390/polym14214615
54. Papathanasiou I, Kamposiora P, Papavasiliou G, Ferrari M. The use of PEEK in digital prosthodontics: A narrative review. *BMC Oral Health*. 2020;20(1):217. doi:10.1186/s12903-020-01202-7

55. Edelhoff D, Schweiger J, Prandtner O, Trimpl J, Stimmelmayr M, Güth JF. CAD/CAM splints for the functional and esthetic evaluation of newly defined occlusal dimensions. *Quintessence Int*. 2017;48(3):181-191. doi:10.3290/j.qi.a37641
56. Gibreel M, Perea-Lowery L, Vallittu PK, Garoushi S, Lassila L. Two-body wear and surface hardness of occlusal splint materials. *Dent Mater J*. 2022;41(6):916-922. doi:10.4012/dmj.2022-100
57. Fasbinder DJ. Digital dentistry: innovation for restorative treatment. *Compend Contin Educ Dent*. 2010;31 Spec No 4:2-11; quiz 12.
58. Beuer F, Schweiger J, Edelhoff D. Digital dentistry: an overview of recent developments for CAD/CAM generated restorations. *Br Dent J*. 2008;204(9):505-511. doi:10.1038/sj.bdj.2008.350
59. Sulaiman TA. Materials in digital dentistry—A review. *Journal of Esthetic and Restorative Dentistry*. 2020;32(2):171-181. doi:10.1111/jerd.12566
60. Bae EJ, Jeong ID, Kim WC, Kim JH. A comparative study of additive and subtractive manufacturing for dental restorations. *J Prosthet Dent*. 2017;118(2):187-193. doi:10.1016/j.prosdent.2016.11.004
61. Mörmann WH, Brandestini M, Lutz F, Barbakow F. Chairside computer-aided direct ceramic inlays. *Quintessence Int*. 1989;20(5):329-339.
62. Davidowitz G, Kotick PG. The Use of CAD/CAM in Dentistry. *Dent Clin North Am*. 2011;55(3):559-570. doi:10.1016/j.cden.2011.02.011
63. Henkel GL. A comparison of fixed prostheses generated from conventional vs digitally scanned dental impressions. *Compend Contin Educ Dent*. 2007;28(8):422-424, 426-428, 430-431.
64. Kang SY, Lee HN, Kim JH, Kim WC. Evaluation of marginal discrepancy of pressable ceramic veneer fabricated using CAD/CAM system: Additive and subtractive manufacturing. *J Adv Prosthodont*. 2018;10(5):347-353. doi:10.4047/jap.2018.10.5.347
65. Kanazawa M, Inokoshi M, Minakuchi S, Ohbayashi N. Trial of a CAD/CAM system for fabricating complete dentures. *Dent Mater J*. 2011;30(1):93-96. doi:10.4012/dmj.2010-112
66. Baba NZ, Goodacre BJ, Goodacre CJ, Müller F, Wagner S. CAD/CAM Complete Denture Systems and Physical Properties: A Review of the Literature. *Journal of Prosthodontics*. 2021;30(S2):113-124. doi:10.1111/jopr.13243

67. Goodacre CJ, Garbacea A, Naylor WP, Daher T, Marchack CB, Lowry J. CAD/CAM fabricated complete dentures: concepts and clinical methods of obtaining required morphological data. *J Prosthet Dent*. 2012;107(1):34-46. doi:10.1016/S0022-3913(12)60015-8
68. Lutz AM, Hampe R, Roos M, Lümkemann N, Eichberger M, Stawarczyk B. Fracture resistance and 2-body wear of 3-dimensional-printed occlusal devices. *J Prosthet Dent*. 2019;121(1):166-172. doi:10.1016/j.prosdent.2018.04.007
69. Kollenberg W. Keramik und Multi-Material 3D-Druck. *Keramische Zeitschrift*. 2014;66(4):233-236. doi:10.1007/BF03400217
70. Schweiger J, Edelhoff D, Güth JF. 3D Printing in Digital Prosthetic Dentistry: An Overview of Recent Developments in Additive Manufacturing. *J Clin Med*. 2021;10(9):2010. doi:10.3390/jcm10092010
71. nternational Organization for Standardization. ISO/ASTM52900:2015 (ASTM F2792). Additive manufacturing—generalprinciples and terminology. https://www-iso-org.offcampus.lib.washington.edu/standard/69669.html?browse=tc.
72. Alammar A, Kois JC, Revilla-León M, Att W. Additive Manufacturing Technologies: Current Status and Future Perspectives. *Journal of Prosthodontics*. 2022;31(S1):4-12. doi:10.1111/jopr.13477
73. Piedra-Cascón W, Krishnamurthy VR, Att W, Revilla-León M. 3D printing parameters, supporting structures, slicing, and post-processing procedures of vat-polymerization additive manufacturing technologies: A narrative review. *J Dent*. 2021;109:103630. doi:10.1016/j.jdent.2021.103630
74. ISO/ASTM Additive Manufacturing—General Principles—Part 2: Overview of Process Categories and Feedstock. *Beuth*. 2016;17296(2).
75. Revilla-León M, Özcan M. Additive Manufacturing Technologies Used for Processing Polymers: Current Status and Potential Application in Prosthetic Dentistry. *J Prosthodont*. 2019;28(2):146-158. doi:10.1111/jopr.12801
76. Revilla-León M, Sadeghpour M, Özcan M. An update on applications of 3D printing technologies used for processing polymers used in implant dentistry. *Odontology*. 2020;108(3):331-338. doi:10.1007/s10266-019-00441-7
77. Alharbi N, Osman R, Wismeijer D. Effects of build direction on the mechanical properties of 3D-printed complete coverage interim dental restorations. *J Prosthet Dent*. 2016;115(6):760-767. doi:10.1016/j.prosdent.2015.12.002

