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UNIVERSITY OF TURKU

FIBER-REINFORCED IMPLANT-RETAINED OVERDENTURES

Studies of their mechanical behavior and deformation in relation with attachment systems

Mona Fathy Elsaed Mohammed Gibreel

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University of Turku

Faculty of Medicine
Institute of Dentistry
Department of Biomaterials Science
Finnish Doctoral Programme in Oral Sciences – FINDOS Turku

Supervised by

Professor, Pekka K. Vallittu
Department of Biomaterials science
Institute of Dentistry
University of Turku
Turku, Finland

Docent, Leila Perea-Lowery
Department of Biomaterials science
Institute of Dentistry
University of Turku
Turku, Finland

Reviewed by

Professor, Gustavo Mendonca
Biologic and Materials Sciences
University of Michigan
School of Dentistry
Michigan, USA

Professor, Jason A. Griggs
Biomedical Materials Science
School of Dentistry
University of Mississippi Medical Center
Mississippi, USA

Opponent

Professor, Per Vult von Steyern
Department of Materials Science and Technology
Faculty of Odontology
Malmö University
Malmö, Sweden

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To all colleagues, students and whom it might help

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ABSTRACT

The fracture incidence is commonly noticed at the attachment area of implant-retained overdentures (IODs) due stress concentration. This series of studies aimed to investigate the effect of polymer-preimpregnated fiber reinforcement on mechanical properties and denture base strains of IODs.

In the four experimental studies, acrylic resin specimens and experimental models simulating IODs were fabricated. They were reinforced with glass fibers and connected to the implants with stud attachments. Study I evaluated the influence of the quantity and position of bidirectional woven glass fiber reinforcing layers on the load-bearing capacity of simulated locator-retained overdentures. Study II evaluated the effect of bidirectional glass fiber reinforcement's positioning on the fatigue resistance of simulated single locator-retained overdentures. Study III compared the flexural strength and modulus between soft liner-retained and ball-and-socket–retained overdentures, as well as the effect of using unidirectional and bidirectional glass fiber reinforcements on the mechanical properties of soft liner-retained overdentures. Finally, study IV evaluated the effect of unidirectional glass fiber reinforcement on the mid-line denture base strains of overdentures retained with a single implant. Results were statistically analyzed by analysis of variance (ANOVA) and post hoc Tukey's test using statistical software. The results showed a significant increase in the flexural strength and fatigue resistance of the overdenture specimens reinforced with 4 layers of bidirectional E-glass fiber weaves or one bundle of unidirectional E-glass fibers placed above the attachment housing. Also, the latter type of reinforcement significantly reduced the midline strains of the single implant-supported overdenture base by almost 50%.

It can be concluded that a proper amount of polymer-preimpregnated glass fiber reinforcement (4 layers of bidirectional fiber weaves or one bundle of unidirectional fibers) placed above the attachment can significantly improve the mechanical properties of IODs and reduce the denture base strains.

KEYWORDS: implant, overdenture, attachment, fiber, flexural strength, silicone, liner, strain

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TIIVISTELMÄ

Implanttikantoisen peittoproteesin pohjalevyn murtumat ovat yleisiä erikoiskiinnikkeen alueella, johon muodostuu purentakuormituksen aikana jännityskeskittymiä. Tämän tutkimuksen tarkoituksena oli selvittää muovilla esikyllästettyjen kuitulujitteiden vaikutusta peittoproteesin pohjalevyn ominaisuuksiin erityisesti erikoiskiinnittimen matriisin kohdalla.

Kokeellisessa neljän osatyön kokonaisuudessa akryylista valmistetuilla koekappaleilla jäljiteltiin peittoproteesin rakennetta. Koekappaleet lujitettiin lasikuiduilla. Ensimmäisessä osatyössä käytettiin lujitteena lasikuitukangaskudosta, joka sijoitettiin eri tasoille koekappaleeseen erikoiskiinnikkeen matriisiosaan nähden. Kappaletta kuormitettiin staattisesti. Toisessa osatyössä kuormitus oli dynaaminen eli kuitulujitteiden vaikutus väsymiskestävyyteen oli tutkimuksen kohteena. Kolmannessa työssä jäljitettiin proteesityyppiä, jossa pohja on pehmeää elastomeeria, joka toimii samalla erikoiskiinnikkeen matriisina. Myös tämä simulaatiotilanne pyrki selvittämään laskikuitulujitteen vaikutusta murtuman etenemisen estämisessä dynaamisessa väsytystilanteessa. Neljännessä osatyössä koekappaleen muoto vastasi alaleuan kokoproteesin muotoa ja proteesi oli tuettu yhdellä implantilla ja erikoiskiinnikkeellä. Proteesin pohjalevyn taipumajännitykset mitattiin venymäliuskoilla. Osatöiden tulokset analysoitiin tilastollisesti varianssianalyysilla. Tulokset osoittivat tilastollisesti merkitsevää taivutuslujuuden ja väsymislujuuden lisääntymistä mikäli koekappale oli lujitettu neljällä kerroksella e-lasikuitukangaskudosta. Vastaava lujitemalli vähensi myös proteesin keskilinjassa olevaa vetojännitystä simuloidun purentakuormituksen aikana.

Yhteenvetona voidaan todeta, että käyttämällä riittävää määrää akryylimuovilla esikyllästettyä lasikuitukangaskudosta implanttikantoisen peittoproteesin lujitteena saadaan materiaalin ja proteesin lujuutta lisättyä merkittävästi.

AVAINSANAT: Implantti, peittoproteesi, erikoiskiinnikke, kuito, taivutuslujuutta, Silikoni, rasitusta

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Abbreviations

ANOVA	Analysis of variance
CAD/CAM	Computer-aided design/computer adid manufacturing
CI	Confidence interval
CFL	Compressive cyclic fatigue limits
E-glass fiber	Electric glass fiber
FRC	Fiber-reinforced composite
FS	Flexural strength
FM	Flexural modulus
g	Gram
GPa	Gigapascal
kPa	Kilopascal
IOD	Implant-retained overdenture
IODs	Implant-retained overdentures
L	Liter
min	Minute
mm	Millimeter
mL	Milliliter
MPa	Megapascal
N	Newton
PFFS	Post fatigue flexural strength
PFR	Partial fiber reinforcement
PMMA	Poly(methyl methacrylate)
SD	Standard deviation
SE	Standard error
S-glass fiber	High strength glass fiber
SEM	Scanning electron microscope
SIO	Single implant-supported overdenture
SN	Stick Net
TFR	Total fiber reinforcement
3D	3 dimensional

UHMWPE	Ultra-high molecular weight polyethylene
Wt.	Weight

List of Original Publications

This dissertation is based on the following original publications, which are referred to in the text by their Roman numerals:

I **Gibreel M**, Lassila LVJ, Närhi TO, Perea-Lowery L, Vallittu PK. Load-bearing capacity of simulated Locator-retained overdenture system. J Prosthet Dent 2018; 120:558-564.

II **Gibreel M**, Lassila LVJ, Närhi TO, Perea-Lowery L, Vallittu PK. Fatigue resistance of simulated locator –retained overdenture system: An in vitro study. J Prosthet Dent 2019;122: 557-563. doi: 10.1016/j.prosdent.2018.11.013.

III **Gibreel M**, Lassila LVJ, Närhi TO, Perea-Lowery L, Vallittu PK. Flexural strength and flexural modulus of fiber-reinforced, soft-liner retained implant overdenture. Int J Prosthodont 2021. doi: 10.11607/ijp.6677. Online ahead of print

IV **Gibreel M**, Lassila LVJ, Närhi TO, Perea-Lowery L, Vallittu PK. Midline denture base strains of glass fiber-reinforced single implant-supported overdentures. J Prosthet Dent. 2020: S0022-3913(20)30389-9. doi: 10.1016/j.prosdent.2020.05.018. Online ahead of print.