78. Kim GB, Lee S, Kim H, et al. Three-Dimensional Printing: Basic Principles and Applications in Medicine and Radiology. *Korean J Radiol*. 2016;17(2):182-197. doi:10.3348/kjr.2016.17.2.182
79. Emami MM, Barazandeh F, Yaghmaie F. Scanning-projection based stereolithography: Method and structure. *Sens Actuators A Phys*. 2014;218:116-124. doi:10.1016/j.sna.2014.08.002
80. Stansbury JW, Idacavage MJ. 3D printing with polymers: Challenges among expanding options and opportunities. *Dent Mater*. 2016;32(1):54-64. doi:10.1016/j.dental.2015.09.018
81. Hazeveld A, Huddleston Slater JJR, Ren Y. Accuracy and reproducibility of dental replica models reconstructed by different rapid prototyping techniques. *Am J Orthod Dentofacial Orthop*. 2014;145(1):108-115. doi:10.1016/j.ajodo.2013.05.011
82. Bogue R. 3D printing: the dawn of a new era in manufacturing? *Assembly Automation*. 2013;33(4):307-311. doi:10.1108/AA-06-2013-055
83. Monzón MD, Paz R, Pei E, et al. 4D printing: processability and measurement of recovery force in shape memory polymers. *The International Journal of Advanced Manufacturing Technology*. 2017;89(5-8):1827-1836. doi:10.1007/s00170-016-9233-9
84. Tumbleston JR, Shirvanyants D, Ermoshkin N, et al. Continuous liquid interface production of 3D objects. *Science (1979)*. 2015;347(6228):1349-1352. doi:10.1126/science.aaa2397
85. Wong K V., Hernandez A. A Review of Additive Manufacturing. *ISRN Mechanical Engineering*. 2012;2012:1-10. doi:10.5402/2012/208760
86. Gao W, Zhang Y, Nazzetta DC, Ramani K, Cipra RJ. RevoMaker. In: *Proceedings of the 28th Annual ACM Symposium on User Interface Software & Technology*. New York, NY, USA: ACM; 2015:437-446. doi:10.1145/2807442.2807476
87. Guo C, Zhang M, Bhandari B. Model Building and Slicing in Food 3D Printing Processes: A Review. *Compr Rev Food Sci Food Saf*. 2019;18(4):1052-1069. doi:10.1111/1541-4337.12443
88. Pham DT, Gault RS. A comparison of rapid prototyping technologies. *Int J Mach Tools Manuf*. 1998;38(10-11):1257-1287. doi:10.1016/S0890-6955(97)00137-5

89. Fass R, Boeckxstaens GE, El-Serag H, Rosen R, Sifrim D, Vaezi MF. Gastro-oesophageal reflux disease. *Nat Rev Dis Primers*. 2021;7(1):55. doi:10.1038/s41572-021-00287-w
90. Clarrett DM, Hachem C. Gastroesophageal Reflux Disease (GERD). *Mo Med*. 2018;115(3):214-218.
91. Ferriolli E, Oliveira RB, Matsuda NM, Braga FJHN, Dantas RO. Aging, Esophageal Motility, and Gastroesophageal Reflux. *J Am Geriatr Soc*. 1998;46(12):1534-1537. doi:10.1111/j.1532-5415.1998.tb01538.x
92. Zheng Z, Nordenstedt H, Pedersen NL, Lagergren J, Ye W. Lifestyle Factors and Risk for Symptomatic Gastroesophageal Reflux in Monozygotic Twins. *Gastroenterology*. 2007;132(1):87-95. doi:10.1053/j.gastro.2006.11.019
93. Chhabra P, Ingole N. Gastroesophageal Reflux Disease (GERD): Highlighting Diagnosis, Treatment, and Lifestyle Changes. *Cureus*. August 2022. doi:10.7759/cureus.28563
94. Ronkainen J, Aro P, Storskrubb T, et al. High prevalence of gastroesophageal reflux symptoms and esophagitis with or without symptoms in the general adult Swedish population: A Kalixanda study report. *Scand J Gastroenterol*. 2005;40(3):275-285. doi:10.1080/00365520510011579
95. Richter JE. Gi Problems in Geriatric Patients. *American Journal of Gastroenterology*. 2000;95(2):368-373. doi:10.1111/j.1572-0241.2000.t01-1-01791.x
96. Martin Z, Spry G, Hoult J, et al. What is the efficacy of dietary, nutraceutical, and probiotic interventions for the management of gastroesophageal reflux disease symptoms? A systematic literature review and meta-analysis. *Clin Nutr ESPEN*. 2022;52:340-352. doi:10.1016/j.clnesp.2022.09.015
97. Zerbib F, Roman S, Ropert A, et al. Esophageal pH-Impedance Monitoring and Symptom Analysis in GERD: A Study in Patients off and on Therapy. *Am J Gastroenterol*. 2006;101(9):1956-1963. doi:10.1111/j.1572-0241.2006.00711.x
98. Agrawal A, Tutuian R, Hila A, Freeman J, Castell DO. Ingestion of Acidic Foods Mimics Gastroesophageal Reflux During pH Monitoring. *Dig Dis Sci*. 2005;50(10):1916-1920. doi:10.1007/s10620-005-2961-6
99. Sood S, Bhat V, Prasad Dk, Bhat A. Role of colors in prosthodontics: Application of color science in restorative dentistry. *Indian Journal of Dental Research*. 2011;22(6):804. doi:10.4103/0970-9290.94675

100. Heintze SD, Cavalleri A, Forjanic M, Zellweger G, Rousson V. Wear of ceramic and antagonist--a systematic evaluation of influencing factors in vitro. *Dent Mater*. 2008;24(4):433-449. doi:10.1016/j.dental.2007.06.016
101. Park JH, Park S, Lee K, Yun KD, Lim HP. Antagonist wear of three CAD/CAM anatomic contour zirconia ceramics. *J Prosthet Dent*. 2014;111(1):20-29. doi:10.1016/j.prosdent.2013.06.002
102. Ahmad R, Wu BM, Morgano SM. Polishing Mechanism And Its Effect On The Mechanical Properties Of Ceramic Restorations - A Review Of The Literature. *Ann Dent*. 2001;8(1):57-61. doi:10.22452/adum.vol8no1.10
103. Flury S, Peutzfeldt A, Lussi A. Influence of Surface Roughness on Mechanical Properties of Two Computer-aided Design/Computer-aided Manufacturing (CAD/CAM) Ceramic Materials. *Oper Dent*. 2012;37(6):617-624. doi:10.2341/11-391-L
104. Field J, Waterhouse P, German M. Quantifying and qualifying surface changes on dental hard tissues in vitro. *J Dent*. 2010;38(3):182-190. doi:10.1016/j.jdent.2010.01.002
105. Ccahuana VZS, Ozcan M, Mesquita AMM, Nishioka RS, Kimpara ET, Bottino MA. Surface degradation of glass ceramics after exposure to acidulated phosphate fluoride. *J Appl Oral Sci*. 2010;18(2):155-165. doi:10.1590/s1678-77572010000200010
106. Flury S, Lussi A, Zimmerli B. Performance of different polishing techniques for direct CAD/CAM ceramic restorations. *Oper Dent*. 2010;35(4):470-481. doi:10.2341/09-373-L
107. Al-Shammery HAO, Bubb NL, Youngson CC, Fasbinder DJ, Wood DJ. The use of confocal microscopy to assess surface roughness of two milled CAD-CAM ceramics following two polishing techniques. *Dent Mater*. 2007;23(6):736-741. doi:10.1016/j.dental.2006.06.012
108. Heurich E, Beyer M, Jandt KD, et al. Quantification of dental erosion—A comparison of stylus profilometry and confocal laser scanning microscopy (CLSM). *Dental Materials*. 2010;26(4):326-336. doi:10.1016/j.dental.2009.12.001
109. Whitehead SA, Shearer AC, Watts DC, Wilson NHF. Comparison of two stylus methods for measuring surface texture. *Dental Materials*. 1999;15(2):79-86. doi:10.1016/S0109-5641(99)00017-2

110. Paepegaey AM, Barker ML, Bartlett DW, et al. Measuring enamel erosion: A comparative study of contact profilometry, non-contact profilometry and confocal laser scanning microscopy. *Dental Materials*. 2013;29(12):1265-1272. doi:10.1016/j.dental.2013.09.015
111. Fuzzi M, Zaccheroni Z, Vallania G. Scanning electron microscopy and profilometer evaluation of glazed and polished dental porcelain. *Int J Prosthodont*. 1996;9(5):452-458.
112. Al-Hiyasat AS, Saunders WP, Sharkey SW, Smith GMcR, Gilmour WH. Investigation of human enamel wear against four dental ceramics and gold. *J Dent*. 1998;26(5-6):487-495. doi:10.1016/S0300-5712(97)00041-9
113. Attin T, Wegehaupt FJ. Methods for Assessment of Dental Erosion. In: ; 2014:123-142. doi:10.1159/000360355
114. Holme B, Hove LH, Tveit AB. Using white light interferometry to measure etching of dental enamel. *Measurement*. 2005;38(2):137-147. doi:10.1016/j.measurement.2005.04.003
115. Larimer C, Suter JD, Bonheyo G, Addleman RS. In situ non-destructive measurement of biofilm thickness and topology in an interferometric optical microscope. *J Biophotonics*. 2016;9(6):656-666. doi:10.1002/jbio.201500212
116. Jandt KD. Atomic force microscopy of biomaterials surfaces and interfaces. *Surf Sci*. 2001;491(3):303-332. doi:10.1016/S0039-6028(01)01296-1
117. Barbour ME, Rees JS. The laboratory assessment of enamel erosion: a review. *J Dent*. 2004;32(8):591-602. doi:10.1016/j.jdent.2004.05.001
118. Amaya-Pajares SP, Ritter A V., Vera Resendiz C, Henson BR, Culp L, Donovan TE. Effect of Finishing and Polishing on the Surface Roughness of Four Ceramic Materials after Occlusal Adjustment. *Journal of Esthetic and Restorative Dentistry*. 2016;28(6):382-396. doi:10.1111/jerd.12222
119. Zhang F, Inokoshi M, Batuk M, et al. Strength, toughness and aging stability of highly-translucent Y-TZP ceramics for dental restorations. *Dent Mater*. 2016;32(12):e327-e337. doi:10.1016/j.dental.2016.09.025
120. Oliveira-Junior OB, Buso L, Fujiy FH, et al. Influence of polishing procedures on the surface roughness of dental ceramics made by different techniques. *Gen Dent*. 2013;61(1):e4-8.

121. Fuzzi M, Zaccheroni Z, Vallania G. Scanning electron microscopy and profilometer evaluation of glazed and polished dental porcelain. *Int J Prosthodont*. 1996;9(5):452-458.
122. Milleding P, Karlsson S, Nyborg L. On the surface elemental composition of non-corroded and corroded dental ceramic materials in vitro. *J Mater Sci Mater Med*. 2003;14(6):557-566. doi:10.1023/a:1023416232222
123. Talbot MJ, White RG. Cell surface and cell outline imaging in plant tissues using the backscattered electron detector in a variable pressure scanning electron microscope. *Plant Methods*. 2013;9(1):40. doi:10.1186/1746-4811-9-40
124. Nguyen JNT, Harbison AM. Scanning Electron Microscopy Sample Preparation and Imaging. *Methods Mol Biol*. 2017;1606:71-84. doi:10.1007/978-1-4939-6990-6_5
125. Valian A, Moravej-Salehi E. Surface treatment of feldspathic porcelain: scanning electron microscopy analysis. *J Adv Prosthodont*. 2014;6(5):387-394. doi:10.4047/jap.2014.6.5.387
126. Moravej-Salehi E, Moravej-Salehi E, Valian A. Surface topography and bond strengths of feldspathic porcelain prepared using various sandblasting pressures. *J Investig Clin Dent*. 2016;7(4):347-354. doi:10.1111/jicd.12171
127. Magalhães AC, Romanelli AC, Rios D, et al. Effect of a single application of TiF4 and NaF varnishes and solutions combined with Nd:YAG laser irradiation on enamel erosion in vitro. *Photomed Laser Surg*. 2011;29(8):537-544. doi:10.1089/pho.2010.2886
128. Torres CP, Chinelatti MA, Gomes-Silva JM, et al. Surface and subsurface erosion of primary enamel by acid beverages over time. *Braz Dent J*. 2010;21(4):337-345. doi:10.1590/s0103-64402010000400009
129. Miura H, Miura K, Mizugai H, Arai Y, Umenai T, Isogai E. Chewing ability and quality of life among the elderly residing in a rural community in Japan. *J Oral Rehabil*. 2000;27(8):731-734. doi:10.1046/j.1365-2842.2000.00590.x
130. Watts MW, Tan EK, Jankovic J. Bruxism and cranial-cervical dystonia: is there a relationship? *Cranio*. 1999;17(3):196-201. doi:10.1080/08869634.1999.11746095
131. Marquis PM, Hu X, Shortall AC. Two-body wear of dental composites under different loads. *Int J Prosthodont*. 2000;13(6):473-479.
132. Söderholm KJ, Richards ND. Wear resistance of composites: a solved problem? *Gen Dent*. 1998;46(3):256-263; quiz 264-265.