1 Introduction

Rehabilitation of the edentulous mandible with implant-retained overdenture (IOD) is a well-accepted treatment with long-term effective outcomes. It can successfully overcome the retention and stability problems associated with traditional complete dentures (Burns *et al.*, 1995, 2011). A variety of attachment systems can be used with the IOD which can be either splinted or solitary (Laverty et al., 2017). However, this treatment modality has been associated with some biological and mechanical complications. The most common mechanical complications are loss of attachment system retention, wear of retention elements, frequent overdenture relining or repair, and implant fracture (Vahidi and Pinto-Sinai, 2015). Poly (methyl methacrylate) (PMMA) denture bases are susceptible to crack propagation due to their hard and brittle nature. Denture base fractures are often caused by fatigue mechanisms when subjected to repeated masticatory loads (McCabe and Walls, 2008; Narva et al., 2005a). Therefore, flexural fatigue resistance is considered to be a significant mechanical property that affects the prosthesis's clinical durability (Kelly, 1969).
High masticatory load (Boven et al., 2015; Chee and Jivraj, 2007; Sharma et al., 2017), rigid bone-implant interface (Yoo et al., 2017), and thin denture base area adjacent to the abutment (Gonda et al., 2007) could be the main causes behind the high risk of fracture. Furthermore, stress distribution within the denture base differs when an implant-supported prosthesis is used (Fontijn-Tekamp et al., 2000). Stresses become more concentrated around the attachment components and the abutment which acts as a fulcrum (Gonda et al., 2007), increasing the risk of denture base fracture in this area (Takahashi et al., 2015).

Many studies have recommended the use of glass fibers in reinforcing denture base polymers (Narva et al., 2005b; Vallittu, 1999) due to their good aesthetic and chemical bonding to the resin matrix with the aid of silane coupling agent (Agha et al., 2016; Vallittu, 1999; Vallittu and Narva, 1997). The concept of partial fiber reinforcement (PFR) was developed to overcome the technical sensitivity associated with total fiber reinforcement (TFR). In PFR, a small quantity of reinforcing fibers is inserted into the weak areas of the denture base to enhance the toughness of polymers in thin areas, such as the area around the overdentures' attachments (Vallittu, 1997a).

Hence, accurate positioning of a proper amount of fiber reinforcement within weak areas of the prosthesis is a key factor in managing the overdentures' fracture complication. Therefore, this series of studies attempts to investigate the effect of the quantity and position of fiber reinforcement within the overdenture base polymer construction on its mechanical properties.

2 Review of the Literature

2.1 Implant overdentures

2.1.1 Introduction and definition

Complete edentulism can be defined as the total loss of natural teeth. It affects negatively the social, psychological, and oral general status of the patients due to the disturbance in oral functions, decreased chewing ability, and compromised facial aesthetic (Almusallam and AlRafee, 2020; The Glossary of Prosthodontic Terms, 2017). Conventional dentures are the primary treatment choice for rehabilitating completely edentulous patients. However, patients rehabilitated with complete dentures usually complain of masticatory difficulties, discomfort, and poor adaptation to wearing the denture (Allen and McMillan, 2003; Bellini et al., 2009).

An overdenture can be defined as a removable dental prosthesis that covers and is partially supported by natural teeth, natural tooth roots, and/or dental implants. (The Glossary of Prosthodontic Terms, 2017).

2.1.2 Implant overdenture advantages:

For an edentulous arch, the possible implant-supported prosthesis options can be either fixed, fixed detachable, or removable overdenture prosthesis (Sadowsky, 1997). Although the advantages of fixed implant restorations are undoubted, they might not be a preferable option for many patients (Jivraj et al., 2006).

Differential treatment planning is established based on initial preoperative evaluation of the structure and quality of the edentulous residual ridge, the intermaxillary relationship, aesthetics, phonetics, hygiene, and cost considerations (Zitzmann and Marinello, 2000). Although implant-supported fixed prostheses have better aesthetics, they are costly and not applicable in some clinical situations, such as severe alveolar bone resorption (Sadowsky, 1997). An implant-supported removable prosthesis offers certain advantages over implant-supported fixed restoration in terms of being cost-effective and time-saving (Attard et al., 2005; Emami et al., 2014; Zitzmann et al., 2006).

Implant overdenture treatment has a good impact on patient's quality of life and positive outcomes to the patients, especially in terms of denture retention, stability, and residual ridge preservation (Goodacre, 2018; Kelly and McKenna, 2020). The psychosocial, functional, and anatomic benefits that the patients get from implant overdentures have been reported in several studies. Improved appearance (Heydecke et al., 2003), patient satisfaction (Thomason et al., 2009), and quality of life (Awad et al., 2003) are all examples of psychosocial benefits. Implant overdentures can enhance the patient's appearance when compared with conventional dentures since they allow the teeth to be placed in optimal aesthetic positions without interfering with muscle movement and denture stability (Goodacre, 2018). Furthermore, patients with implant overdenture have a higher biting force than those wearing conventional complete dentures, which in turn improves their food chewing ability (Benzing et al., 1994; Geckili et al., 2012). Consequently, they can eat a wider variety of food options which affects positively their nutrition (Morais et al., 2003).

Some studies showed that the rate of alveolar bone resorption for complete denture wearers is higher than that for patients with implant overdentures (Alsrouji et al., 2019; Wright et al., 2002). In addition, less muscle atrophy around the mouth is encountered as the implant overdenture tends to show better stability that allows for greater muscle activity without denture displacement (Müller et al., 2012).

Because of the bulk of the restoration, many new denture wearers complain of discomfort. Other complaints associated with maxillary dentures include impaired taste and gagging reflex, particularly in patients with a low gagging threshold (Dervis, 2002). An implant-supported overdenture allows prosthesis's flanges (Ceruti et al., 2006) and palatal coverage reduction (Jain et al., 2013; Sadowsky and Zitzmann, 2016) which assists in the management of the aforementioned issues.

2.1.3 Implant overdenture indications:

An implant overdenture removable prosthesis might be considered as a more advantageous treatment option than a fixed prosthesis. This is especially true in the case of patients with an extremely resorbed residual ridge where the support of the lips and soft tissues of the face is lacking. It is also the same for those who are unable to maintain good oral hygiene around the implants/prosthesis. Other indications would be when the number, positioning, or angulation of the implant fixtures are insufficient for a fixed reconstruction, when several surgeries, such as bone grafting, are contraindicated, and when the patient has financial limitations and a limited time frame (Laverty et al., 2017).

In addition, an implant overdenture removable prosthesis is indicated for patients with a superficially located mental nerve, sensitive mucosa, knife-edged ridge, and/or sharp mylohyoid projections (Emami et al., 2014). It is also recommended for patients who have been exposed to trauma or surgery for head and neck cancer that modified the denture bearing anatomy (Schoen et al., 2004), patients with neuromuscular disorders who can't tolerate/control conventional removable prostheses (Laverty et al., 2017; Romero-Pérez et al., 2014), and when a fixed implant reconstruction is contraindicated (Laverty et al., 2017).

2.2 Overdenture attachments

An attachment is a mechanical device (retainer) used for the fixation, retention, and stabilization of an implant overdenture, consisting of a metal receptacle and a closely fitting part. The former (the female or matrix component) and the latter (the male or patrix component) (The Glossary of Prosthodontic Terms, 2017). Besides improving retention, stability, and support, which in turn extends the longevity of the prosthesis and implants, they enable patients to easily place and remove the prosthesis (Trakas et al., 2006).