133. Casey J, Dunn WJ, Wright E. In vitro wear of various orthotic device materials. *J Prosthet Dent*. 2003;90(5):498-502. doi:10.1016/s0022-3913(03)00545-6
134. Gupta BR. Friction and wear mechanism of polymers, their composites and nanocomposites. In: *Tribology of Polymers, Polymer Composites, and Polymer Nanocomposites*. Elsevier; 2023:51-117. doi:10.1016/B978-0-323-90748-4.00012-1
135. Tooth Wear. *Int Dent J*. 2024;74(1):163-164. doi:10.1016/j.identj.2023.10.007
136. Yip KHK, Smales RJ, Kaidonis JA. Differential wear of teeth and restorative materials: clinical implications. *Int J Prosthodont*. 2004;17(3):350-356.
137. Turssi CP, de Moraes Purquerio B, Serra MC. Wear of dental resin composites: Insights into underlying processes and assessment methods—A review. *J Biomed Mater Res B Appl Biomater*. 2003;65B(2):280-285. doi:10.1002/jbm.b.10563
138. Turssi CP, Hara AT, Magalhães CS de, Serra MC, Rodrigues AL. Influence of storage regime prior to abrasion on surface topography of restorative materials. *J Biomed Mater Res B Appl Biomater*. 2003;65B(2):227-232. doi:10.1002/jbm.b.10005
139. de Gee A. Influence of enzymes and plaque acids on in vitro wear of dental composites. *Biomaterials*. 1996;17(13):1327-1332. doi:10.1016/0142-9612(96)88679-0
140. de Gee AJ, van Duinen RNB, Werner A, Davidson CL. Early and Long-term Wear of Conventional and Resin-modified Glass Ionomers. *J Dent Res*. 1996;75(8):1613-1619. doi:10.1177/00220345960750081401
141. Schuurs A. *Pathology of the Hard Dental Tissues*. Wiley; 2012. doi:10.1002/9781118702659
142. Jones RRH, Cleaton-Jones P. Depth and Area of Dental Erosions, and Dental Caries, in Bulimic Women. *J Dent Res*. 1989;68(8):1275-1278. doi:10.1177/00220345890680081201
143. Schroeder PL. Dental Erosion and Acid Reflux Disease. *Ann Intern Med*. 1995;122(11):809. doi:10.7326/0003-4819-122-11-199506010-00001
144. Asher C, Read MJ. Early enamel erosion in children associated with the excessive consumption of citric acid. *Br Dent J*. 1987;162(10):384-387. doi:10.1038/sj.bdj.4806141
145. Barron RP, Carmichael RP, Marcon MA, Sàndor GKB. Dental erosion in gastroesophageal reflux disease. *J Can Dent Assoc*. 2003;69(2):84-89.

146. Schlueter N, Ganss C, Hardt M, Schegietz D, Klimek J. Effect of pepsin on erosive tissue loss and the efficacy of fluoridation measures in dentine *in vitro*. *Acta Odontol Scand*. 2007;65(5):298-305. doi:10.1080/00016350701678733
147. Bitencourt SB, Catanoze IA, da Silva EVF, et al. Effect of acidic beverages on surface roughness and color stability of artificial teeth and acrylic resin. *J Adv Prosthodont*. 2020;12(2):55. doi:10.4047/jap.2020.12.2.55
148. Bettencourt AF, Neves CB, de Almeida MS, et al. Biodegradation of acrylic based resins: A review. *Dental Materials*. 2010;26(5):e171-e180. doi:10.1016/j.dental.2010.01.006
149. ARIMA T, MURATA H, HAMAD T. The effects of cross-linking agents on the water sorption and solubility characteristics of denture base resin. *J Oral Rehabil*. 1996;23(7):476-480. doi:10.1111/j.1365-2842.1996.tb00882.x
150. Dixon DL, Breeding LC, Ekstrand KG. Linear dimensional variability of three denture base resins after processing and in water storage. *J Prosthet Dent*. 1992;68(1):196-200. doi:10.1016/0022-3913(92)90304-S
151. Göpferich A. Mechanisms of polymer degradation and erosion. *Biomaterials*. 1996;17(2):103-114. doi:10.1016/0142-9612(96)85755-3
152. Akova T, Ozkomur A, Uysal H. Effect of food-simulating liquids on the mechanical properties of provisional restorative materials. *Dental Materials*. 2006;22(12):1130-1134. doi:10.1016/j.dental.2005.09.009
153. Prakki A, Cilli R, Mondelli RFL, Kalachandra S, Pereira JC. Influence of pH environment on polymer based dental material properties. *J Dent*. 2005;33(2):91-98. doi:10.1016/j.jdent.2004.08.004
154. Yap A. Influence of dietary simulating solvents on the hardness of provisional restorative materials. *Dental Materials*. 2004;20(4):370-376. doi:10.1016/j.dental.2003.06.001
155. Colombo M, Poggio C, Lasagna A, Chiesa M, Scribante A. Vickers Micro-Hardness of New Restorative CAD/CAM Dental Materials: Evaluation and Comparison after Exposure to Acidic Drink. *Materials*. 2019;12(8):1246. doi:10.3390/ma12081246
156. Koda T, Tsuchiya H, Yamauchi M, Ohtani S, Takagi N, Kawano J. Leachability of denture-base acrylic resins in artificial saliva. *Dental Materials*. 1990;6(1):13-16. doi:10.1016/0109-5641(90)90037-F