A variety of attachment systems have been developed for connecting overdentures to the underlying implants. They are divided into 2 main categories: splinted (bar attachment) and stud (solitary) attachments such as (telescopic, locators, ball, and magnets). According to the allowed movement, they could either be resilient or rigid (Alsabeeha et al., 2009; Sadowsky, 2001; Trakas et al., 2006). Attachments can be either rigid if they do not allow any denture movements, or resilient if they allow translation, rotation, axial, or hinge movements or a combination of them. With rigid attachments, almost 100% of the occlusal load will be delivered to the implants, while with resilient attachments, the occlusal load will be distributed between the implants and the other supporting structures (Martínez-Lage-Azorín et al., 2013).

Many factors must be considered while planning the treatment and selecting the attachment system for an IOD. The number of implants, the cost-effectiveness, the amount of retention needed, the arch geometry, inter-arch distance, bone quantity, bone quality, expected level of oral hygiene, patient's social status, patient's demands, the maxilla-mandibular relationship, the condition of the opposing arch, the inter-implant distance, stress distribution, and maintenance requirements (Bergendal and Engquist, 1998; Laverty et al., 2017; Trakas et al., 2006; Wismeijer et al., 1999) all play a role in the final decision.

2.2.1 Bar attachment

The bar and clip can be either rigid or resilient depending on the shape of the transverse section and the clip material composition (dos Santos et al., 2014). Bar systems can be used alone as direct retainers such as the Hader or Dolder bar systems or in combination with secondary attachments such as locators (Laverty et al., 2017). A cantilever bar extension on the distal of the last implant increases the retention area of the prosthesis. However, the loss of retention over time and a higher biofilm accumulation have been reported with the bar attachment (Burns et al., 2011; Gonçalves et al., 2020).

2.2.2 Stud attachment:

Stud attachments are characterized by having a low profile with reduced leverage action on the abutment and easier hygienic maintenance (Walton et al., 2001). Moreover, they are less technique sensitive and require less space within the prosthesis when compared to splinted designs (MacEntee et al., 2005; Walton et al., 2001). They are suitable for patients with tapered V-shaped arches where a bar attachment is contraindicated (Walton et al., 2001).

2.2.2.1 Ball and socket attachment

Ball attachments consist of a spherical patrix which is usually screwed to the implant fixture and a matrix that fits over the patrix providing retention utilizing spring-action arms or an interchangeable elastic ring (Taddei et al., 2004). They require less space in buccolingual and occlusal directions when compared to bar attachments. Moreover, they are commonly used because of their low cost, variability, ease of handling, and minimal chair-side time requirements (Laverty et al., 2017; Ortensi et al., 2019).

Ball attachments offer variable degrees of free movements and resiliency according to their design (Daas et al., 2008). The main advantages that the ball attachments provide are due to their versatility and applicability in different clinical situations without the need for fabricating new dentures (Shor et al., 2007). The ball/O-ring attachment could transfer less amount of stress to the implants and produce less bending moment than the bar/clip attachment (Tokuhisa et al., 2003). Implant overdentures retained by ball attachment systems recorded high implant survival and prosthetic success rates with minimal complications, high patient satisfaction, and successful biological parameters in the mid-term follow-up (Krennmair et al., 2012; Ortensi et al., 2019).

2.2.2.2 Locator attachment:

The locator attachment is known for its low profile, dual retention, and divergence up to 40 degrees. Moreover, it is characterized by its self-aligning ability which aids patients to position their prostheses and reduces the incidence of wear (Elsyad et al., 2017a; Evtimovska et al., 2009). Therefore, it is a good option for patients with limited inter-arch space and elderly patients with poor manual dexterity (Bilhan et al., 2011).

A locator attachment has nylon inserts inside the metal housing with different colours according to the degree of retention. The nylon replacements are available in 2 types according to the retention they provide: the first type provides internal and external retention and comes in 3 colours (transparent for high retention, pink for medium retention, and blue for low retention). While the second type is intended for external retention only with green, orange, and red colours ranging from high to low retention respectively (Martínez-Lage-Azorín et al., 2013).

2.3 Soft liners

Soft denture liners are materials used to make a cushion layer between the hard denture base and the oral mucosa. They can absorb some of the masticatory loads and distribute them more evenly to the underlying supporting tissues (Braden et al., 1995; Hashem, 2015; McCabe et al., 2002). They can be used with atrophied ridge, thin and non-resilient mucosa, bony undercuts, immediate prosthesis, healing after implant placement, maxillo-facial prostheses, and for patients with bruxism and xerostomia. Moreover, soft liners can be used to minimize the incidence of age-related problems occurring in the denture bearing area (Gjengedal et al., 2013; Hashem, 2015; Lowe, 2004).

Resilient liners can be divided into silicone rubbers or plasticized acrylics and can be either auto-, heat-, or visible light-polymerized (Ergun and Nagas, 2007). The plasticized acrylic resin consists of a powder containing acrylic polymers and copolymers and a liquid containing an acrylic monomer. Plasticizers' function is to keep the material soft. (Gronet et al., 1997). The silicone elastomers are mainly based on dimethylsiloxane polymers, which are similar in chemical composition to silicone impression materials. They can keep their elasticity for a long time since they don't depend on leachable plasticizers (McCabe, 1998). An ideal soft liner material should cushion the mucosa, be dimensionally stable, permanently resilient, with minimal fluid sorption and solubility, and have fungal growth inhibitory properties (Chladek et al., 2014; Hashem, 2015; McCabe et al., 2002). The advantageous mechanical properties are ease of processing, ease of finishing and polishing, and adhesion to the denture resin (Pesun et al., 2001). The use of a soft liner can improve both masticatory efficiency and oral comfort (Palla et al., 2015).

Silicone resilient liners have been used as an alternative to the metal matrix of the ball attachment with the benefit of reducing significantly stresses on the peri-implant tissues (Kanazawa et al., 2007). Moreover, they can be used as an alternative to the metal matrix when the number, angulation, and/or location of implants deviate from the original treatment plan. Their use might be recommended when implants are placed in a grafted bone or bone of poor quality. It is a more simple and cheaper design that needs less prosthetic maintenance (Cain and Mitchell, 1998; Elsyad et al., 2017b). Placing a soft lining material around the implant abutment compensates for the volumetric contraction of the denture base resin, which happens during processing. Therefore, it prevents direct contact between the abutments and the acrylic resin and reduces the possibility of implant overloading (Sakaguchi and Powers, 2012). Because of their viscoelastic properties, soft liners can divide masticatory loads between the implant and the residual ridge (Adrian et al., 1992), reduce the need for prosthetic maintenance (Elsyad et al., 2017b; Elsyad, 2012), and decrease the incidence of peri-implant soft tissue complication (Elsyad et al., 2017b; Gibreel et al., 2017). Their bonding strength values to the denture base acrylic resin are acceptable and can be improved by the use of primer (Lassila et al., 2010; Więckiewicz et al., 2014). Long-term silicone soft liners can last for up to one year (Hashem, 2015).

2.4 The number of implants and their location:

The mandible's compact bone structure and dense cortical plates make it suitable for receiving dental implants. The number of implants required to improve function without subjecting the patient to excessive surgical procedures should be considered during treatment planning for a mandibular implant overdenture. (Mittal et al., 2016; Vahidi and Pinto-Sinai, 2015).

Based on several randomized controlled studies, a review of the McGill and York consensus statement concluded that 2-implant supported overdenture is the minimum standard for patient satisfaction regarding the improvement in functions (Feine et al., 2002). A study conducted by Merickse-Stern (1990) concluded that increasing the number of implants improved overdenture’s retention and stability slightly. Another in vivo study (Bilhan et al., 2012), which compared between 2-, 3-, and 4-implant-supported overdenture found that the number of implants had a non-significant effect on the maximum biting force. However, in some cases, the placement of more than 2 implants might be necessary, such as remaining maxillary natural teeth, that would increase the masticatory forces on the mandible; implants which are less than 8 mm in length or less than 3 mm in width; soft tissues which are sensitive to occlusal loading; high muscle attachments or sharp mylohyoid projections; large V-shaped ridges; and patients’ demand for high retention (Sadowsky, 2001).

2.4.1 Single implant-supported overdenture (SIO):

In 2009, the British Society for the Study of Prosthetic Dentistry conference in York released another statement supporting the McGill consensus statement (Thomason et al., 2009). However, there is still no agreement on the exact number of implants that would provide the best treatment prognosis (Di Francesco et al., 2019).

A mandibular overdenture retained by a single implant placed in the midline area of the edentulous mandible, which was suggested by Cordioli et al. (1997) has been considered as an alternative option with a reduced cost for the 2-implant-supported overdenture. Previous studies (Cordioli et al., 1997; de Souza Batista et al., 2018; Krennmair and Ulm, 2001) reported its favourable outcomes, especially when compared to complete dentures. Moreover, it can be considered as a simple, less invasive, and straightforward prosthetic technique (Liu et al., 2013).

SIO was found to exhibit low strain values on peri-implant bone and adequate denture stability when compared with mandibular overdentures retained by 2, 3, and 4 implants (Liu et al., 2013). A 3-year clinical evaluation study (Harder et al., 2011) concluded that such a treatment was a successful option for the edentulous mandible in terms of oral health-related quality of life and chewing ability. A systematic review study (Padmanabhan et al., 2020) reported high implant survival rates and minimal incidence of complications for SIO.

2.5 Overdenture fracture complications:

Regardless of the implants' number, using implant overdentures leads to an increase in the patients' masticatory efficiency (Bakke et al., 2002). Unfortunately, this increase can cause extra mechanical complications making the prosthesis more vulnerable to failure as the yield stress of the material is exceeded. Implant prosthesis's complications can be classified into 6 categories: surgical complications, implant loss, bone loss, peri-implant soft tissue complications, aesthetic/phonetic complications, and mechanical complications such as overdenture fractures (Goodacre et al., 2003).

Fracture of an implant overdenture can result from impact force or flexural fatigue after repeated loading (Goodacre et al., 2003; Jagger et al., 1999). Impact force failures are caused by a sudden blow to the denture, such as accidental dropping while flexural fatigue occurs after repeated loading of a material (Uzun and Hersek, 2002). Most of the overdenture fractures were noticed in the thinner denture base area surrounding the abutments as a large space is occupied by the attachment matrix (Gonda et al., 2010; Rodrigues, 2000). That area is occupied by a weaker autopolymerizing resin, which is used for attachment pick-up procedures in most of the denture bases (Tokgoz et al., 2019).

The incidence of overdenture fractures has been reported in several studies (Chhabra et al., 2019; Nedir et al., 2006). A 34% overdenture base fracture located above the abutment over a 5-years observation period was reported (Chhabra et al., 2019). When Nedir et al. (2006) evaluated the prosthetic complications associated with dental implants, they reported the incidence of fracture failure in 10.9% of the prostheses after 5 to 6 years. A literature review (Goodacre et al., 2003) identified overdentures' fracture as one of the mechanical complications that affected 12% of prostheses. The difference in fracture incidence rate between SIO and those retained by 2 implants was statistically non-significant, while fracture was noticed in the matrix areas adjacent to implants (Gonda et al., 2010).

Different factors tend to play a role in the incidence of overdenture base fracture, such as thickness (Tokgoz et al., 2019), housing retaining material (Ozkir and Yilmaz, 2017), and the coping height (Dong et al., 2006). A higher abutment tends to increase the strains near the top of it, while a short abutment can direct more strains toward the middle part of the denture (Dong et al., 2006). Another factor could be the attachment type (Elsyad et al., 2016, 2020). A study comparing clinical denture base deformation between overdentures retained with either balls or locators concluded that ball attachments caused a significant deformation in the mandibular denture base and high tensile strains in the area above the implants (Elsyad et al., 2016). That was attributed to the lack of resiliency and intimate contact between the ball and socket generating a fulcrum of overdenture rotation and concentrating strains in these areas. These strains may cause crack initiation that can result in denture base fracture (Elsyad et al., 2013, 2016). Another study (Elsyad et al., 2020) concluded that telescopic and stud attachments were associated with less denture base deformation and strains when compared to the bar, which occupied a greater space inside the denture base.

Since PMMA, which is the most commonly used polymer for denture fabrication tends to be brittle, meaning that it is stronger under compression rather than under tension (Sakaguchi and Powers, 2012). Therefore, some studies (Elsyad et al., 2016, 2020; Sadowsky, 2001) recommended the reinforcement of overdenture base in areas that are subjected to tensile stresses to increase their fracture resistance.

2.6 Overdenture reinforcement:

Generally, 3 methods have been proposed for denture base reinforcement: chemical modification of the acrylic, filler particles addition, addition of fibers to the acrylic, and the incorporation of a metal substructure (cast or wrought) in the acrylic resin (Jagger et al., 1999).

2.6.1 Chemical modification of the acrylic (high impact resin)

In high-impact strength acrylic resins, the PMMA polymer is modified by adding a rubber compound to enhance their mechanical properties (Meng & Latta, 2005). Alternatively, acrylic-elastomer copolymers such as methyl methacrylate–butadiene–styrene are added to the powder. In these denture base materials, the rubber is grafted with a methacrylate group allowing the particles to form a covalent bond with the polymer network (Abdulwahhab, 2013). These rubber compounds can absorb the crack energy and arrest or slow its propagation through the acrylic denture base. They can be incorporated within the PMMA by up to 30% without changing the handling properties such as the viscosity (Gad et al., 2017). Some studies (Narendra et al., 2013; Uzun and Hersek, 2002) showed that high impact resins provided higher flexural strength and fracture toughness when compared with the traditional heat-cured and injection-molded resins. However, its use is limited due to the high cost, which is sometimes up to 20 times that of conventional resin (Jagger et al., 1999).

2.6.2 Filler addition

Several studies (Asar et al., 2013; Casemiro et al., 2008; Gad et al., 2017; Sehajpal and Sood, 1989; Zuccari et al., 1997) have tested the use of fillers such as metal oxides, noble metals, minerals, and carbon to enhance the denture base resin strength. A considerable improvement in its characteristics was noticed when using metal oxides, particularly ZrO_2, which improved the material's physical and mechanical characteristics, as well as patients' perception of heat and cold stimuli (Asar et al., 2013; Gad et al., 2017). The nano-sized filler particles were used for the same purpose, since they can provide a high surface area, fine size, and more homogenous distribution. The size, shape, type, and concentration of the added nano-filler particles affect the properties of the reinforced resin (Gad et al., 2016, 2018).

2.6.3 Metal incorporation

Metallic wrought wires and cast frameworks have been used for reinforcing the implant overdentures and complete dentures (Amaral et al., 2018; Balch et al., 2013; Murthy et al., 2015; Rodrigues, 2000). However, the lack of chemical bonding between the metal and resin can cause defects and thereby concentrate stress, leading to a higher rate of failure (Jagger et al., 1999). Therefore, different methods for improving the bond between metal and acrylic resin have been investigated. Vallittu and Lassila (1992b) studied the effect of sandblasting the metal surface on wires. They concluded that roughening of the metal surfaces increases the fracture resistance of metallic wires-reinforced dentures as it improves the mechanical bonding with acrylic resin.