157. Tseng CC, Lin PY, Kirankumar R, Chuang ZW, Wu IH, Hsieh S. Surface degradation effects of carbonated soft drink on a resin based dental compound. *Heliyon*. 2021;7(3):e06400. doi:10.1016/j.heliyon.2021.e06400
158. Backer AD, Münchow EA, Eckert GJ, Hara AT, Platt JA, Bottino MC. Effects of Simulated Gastric Juice on CAD/CAM Resin Composites—Morphological and Mechanical Evaluations. *Journal of Prosthodontics*. 2017;26(5):424-431. doi:10.1111/jopr.12420
159. Donovan T, Nguyen-Ngoc C, Abd Alraheam I, Irusa K. Contemporary diagnosis and management of dental erosion. *Journal of Esthetic and Restorative Dentistry*. 2021;33(1):78-87. doi:10.1111/jerd.12706
160. Alencar MF, Pereira MT, De-Moraes MDR, Santiago SL, Passos VF. The effects of intrinsic and extrinsic acids on nanofilled and bulk fill resin composites: Roughness, surface hardness, and scanning electron microscopy analysis. *Microsc Res Tech*. 2020;83(2):202-207. doi:10.1002/jemt.23403
161. Habsha E. The etiology and pathogenesis of tooth wear: Part I. In: *Oral Health*. Vol 10. ; 1999:83-92.
162. Lazarchik D. Dental erosion: predominant oral lesion in gastroesophageal reflux disease. *Am J Gastroenterol*. 2000;95(1):S33-S38. doi:10.1016/S0002-9270(00)01076-5
163. Bartlett D, Smith B. Clinical investigations of gastro-oesophageal reflux: Part 1. *Dent Update*. 1996;23(5):205-208.
164. Newton JL, James OFW, Williams G V, Allen A. The diurnal profile of gastric pepsin activity is reduced with Helicobacter pylori infection. *Dig Dis Sci*. 2004;49(7-8):1103-1108. doi:10.1023/b:ddas.0000037795.92727.9f
165. Kleter GA, Damen JJ, Buijs MJ, Ten Cate JM. The Maillard reaction in demineralized dentin in vitro. *Eur J Oral Sci*. 1997;105(3):278-284. doi:10.1111/j.1600-0722.1997.tb00212.x
166. Tonami K ichi, Ericson D. Protein profile of pepsin-digested carious and sound human dentine. *Acta Odontol Scand*. 2005;63(1):17-20. doi:10.1080/00016350410010090
167. Ganss C, Klimek J, Starck C. Quantitative analysis of the impact of the organic matrix on the fluoride effect on erosion progression in human dentine using longitudinal microradiography. *Arch Oral Biol*. 2004;49(11):931-935. doi:10.1016/j.archoralbio.2004.05.010

168. Pace F, Pallotta S, Tonini M, Vakil N, Bianchi Porro G. Systematic review: gastro-oesophageal reflux disease and dental lesions. *Aliment Pharmacol Ther*. 2008;27(12):1179-1186. doi:10.1111/j.1365-2036.2008.03694.x
169. Sulaiman TA, Abdulmajeed AA, Shahramian K, et al. Impact of gastric acidic challenge on surface topography and optical properties of monolithic zirconia. *Dent Mater*. 2015;31(12):1445-1452. doi:10.1016/j.dental.2015.09.010
170. Harryparsad A, Dullabh H, Sykes L, Herbst D. The effects of hydrochloric acid on all-ceramic restorative materials: an in-vitro study. *SADJ*. 2014;69(3):106-111.
171. HEINTZE SD, REICHL FX, HICKEL R. Wear of dental materials: Clinical significance and laboratory wear simulation methods —A review. *Dent Mater J*. 2019;38(3):343-353. doi:10.4012/dmj.2018-140
172. Eisenburger M, Shellis RP, Addy M. Comparative Study of Wear of Enamel Induced by Alternating and Simultaneous Combinations of Abrasion and Erosion in vitro. *Caries Res*. 2003;37(6):450-455. doi:10.1159/000073399
173. Tanoue N, Matsumura H, Atsuta M. Analysis of composite type and different sources of polymerization light on in vitro toothbrush/dentifrice abrasion resistance. *J Dent*. 2000;28(5):355-359. doi:10.1016/S0300-5712(00)00014-2
174. Momoi Y, Hirosaki K, Kohno A, McCabe JF. In vitro toothbrush-dentifrice abrasion of resin-modified glass ionomers. *Dental Materials*. 1997;13(2):82-88. doi:10.1016/S0109-5641(97)80016-4
175. Piddock V, Qualtrough AJE. Dental ceramics—an update. *J Dent*. 1990;18(5):227-235. doi:10.1016/0300-5712(90)90019-B
176. Chitwadgi R, Siddesh B, Shankar BL, Suresh R, Siddeshkumar NG. Optimization and analysis of dry sliding wear behaviour of N-B4C/MOS2 unreinforced AA2219 nano hybrid composites using response surface methodology. *Metallurgical and Materials Engineering*. 2022;28(3):469-485. doi:10.30544/840
177. Paulino MR, Alves LR, Gurgel BC V, Calderon PS. Simplified versus traditional techniques for complete denture fabrication: a systematic review. *J Prosthet Dent*. 2015;113(1):12-16. doi:10.1016/j.prosdent.2014.08.004
178. Quirynen M, Bollen CM, Papaioannou W, Van Eldere J, van Steenberghe D. The influence of titanium abutment surface roughness on plaque accumulation and gingivitis: short-term observations. *Int J Oral Maxillofac Implants*. 1996;11(2):169-178.

179. Nishigawa K, Bando E, Nakano M. Quantitative study of bite force during sleep associated bruxism. *J Oral Rehabil*. 2001;28(5):485-491. doi:10.1046/j.1365-2842.2001.00692.x
180. Calderon P dos S, Kogawa EM, Lauris JRP, Conti PCR. The influence of gender and bruxism on the human maximum bite force. *Journal of Applied Oral Science*. 2006;14(6):448-453. doi:10.1590/S1678-77572006000600011
181. Ferrario VF, Sforza C, Zanotti G, Tartaglia GM. Maximal bite forces in healthy young adults as predicted by surface electromyography. *J Dent*. 2004;32(6):451-457. doi:10.1016/j.jdent.2004.02.009
182. de Abreu RAM, Pereira MD, Furtado F, Prado GPR, Mestriner W, Ferreira LM. Masticatory efficiency and bite force in individuals with normal occlusion. *Arch Oral Biol*. 2014;59(10):1065-1074. doi:10.1016/j.archoralbio.2014.05.005
183. Oh W suck, DeLong R, Anusavice KJ. Factors affecting enamel and ceramic wear: A literature review. *J Prosthet Dent*. 2002;87(4):451-459. doi:10.1067/mpr.2002.123851
184. Lambrechts P, Braem M, Vuylsteke-Wauters M, Vanherle G. Quantitative in vivo Wear of Human Enamel. *J Dent Res*. 1989;68(12):1752-1754. doi:10.1177/00220345890680120601
185. XHONGA FA. Bruxism and its effect on the teeth. *J Oral Rehabil*. 1977;4(1):65-76. doi:10.1111/j.1365-2842.1977.tb00967.x
186. Gibbs CH, Mahan PE, Mauderli A, Lundeen HC, Walsh EK. Limits of human bite strength. *J Prosthet Dent*. 1986;56(2):226-229. doi:10.1016/0022-3913(86)90480-4
187. Rosar JV, Barbosa T de S, Dias IOV, et al. Effect of interocclusal appliance on bite force, sleep quality, salivary cortisol levels and signs and symptoms of temporomandibular dysfunction in adults with sleep bruxism. *Arch Oral Biol*. 2017;82:62-70. doi:10.1016/j.archoralbio.2017.05.018
188. Domagała I, Przystupa K, Firlej M, et al. Analysis of the Statistical Comparability of the Hardness and Wear of Polymeric Materials for Orthodontic Applications. *Materials*. 2021;14(11):2925. doi:10.3390/ma14112925
189. Pizolato RA, Gavião MBD, Berretin-Felix G, Sampaio ACM, Trindade Junior AS. Maximal bite force in young adults with temporomandibular disorders and bruxism. *Braz Oral Res*. 2007;21(3):278-283. doi:10.1590/s1806-83242007000300015