Regarding metal-reinforced overdentures, Gonda et al. (2007) studied the effect of 3 designs of metal on midline strain and strain around the retentive copings of mandibular teeth-supported overdenture. They concluded that a chrome-cobalt framework design extending over the ridge and the retentive copings could provide better strain resistance in comparison with the other designs. The effect of different types of reinforcements on the dynamic and static strength of a simulated implant-supported overdenture was investigated in a previous study (Rached et al., 2011). The evaluated types of reinforcement were: braided stainless steel bar, steel mesh, unidirectional E-glass fibers, E-glass fiber mesh, woven polyethylene braids, and polyaramid fibers. The authors concluded that E-glass fibers, woven polyethylene braids, and polyaramid fibers increased the flexural strength and fatigue resistance of the implant-supported overdenture. Therefore, fibers have been considered more advantageous and successful over metallic reinforcements (Dyer et al., 2005; Vallittu, 1999; Vallittu, 2018). However, a metal framework is still valid as a reinforcement option (Grageda and Rieck, 2014; Ahuja et al., 2012).

2.6.4 Fiber incorporation

Fibers fabricated from different materials such as glass, carbon/graphite, aramid, or ultra-high molecular weight polyethylene (UHMWPE) have been used as a denture base reinforcement. Polyamide fibers, including nylon and aramid, are biocompatible and can increase the flexural strength and modulus of denture base resins (Chen et al., 2001). However, aramid fibers possess undesirable aesthetics especially for the anterior region due to their yellowish colour (Soygun et al., 2013). Also, the black colour of carbon fibers has been considered as a limiting factor against their clinical use (Vallittu, 1996).

Ultra-high molecular weight polyethylene fiber is a natural crystalline polymer that can increase the impact strength and toughness of denture base resins (Gad et al., 2017; Yu et al., 2012). However, due to the weak bond between these fibers and PMMA, surface etching of the fibers using electrical plasma may be needed to achieve mechanical bonding (Uzun et al., 1999). Unfortunately, this etching may negatively affect the mechanical properties (Takagi et al., 1996).

Glass fiber reinforcement is the most commonly used type of reinforcement. They are characterized by their high tensile strength. They have excellent compression and impact properties, high modulus of elasticity, and good bending strength. They stretch uniformly under stress to their breaking point without yielding and when the tensile load is removed before reaching the breaking point, they return to their original length. Therefore, these fibers can store and release a large amount of energy (Murphy, 1994). Their transparency makes them suitable for dental applications. Glass fibers differ according to their composition and the most commonly used glass fibers in the dental application are electrical (E) glass and high strength (S) glass since they are chemically stable and durable in the pH range of 4-11. E-glass fiber consists of alumina-borosilicate with less than 1 wt. % alkali oxides. It has good tensile and compressive strength, but poor impact strength. E-glass fiber can be silanized and adhered to the resin matrix of the fiber-reinforced composite (FRC) (Vallittu and Özcan, 2017). The tensile strength of the fiber should be higher than that of the polymer matrix, while its elongation should be equal to or lower than that of the matrix.

The use of glass fiber has been found to significantly improve the mechanical properties (Farina et al., 2012; Narva et al., 2005a, 2005b) of acrylic resin base and reduce its deformation (Kanie et al., 2005). The addition of continuous unidirectional glass fiber improved the transverse strength (Vallittu et al., 1994a), static strength (Narva et al., 2005b), and fatigue resistance (Narva et al., 2005a) of the denture base resin. They seemed to be durable and clinically successful when used in the repair of acrylic resin removable dentures (Narva et al., 2001; Vallittu, 1997a). Also, woven fibers were able to improve the mechanical properties of denture base polymers such as impact strength (Kanie et al., 2000), flexural strength, and flexural modulus (Kanie et al., 2002; Vallittu, 1999). However, their reinforcing effect was lower than the unidirectional fibers (Vallittu, 1999).

2.7 Mechanical testing.

A proposed list of materials' tests that may predict their relevant clinical factors has been developed with the aim of correlating the in vitro and clinical performance of materials with each other (Sarrett, 2005).

2.7.1 Flexural strength (FS) and flexural modulus (FM)

Compressive, tensile, and shear strengths are evaluated using the 3-point bending test performed on rectangular specimens. Loading generates compressive stresses on the upper side of the specimen, tensile stresses on the lower side of the specimen, and shear stresses at the ends of the specimen. The 3-point bending test evaluates the maximum stress that can be tolerated by the material i.e. the flexural strength of the material, and is known as transverse strength or as modulus of rupture (Sakaguchi and Powers, 2012). Flexural strength is the primary mode of evaluation for any denture base material additions, reinforcements, modifications, and composition changes (Gad et al., 2017). The flexural modulus describes the ability of the material to withstand temporary changes during tension or compression. Resilient (flexible) materials has a low flexural modulus and undergoes reversible elastic deformation. Such a material would deform under pressure (such as masticatory stresses), and return to its original size when stress is removed. While a material with high modulus tends to be rigid and brittle, therefore it will break rather than deform (Anusavice et al., 2012; Sakaguchi and Powers, 2012).

The use of bend testing in the assessment of acrylic denture bases was first made by the National Bureau of Standards in the United States of America. Later, it was followed by further development in the standards and testing equipment (Stafford et al., 1980). The ISO protocol evaluation guidelines have been accepted and amended to meet the needs and requirements for simulating clinical situations (Chander et al., 2019; Kelly et al., 2012). Three-point and 4-point bending tests are the commonly used tests for evaluating the flexural strength of PMMA denture base resin (Chander et al., 2019; Ucar et al., 2012).

2.7.2 Fatigue strength

Usually during function, materials are subjected to slow and repetitive cycling loading, rather than static single loading. Fatigue tests aim to evaluate the materials' resistance to such repetitive cyclic loading. Fatigue can be defined as "progressive fracture under repeated loading", while fatigue strength is "the stress level at which material fails under repeated loading" (Sakaguchi and Powers, 2012). Commonly, fatigue failure results from a flaw progressing into a critical crack or from coalescing of multiple cracks over many cycles that can lead to premature failure (Anusavice et al., 2012).

Fatigue failures can occur even in materials with high strength when subjected to static loading. The endurance limit can be defined as the stress level at which the material can provide maximum service without failure (Anusavice et al., 2012). Defects and surface flaws are critical for the material's fatigue properties since they can initiate cracks that can grow over time and progress into macroscopic cracks, resulting in fracture. Therefore, identifying the fatigue resistance of a material can provide a better understanding of the material behaviour under clinical situations as well as the causes of failure.

2.8 Principles of the effectiveness of fiber reinforcements

Different Factors must be considered to obtain an effective reinforcement with fibers.

2.8.1 Fiber quantity and placement

Two approaches have been identified for denture base reinforcement with fibers. One was described by Ladizesky et al. (1990) which is known as the total fiber reinforcement where the whole denture base is reinforced with fibers. However, some clinical and technical problems were associated with this approach such as fiber protrusion and difficulties in polishing the denture surface. Another approach that was described by Vallittu and Lassila (1992a, 1992b) depends on the precise placement of oriented fibers only within the weakest areas of the denture base. This approach is called partial fiber reinforcement. To use this approach successfully, it is necessary to have prior knowledge of the existing or supposed path of the fracture line. Glass fibers increased the polymer toughness and its fracture resistance when placed in weak and thin areas of the prosthesis. Therefore, fibers should be placed on the tension side during mastication and at the right angle to the possible fracture line (Vallittu, 1997a).