190. Gibbs CH, Mahan PE, Lundeen HC, Brehnan K, Walsh EK, Holbrook WB. Occlusal forces during chewing and swallowing as measured by sound transmission. *J Prosthet Dent*. 1981;46(4):443-449. doi:10.1016/0022-3913(81)90455-8
191. Gil-Martínez A, Paris-Alemany A, López-de-Uralde-Villanueva I, La Touche R. Management of pain in patients with temporomandibular disorder (TMD): challenges and solutions. *J Pain Res*. 2018;11:571-587. doi:10.2147/JPR.S127950
192. Benli M, Olson J, Huck O, Özcan M. A novel treatment modality for myogenous temporomandibular disorders using aromatherapy massage with lavender oil: A randomized controlled clinical trial. *Cranio*. 2023;41(1):48-58. doi:10.1080/08869634.2020.1819067
193. Riley P, Glenny AM, Worthington H V, et al. Oral splints for temporomandibular disorder or bruxism: a systematic review. *Br Dent J*. 2020;228(3):191-197. doi:10.1038/s41415-020-1250-2
194. Osiewicz MA, Werner A, Roeters FJM, Kleverlaan CJ. Effects of occlusal splint therapy on opposing tooth tissues, filling materials and restorations. *J Oral Rehabil*. 2021;48(10):1129-1134. doi:10.1111/joor.13235
195. Huettig F, Kustermann A, Kuscu E, Geis-Gerstorfer J, Spintzyk S. Polishability and wear resistance of splint material for oral appliances produced with conventional, subtractive, and additive manufacturing. *J Mech Behav Biomed Mater*. 2017;75:175-179. doi:10.1016/j.jmbbm.2017.07.019
196. Ekberg E, Nilner M. Treatment outcome of appliance therapy in temporomandibular disorder patients with myofascial pain after 6 and 12 months. *Acta Odontol Scand*. 2004;62(6):343-349. doi:10.1080/00016350410010063
197. Taraszewska A. Risk factors for gastroesophageal reflux disease symptoms related to lifestyle and diet. *Rocz Panstw Zakl Hig*. 2021;72(1):21-28. doi:10.32394/rpzh.2021.0145
198. Ohmure H, Kanematsu-Hashimoto K, Nagayama K, et al. Evaluation of a Proton Pump Inhibitor for Sleep Bruxism: A Randomized Clinical Trial. *J Dent Res*. 2016;95(13):1479-1486. doi:10.1177/0022034516662245
199. On ZX, Grant J, Shi Z, et al. The association between gastroesophageal reflux disease with sleep quality, depression, and anxiety in a cohort study of Australian men. *J Gastroenterol Hepatol*. 2017;32(6):1170-1177. doi:10.1111/jgh.13650

200. Fujiwara Y, Arakawa T, Fass R. Gastroesophageal reflux disease and sleep. *Gastroenterol Clin North Am*. 2013;42(1):57-70. doi:10.1016/j.gtc.2012.11.011
201. Nandurkar S, Talley NJ. Epidemiology and natural history of reflux disease. *Baillieres Best Pract Res Clin Gastroenterol*. 2000;14(5):743-757. doi:10.1053/bega.2000.0122
202. Locke GR, Talley NJ, Fett SL, Zinsmeister AR, Melton LJ. Prevalence and clinical spectrum of gastroesophageal reflux: a population-based study in Olmsted County, Minnesota. *Gastroenterology*. 1997;112(5):1448-1456. doi:10.1016/s0016-5085(97)70025-8
203. Bloom BS, Glise H. What do we know about gastroesophageal reflux disease? *Am J Gastroenterol*. 2001;96(8 Suppl):S1-6. doi:10.1016/s0002-9270(01)02576-x
204. Scheutzel P. Etiology of dental erosion – intrinsic factors. *Eur J Oral Sci*. 1996;104(2):178-190. doi:10.1111/j.1600-0722.1996.tb00066.x
205. Zieliński G, Pająk A, Wójcicki M. Global Prevalence of Sleep Bruxism and Awake Bruxism in Pediatric and Adult Populations: A Systematic Review and Meta-Analysis. *J Clin Med*. 2024;13(14). doi:10.3390/jcm13144259
206. Miyawaki S, Lavigne GJ, Pierre M, Guitard F, Montplaisir JY, Kato T. Association between sleep bruxism, swallowing-related laryngeal movement, and sleep positions. *Sleep*. 2003;26(4):461-465.
207. Grymak A, Aarts JM, Ma S, Waddell JN, Choi JJE. Wear Behavior of Occlusal Splint Materials Manufactured By Various Methods: A Systematic Review. *J Prosthodont*. 2022;31(6):472-487. doi:10.1111/jopr.13432
208. Gupt P, Nagpal A, Samra RK, Verma R, Kaur J, Abrol S. A comparative study to check fracture strength of provisional fixed partial dentures made of autopolymerizing polymethylmethacrylate resin reinforced with different materials: An in vitro study. *J Indian Prosthodont Soc*. 2017;17(3):301-309. doi:10.4103/jips.jips_79_17
209. Singh A, Garg S. Comparative Evaluation of Flexural Strength of Provisional Crown and Bridge Materials-An Invitro Study. *J Clin Diagn Res*. 2016;10(8):ZC72-7. doi:10.7860/JCDR/2016/19582.8291
210. Reyes-Sevilla M, Kuijs RH, Werner A, Kleverlaan CJ, Lobbezoo F. Comparison of wear between occlusal splint materials and resin composite materials. *J Oral Rehabil*. 2018;45(7):539-544. doi:10.1111/joor.12636