The position of glass fibers within the denture base affects its flexural properties. The fibers tended to be more effective when they were placed close to the tensile stress side of the denture base (Agha et al., 2016). Placing glass fibers in the area of neutral stress increased fracture toughness only, while placing them near the compressive side increased the flexural modulus (Dalkiz et al., 2012).

2.8.2 Adhesion to the polymer matrix and impregnation

Adequate adhesion of the fibers to the polymer matrix is a crucial factor for the strength of the fiber-reinforced structure. Silane coupling agent has been used successfully in improving the adhesion between polymers and glass fibers (Basant and Reddy, 2011; Solnit, 1991; Vallittu, 1997b). PMMA-preimpregnated glass fibers allow interfacial adhesion in order to transfer stresses from the weak polymer matrix to the stronger fibers. The preimpregnated fibers need further impregnation with a thin mixture of the polymer used in the final construction. The remaining small quantities of free monomers within that thin mixture penetrate, dissolve, and plasticize the porous preimpregnation polymer of the reinforcement (Vallittu, 1999, 2018).

2.8.3 Fiber orientation

Regarding the fiber orientation, they can be continuous unidirectional (rovings and yarns), continuous bidirectional (weaves and fabrics), continuous random oriented (mat), or discontinuous (short and chopped) random or oriented fibers. Continuous unidirectional fibers are anisotropic, with high strength and stiffness only in 1 direction (fibers direction) and therefore, they are suitable for applications when the direction of the highest stress is known. Their reinforcing efficiency (Krenchel's factor) is theoretically 100%. While woven fibers are orthotropic, reinforcing the polymer in 2 directions with a theoretical reinforcing efficiency of 50% (Narva et al., 2001; Vallittu, 1999).

Although continuous fibers provide excellent reinforcement (Vallittu, 1997a), some technical difficulties have been encountered while placing them within the high-viscosity dough of multiphase polymers such as PMMA (Vallittu et al., 1994b). Alternatively, low concentration woven fibers can enhance the mechanical features while maintaining ease of handling (Vallittu, 1999). Their stretching ability enhances the polymer toughness and resists crack propagation. They are also suitable in cases where the load direction is unknown or in limited spaces where the use of unidirectional fibers may not be possible. The polymer strain at fracture increased when woven reinforcement was used. Then, from a clinical perspective, they can be placed in certain areas within the prosthesis where greater toughness is needed, such as in areas of precision overdenture attachments (Vallittu, 1999).

Randomly-oriented fibers or short fibers have similar isotropic properties in all directions. This means that the strength of the reinforced structure is not related to the fracture force (Lastumäki et al., 2001).

2.9 The use of strain gauge for strain measurement

2.9.1 Strain

Strain (ε) is a measurement of deformation within an object during force application. It is the fractional change in length, width, or height of an object when a force is applied along that dimension. It can be calculated from the following equation:

$$\text{Strain } (\varepsilon) = \Delta L / L \qquad (1),$$

where: L is the length and ΔL is the change in length.

From the previous equation, strain can be defined as the ratio of change in length to the initial length. When ΔL is positive, the object is under tension and this deformation is known as tensile strain. When ΔL is negative, the object is under compression and this deformation is known as compressive strain. Since the obtained magnitude is usually small, the strain is often reported as microstrain (με = ε x 10-6) and is a dimensionless quantity (Darvell, 2018). Detecting a minor mechanical deformation that occurs during force application is a sensitive procedure. As a result, strain gauges have long been regarded as the most accurate and widely used method for performing such analyses (Elsyad et al., 2016).

2.9.2 Strain gauge

The strain gauge was introduced by Edward E. Simmons in 1938 (Perry, 1984). It is an electrical device that can transform the mechanical deformation within a body into a measurable signal (Craig et al., 1967). It is a sensor that responds to a material's expansion or contraction (Segil, 2019). The strain gauge has a sensitive element in the form of a small wire (metallic foil) bonded to a baking material fabricated from polyimide (Figure 1).

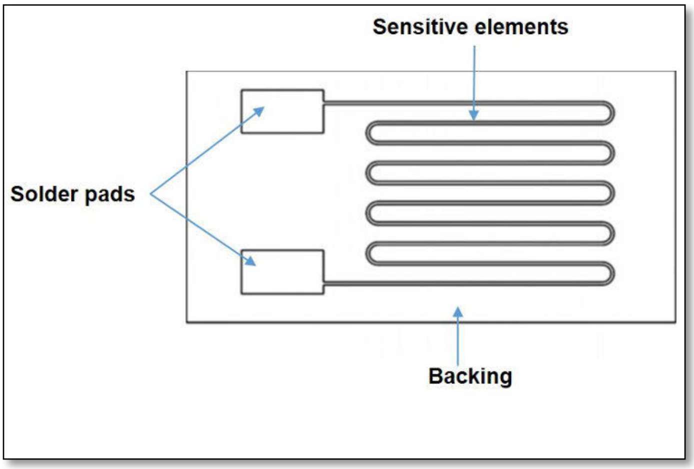

Figure 1. Schematic representation of a strain gauge. Modified from (Segil, 2019).

The resistance of the gauge is detected by the length of the looped wire. The longer the wire, the greater the resistance. This structure provides a strain/resistance relationship that is determined by the manufacturer. This relationship is called gauge factor (GF) that can be defined as the ratio of fractional change in resistance to the relative change in its length (Bøving, 1989; Montero et al., 2011). The gauge factor equations are:

$$GF = \frac{\Delta R/R}{\Delta L/L} \tag{2}$$

Or

$$GF = \frac{\Delta R/R}{\varepsilon} \quad (3),$$

where: GF is the gauge factor, ΔR is the change in resistance, R is the initial resistance, ΔL is the change in length, L is the initial length, and ε is the strain.

The electrical resistance used for strain gauges is usually either 120 Ω or 350 Ω, while the commonly used gauge factor is around 2, according to the purpose of the experiment and operation system used for strain measurement (Bolton, 2015).

As the material expands or contracts, the metal wire gets longer or shorter with the material, changing the electrical resistance of the wire. In this way, the voltage change in the wires can be related to the change in strain (Segil, 2019). Most of the strain gauge measurement devices automatically link the voltage change to the strain, so the device output is the actual strain.

2.9.3 Strain measurement

Detecting small mechanical deformations needs a sensitive method that can measure small changes in resistance. The "Wheatstone bridge circuit" is regarded as a reliable method for strain measurement within a body. The simple Wheatstone bridge circuit has 4 resistors (R_1, R_2, R_3, and R_4), 2 parallel legs with 2 resistors in each series, an excitation voltage or input voltage (V_{in}) is applied across the bridge from top to bottom, and a measured voltage or output voltage (V_{out}) is applied across the bridge from right to left (Figure 2).

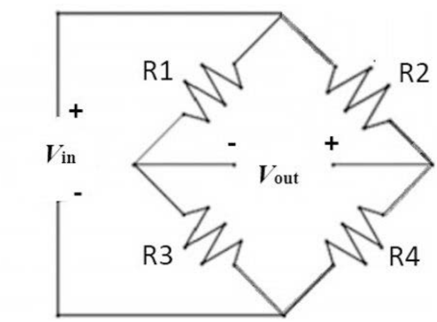

Figure 2. Basic Wheatstone bridge arrangement modified from (Segil, 2019).