211. Gibreel M, Perea-Lowery L, Vallittu PK, Lassila L. Characterization of occlusal splint materials: CAD-CAM versus conventional resins. *J Mech Behav Biomed Mater*. 2021;124:104813. doi:10.1016/j.jmbbm.2021.104813
212. Benli M, Al-Haj Husain N, Ozcan M. Mechanical and chemical characterization of contemporary occlusal splint materials fabricated with different methods: a systematic review. *Clin Oral Investig*. 2023;27(12):7115-7141. doi:10.1007/s00784-023-05360-0
213. Schmeiser F, Baumert U, Stawarczyk B. Two-body wear of occlusal splint materials from subtractive computer-aided manufacturing and three-dimensional printing. *Clin Oral Investig*. 2022;26(9):5857-5866. doi:10.1007/s00784-022-04543-5
214. Lutz AM, Hampe R, Roos M, Lümkemann N, Eichberger M, Stawarczyk B. Fracture resistance and 2-body wear of 3-dimensional-printed occlusal devices. *J Prosthet Dent*. 2019;121(1):166-172. doi:10.1016/j.prosdent.2018.04.007
215. Patzelt SBM, Krügel M, Wesemann C, et al. In Vitro Time Efficiency, Fit, and Wear of Conventionally- versus Digitally-Fabricated Occlusal Splints. *Materials (Basel)*. 2022;15(3). doi:10.3390/ma15031085
216. Han Y, Bai Y, Bian J, Guo X, Liu B, Wang Z. Polydimethylsiloxane-Based Mechanoluminescent Occlusal Splint with the Visualization of Occlusal Force. *ACS Appl Polym Mater*. 2021;3(10):5180-5187. doi:10.1021/acsapm.1c00917
217. Grymak A, Waddell JN, Aarts JM, Ma S, Choi JJE. Evaluation of wear behaviour of various occlusal splint materials and manufacturing processes. *J Mech Behav Biomed Mater*. 2022;126:105053. doi:10.1016/j.jmbbm.2021.105053
218. Heintze SD, Zappini G, Rousson V. Wear of ten dental restorative materials in five wear simulators--results of a round robin test. *Dent Mater*. 2005;21(4):304-317. doi:10.1016/j.dental.2004.05.003
219. Abuzar MA, Bellur S, Duong N, et al. Evaluating surface roughness of a polyamide denture base material in comparison with poly (methyl methacrylate). *J Oral Sci*. 2010;52(4):577-581. doi:10.2334/josnusd.52.577
220. Kurt H, Erdelt KJ, Cilingir A, et al. Two-body wear of occlusal splint materials. *J Oral Rehabil*. 2012;39(8):584-590. doi:10.1111/j.1365-2842.2012.02301.x
221. Wesemann C, Spies BC, Sterzenbach G, et al. Polymers for conventional, subtractive, and additive manufacturing of occlusal devices differ in hardness and

flexural properties but not in wear resistance. *Dent Mater*. 2021;37(3):432-442. doi:10.1016/j.dental.2020.11.020

222. Wang S, Li Z, Ye H, Zhao W, Liu Y, Zhou Y. Preliminary clinical evaluation of traditional and a new digital PEEK occlusal splints for the management of sleep bruxism. *J Oral Rehabil*. 2020;47(12):1530-1537. doi:10.1111/joor.13083
223. Kern M, Strub JR, Lü XY. Wear of composite resin veneering materials in a dual-axis chewing simulator. *J Oral Rehabil*. 1999;26(5):372-378. doi:10.1046/j.1365-2842.1999.00416.x
224. Issar-Grill N, Roberts HW, Wright EF, Dixon SA, Vandewalle KS. Volumetric wear of various orthotic appliance materials. *Cranio*. 2013;31(4):270-275. doi:10.1179/crn.2013.31.4.004
225. Grymak A, Aarts JM, Ma S, Waddell JN, Choi JJE. Comparison of hardness and polishability of various occlusal splint materials. *J Mech Behav Biomed Mater*. 2021;115:104270. doi:10.1016/j.jmbbm.2020.104270
226. Benli M, Eker Gümüş B, Kahraman Y, et al. Thermal, structural and morphological characterization of dental polymers for clinical applications. *J Prosthodont Res*. 2021;65(2):176-185. doi:10.2186/jpr.JPOR_2019_534
227. Taneva I, Uzunov T. Influence of post-polymerization processing on the mechanical characteristics of 3D-printed occlusal splints. *J Phys Conf Ser*. 2020;1492(1):012018. doi:10.1088/1742-6596/1492/1/012018
228. Wada J, Wada K, Gibreel M, et al. Effect of 3D Printer Type and Use of Protection Gas during Post-Curing on Some Physical Properties of Soft Occlusal Splint Material. *Polymers (Basel)*. 2022;14(21). doi:10.3390/polym14214618
229. Wada J, Wada K, Gibreel M, et al. Effect of Nitrogen Gas Post-Curing and Printer Type on the Mechanical Properties of 3D-Printed Hard Occlusal Splint Material. *Polymers (Basel)*. 2022;14(19). doi:10.3390/polym14193971
230. de Almeida Salles C, de Moraes Melo Neto CL, de Carvalho Dekon SF, et al. Influence of thermocycling and disinfection on the color stability and hardness of thermoplastic sheets used for occlusal splint fabrication. *Gen Dent*. 2021;69(3):42-45.
231. Perea-Lowery L, Gibreel M, Vallittu PK, Lassila L. Evaluation of the mechanical properties and degree of conversion of 3D printed splint material. *J Mech Behav Biomed Mater*. 2021;115:104254. doi:10.1016/j.jmbbm.2020.104254

232. Rosentritt M, Huber C, Strasser T, Schmid A. Investigating the mechanical and optical properties of novel Urethandimethacrylate (UDMA) and Urethanmethacrylate (UMA) based rapid prototyping materials. *Dental Materials*. 2021;37(10):1584-1591. doi:10.1016/j.dental.2021.08.009
233. Reymus M, Stawarczyk B. In vitro study on the influence of postpolymerization and aging on the Martens parameters of 3D-printed occlusal devices. *J Prosthet Dent*. 2021;125(5):817-823. doi:10.1016/j.prosdent.2019.12.026
234. Gautam R, Singh RD, Sharma VP, Siddhartha R, Chand P, Kumar R. Biocompatibility of polymethylmethacrylate resins used in dentistry. *J Biomed Mater Res B Appl Biomater*. 2012;100(5):1444-1450. doi:10.1002/jbm.b.32673
235. Väyrynen VOE, Tanner J, Vallittu PK. The anisotropicity of the flexural properties of an occlusal device material processed by stereolithography. *J Prosthet Dent*. 2016;116(5):811-817. doi:10.1016/j.prosdent.2016.03.018
236. Lutz F, Krejci I, Barbakow F. Chewing pressure vs. wear of composites and opposing enamel cusps. *J Dent Res*. 1992;71(8):1525-1529. doi:10.1177/00220345920710081201
237. Ayazi S, Lipham JC, Hagen JA, et al. A new technique for measurement of pharyngeal pH: normal values and discriminating pH threshold. *J Gastrointest Surg*. 2009;13(8):1422-1429. doi:10.1007/s11605-009-0915-6
238. Alnasser M, Finkelman M, Papathanasiou A, Suzuki M, Ghaffari R, Ali A. Effect of acidic pH on surface roughness of esthetic dental materials. *J Prosthet Dent*. 2019;122(6):567.e1-567.e8. doi:10.1016/j.prosdent.2019.08.022
239. Tinastepe N, Malkondu O, Kazazoglu E. Hardness and surface roughness of differently processed denture base acrylic resins after immersion in simulated gastric acid. *J Prosthet Dent*. 2023;129(2):364.e1-364.e9. doi:10.1016/j.prosdent.2022.12.001
240. Turker I, Kursoglu P. Wear evaluation of CAD-CAM dental ceramic materials by chewing simulation. *J Adv Prosthodont*. 2021;13(5):281-291. doi:10.4047/jap.2021.13.5.281
241. Chaput JP, Dutil C, Featherstone R, et al. Sleep duration and health in adults: an overview of systematic reviews. *Appl Physiol Nutr Metab*. 2020;45(10 (Suppl. 2)):S218-S231. doi:10.1139/apnm-2020-0034

242. Cosme DC, Baldisserotto SM, Canabarro S de A, Shinkai RS. Bruxism and voluntary maximal bite force in young dentate adults. *Int J Prosthodont*. 2005;18(4):328-332.
243. Kern M, Strub JR, Lü XY. Wear of composite resin veneering materials in a dual-axis chewing simulator. *J Oral Rehabil*. 1999;26(5):372-378. doi:10.1046/j.1365-2842.1999.00416.x
244. Ramp MH, Suzuki S, Cox CF, Lacefield WR, Koth DL. Evaluation of wear: Enamel opposing three ceramic materials and a gold alloy. *J Prosthet Dent*. 1997;77(5):523-530. doi:10.1016/S0022-3913(97)70146-X
245. Straffelini G. *Friction and Wear. Methodologies for Design and Control.* . Switzerland: Springer International Publishing AG Switzerland; 2015.
246. Leinfelder KF, Taylor DF, Barkmeier WW, Goldberg AJ. Quantitative wear measurement of posterior composite resins. *Dent Mater*. 1986;2(5):198-201. doi:10.1016/S0109-5641(86)80013-6
247. Perry R, Kugel G, Kunzelmann KH, Flessa HP, Estafan D. Composite restoration wear analysis: conventional methods vs. three-dimensional laser digitizer. *J Am Dent Assoc*. 2000;131(10):1472-1477. doi:10.14219/jada.archive.2000.0060
248. Okeson JP. The effects of hard and soft occlusal splints on nocturnal bruxism. *J Am Dent Assoc*. 1987;114(6):788-791. doi:10.14219/jada.archive.1987.0165
249. Schulte JK, Anderson GC, Sakaguchi RL, DeLong R. Wear resistance of isosit and polymethyl methacrylate occlusal splint material. *Dent Mater*. 1987;3(2):82-84. doi:10.1016/s0109-5641(87)80010-6
250. Zhang F, Spies BC, Vleugels J, et al. High-translucent yttria-stabilized zirconia ceramics are wear-resistant and antagonist-friendly. *Dent Mater*. 2019;35(12):1776-1790. doi:10.1016/j.dental.2019.10.009
251. Wimmer T, Huffmann AMS, Eichberger M, Schmidlin PR, Stawarczyk B. Two-body wear rate of PEEK, CAD/CAM resin composite and PMMA: Effect of specimen geometries, antagonist materials and test set-up configuration. *Dent Mater*. 2016;32(6):e127-36. doi:10.1016/j.dental.2016.03.005
252. Bates JF, Stafford GD, Harrison A. Masticatory function--a review of the literature. 1. The form of the masticatory cycle. *J Oral Rehabil*. 1975;2(3):281-301. doi:10.1111/j.1365-2842.1975.tb00921.x

253. Preis V, Behr M, Handel G, Schneider-Feyrer S, Hahnel S, Rosentritt M. Wear performance of dental ceramics after grinding and polishing treatments. *J Mech Behav Biomed Mater*. 2012;10:13-22. doi:10.1016/j.jmbbm.2012.03.002
254. Matzinger M, Hahnel S, Preis V, Rosentritt M. Polishing effects and wear performance of chairside CAD/CAM materials. *Clin Oral Investig*. 2019;23(2):725-737. doi:10.1007/s00784-018-2473-3
255. Papathanasiou I, Papavasiliou G, Kamposiora P, Zoidis P. Effect of Staining Solutions on Color Stability, Gloss and Surface Roughness of Removable Partial Dental Prosthetic Polymers. *J Prosthodont*. 2022;31(1):65-71. doi:10.1111/jopr.13360
256. Mickeviciute E, Ivanauskiene E, Noreikiene V. In vitro color and roughness stability of different temporary restorative materials. *Stomatologija*. 2016;18(2):66-72.
257. Nezir M, Ahisha CD, Özcan S, Üçtaşli MB. The effect of detox solution on color stability, roughness, and microhardness of monochromatic universal composite resins. *BMC Oral Health*. 2024;24(1):789. doi:10.1186/s12903-024-04587-x
258. Zakir T, Dandekeri S, Suhaim K, Shetty NG, Ragher M, Shetty S. Influence of aerated drink, mouthwash, and simulated gastric acid on the surface roughness of dental ceramics: A comparative In Vitro study. *J Pharm Bioallied Sci*. 2020;12(5):480. doi:10.4103/jpbs.JPBS_143_20
259. Sagsoz O, Polat Sagsoz N. Chemical degradation of dental CAD/CAM materials. *Biomed Mater Eng*. 2019;30(4):419-426. doi:10.3233/BME-191063
260. Ranjitkar S, Kaidonis JA, Townsend GC, Vu AM, Richards LC. An in vitro assessment of the effect of load and pH on wear between opposing enamel and dentine surfaces. *Arch Oral Biol*. 2008;53(11):1011-1016. doi:10.1016/j.archoralbio.2008.05.013