The bridge is under a state of balance when the 4 resistors (R_1, R_2, R_3, and R_4) are equal to each other. In this case, the output voltage is equal to zero as $R_1R_3 = R_2R_4$. In other order, R_1R_3 - R_2R_4 = zero. Based on this relationship, the output voltage can be calculated from Ohm's law by using equation 4 as follows:

$$V_{out} = V_{in}\left[\frac{R_1R_3 - R_4R_2}{(R_2 + R_3)(R_1 + R_4)}\right] \tag{4}$$

Practically, a load applied to the body of the bridge makes it unbalanced. The changes in resistance at any leg of the Wheatstone bridge will result in a non-zero output voltage and then the bridge will be unbalanced. One, 2, or 4 strain gauges can be placed within the Wheatstone bridge according to the purpose of measurement (Keil, 2017). In 1 gauge configuration (Quarter Bridge), 1 of the 4 resistors is replaced with a strain gauge for measuring either tension or compression. While in 2 gauges configuration (Half Bridge), 2 of the 4 resistors are replaced with 2 strain gauges to measure either tension, compression, or even both of them. Finally, in 4 gauges configuration (Full Bridge), all of the 4 resistors are replaced with 4 strain gauges. R_1 and R_3 will measure tensile strain while R_2 and R_4 will measure compressive strain.

2.9.4 Application of strain gauge in dental studies

Denture base strain has been evaluated in many in vitro studies (Elsyad et al., 2016; Gonda et al., 2007; Takahashi et al., 2013, 2015) and in vivo studies (Elsyad et al., 2013; Elsyad et al., 2020) using strain gauges. Gonda et al. (2007) used strain gauges for measuring the denture base deformation of reinforced mandibular overdentures. Clinical denture base deformation was evaluated for mandibular 2-implant-supported overdentures with 3 different attachments using 6 linear strain gauges fixed to the overdenture lingual surface opposite to the abutments and the midline (Elsyad et al., 2020).

In addition to denture base deformation, strain gauges were used to detect dental implant strains (Grobecker-Karl et al., 2020; Karl et al., 2005; Pham et al., 2019), bone strains after implant placement (Akça et al., 2005, 2007; Cehreli et al., 2005), and to assess the fit of implant frameworks fabricated using computer-aided design/computer-aided manufacturing (CAD/CAM) (Abduo et al., 2011).

3 Aims

The series of studies included within the current thesis were conducted with the aim to investigate the effect of fiber reinforcement on the mechanical properties and strains of implant overdentures. The specific aims of the included studies were:

1. To evaluate the effect of using a different number of layers of bidirectional woven E-glass fiber reinforcement placed at different positions on the load-bearing capacity of simulated locator-retained overdenture specimens (study I).
2. To evaluate the effect of bidirectional E-glass fiber reinforcements position on the fatigue resistance of a simulated single locator-retained overdenture (study II).
3. To compare the flexural strength and modulus of soft liner-retained overdentures with ball-and-socket–retained overdentures and to evaluate the effect of using unidirectional and bidirectional E-glass fiber reinforcements on the mechanical properties of soft liner-retained overdentures (study III).
4. To evaluate the effect of unidirectional E-glass fiber reinforcement on the mid-line denture base strains of SIO (study IV).

4 Materials and Methods

4.1 Study I

4.1.1 Specimens fabricated for testing

For this study, locator-retained overdenture-simulating specimens were designed. A clear autopolymerizing denture base resin (Palapress; Kulzer GmbH) was used to obtain seventy specimens (65 mm long, 5 mm high, and 10 mm wide). The mixture was prepared according to the manufacturing instructions. The powder to liquid ratio of the autopolymerizing resin was 10 g: 7.0 mL. Each specimen had a locator's metal housing in the middle. A 4 mm diameter locator model analog and a titanium matrix housing (2.3 mm height x 5.5 mm diameter) with inner retention insert (regular retention) were used and purchased from the same manufacturer (Zest Anchors LLC).

The fiber reinforcement system (Stick Net; StickTech Ltd) (SN) selected for this study is a bidirectional silanated E-glass fiber weave preimpregnated with porous PMMA. The thickness of a single fiber weave is 0.06 mm.

Three groups were designed for testing. The 1st group was the control, which had no reinforcement (n=10). The 2nd group (2L) with 2 layers of SN fiber weaves was subdivided according to the location of the fiber weaves into 3 subgroups: 2L-A subgroup with 2 SN fiber layers above the metal housing (n=10); 2L-N subgroup with 2 SN fiber layers next to the metal housing (n=10); and 2L-A+2L-N subgroup with a combination of 2 SN fiber layers above the metal housing and 2 SN fiber layers next to it (n=10). The 3rd group (4L) with 4 layers of SN fiber weaves was subdivided according to fiber weaves location into 4L-A subgroup with 4 SN fiber layers above the metal housing (n=10); 4L-N subgroup with 4 SN fiber layers next to the metal housing (n=10); and 4L-A+4L-N subgroup with a combination of 4 SN fiber layers above the metal housing and 4 SN fiber layers next to it (n=10). (Figures 3 and 4).

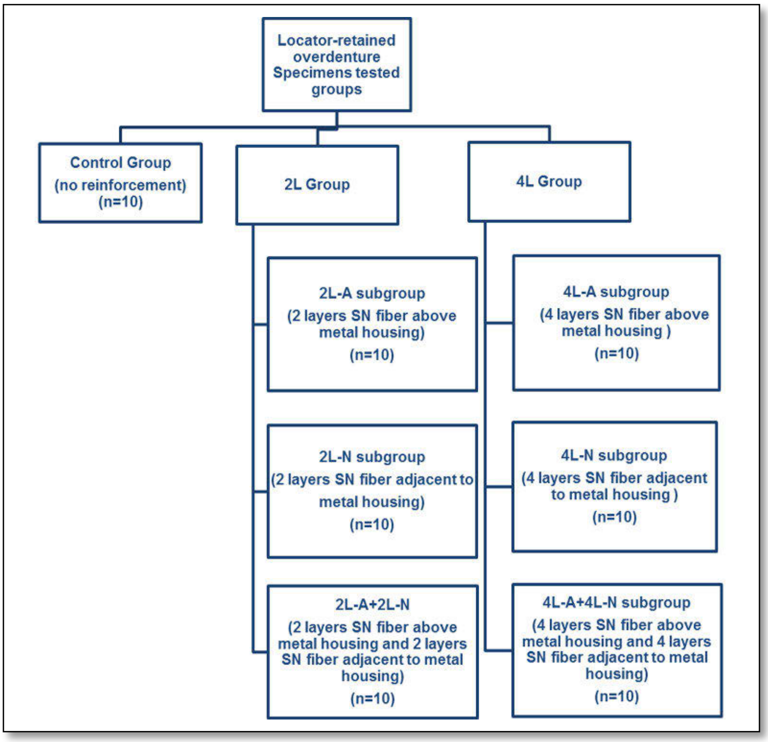

Figure 3. Diagram of the test groups of locator-retained overdenture specimens according to the number and location of SN fiber layers. Adopted from original publication I.

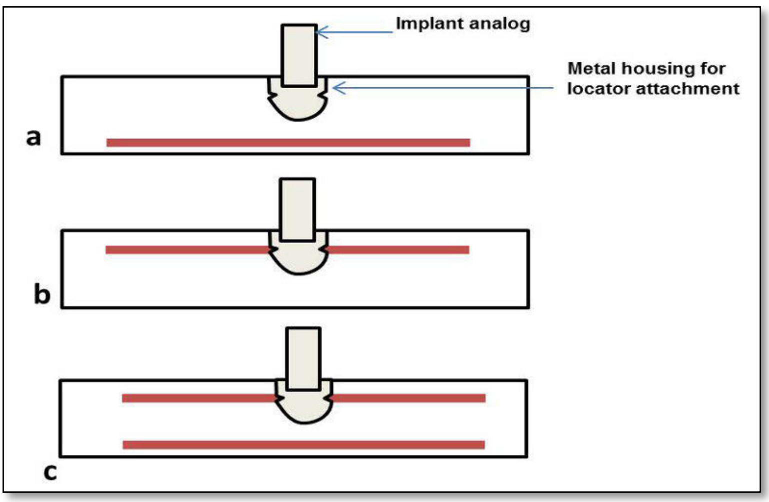

Figure 4. Schematic view of SN fiber weaves positioning (red lines) for each subgroup of groups 2L and 4L: a. above the metal housing in subgroups 2L-A and 4L-A, b. Next to the metal housing in subgroups 2L-N and 4L-N, c. Next to and above the metal housing in subgroups 2L-A+2L-N and 4L-A+4L-N. Adopted from original publication I.

The control group was fabricated by inserting the metal housing into the center of a polyvinyl siloxane lab-putty mold (Lab Putty; Coltène) (5.2 × 10.2 × 65.2 mm). Then, the whole mold was filled with an acrylic resin mixture.

For the fiber-reinforced test specimens, equal layers of SN fiber sheets (60 mm length × 9 mm width) were cut with scissors and wet with a thin powder-liquid mixture of auto-polymerizing acrylic resin between 2 plastic sheets. The specimens of the 2L-A subgroup were fabricated by partially filling the lab-putty mold with a 4 mm layer of acrylic resin while the metal housing was centralized in the mold. Then, 2 layers of wet SN fiber weaves were placed on top of each other, followed by another layer of resin mixture. The 2L-N subgroup specimens were obtained by creating a hole in the middle of 2 fiber sheets using an explorer (LM 5-8 Si; LM-DENTAL). Gentle pressure was applied with the probe to displace the fibers laterally creating a space (<5.5 mm in diameter) for holding the metal housing with some friction. The housing and the surrounding wet SN fiber layers were placed as 1 piece inside the mold while the rest of the mold was filled with an acrylic resin mix. To get the specimens of the 2L-A+2L-N subgroup, the same steps as for the 2L-N subgroup specimens were followed, plus adding 2 extra layers of SN fiber weaves above the metal housing after filling the mold with a 4 mm layer of acrylic resin, and finally pouring another layer of resin mix to cover the fibers and fill the mold. The previous procedures were repeated using 4 layers of SN fiber weaves to fabricate the test specimens for the other 3 subgroups 4L-A, 4L-N, and 4L-A+4L-N.

Glass plates were used for covering the molds while polymerizing the specimens in distilled water maintained at 55 ±2 °C under air pressure of 300 kPa for fifteen minutes in a pneumatic polymerizing unit (Ivomat IP3; Ivoclar Vivadent AG). Successively finer grades of silicon carbide abrasive papers from P300 to P1200 were used to wet ground the cured specimens to the predetermined dimensions (5×10×65 mm), and then the samples were stored dry at room temperature (23 ±1 ºC) for 24 hours before testing.

4.1.2 Mechanical testing

The specimens were submitted to a static 3-point bending test in the air to detect the flexural strength, modulus, and strain values at a speed of 5 mm/min using the implant analog for load application (Figure 5). The distance between the supports of the test specimens was adjusted to be 50 mm.

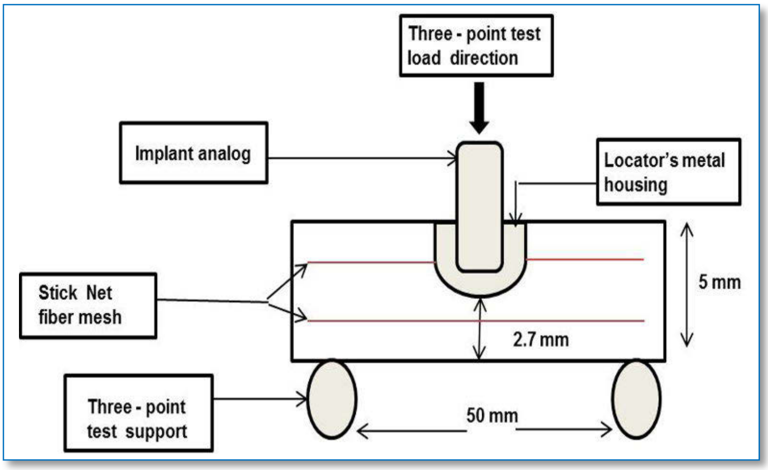

Figure 5. Flexural strength test set up. Adopted from original publication I.

4.1.3 Visual examination

After testing, tested specimens were visually examined to detect any differences in failure modes.

4.1.4 Scanning electron microscope (SEM) examination

Before SEM examination, representative test specimens were wet ground with decreasing abrasiveness silicon carbide papers (1000-, 1200-, 4000-grit) and gold sputter coated.

4.2 Study II

4.2.1 Specimens fabricated for testing

Forty-eight specimens mimicking overdentures with locator's metal housing in the middle were tested. The resin type, specimen dimensions, attachment, and the reinforcing fiber system used in this study were the same as those used in study I.

Two test groups were assessed: control group without reinforcement (n=12) and 4L group, which had 4 layers of SN fiber weaves. The 4L group was subdivided into 3 subgroups according to the positioning of the fiber weaves. The 1st subgroup (4L-A) had 4 SN layers above the metal housing (n=12). The 2nd subgroup (4L-N) had 4 SN layers surrounding the metal housing (n=12). The 3rd subgroup (4L-A+4L-N) had a combination of 4 SN layers above the metal housing and 4 SN layers surrounding it (n=12) (Figure 4). Test specimens were fabricated in the same way as explained before in study I.

4.2.2 Mechanical testing

The staircase method was used to evaluate the compressive fatigue limits (CFL) for the test specimens at 10 000 cycles. A universal testing machine (Model LRX; Lloyds Instruments Ltd) was used to conduct the fatigue resistance test at a crosshead speed of 60 mm/min and a frequency of 0.5 HZ in a water bath at 37ºC. The load was applied at the locator metal housing using an implant analog (Figure 6).

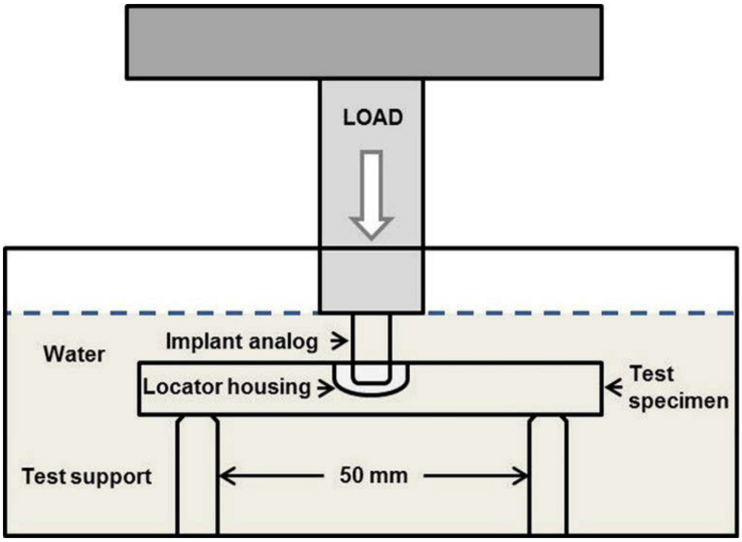

Figure 6. Flexural fatigue test set up in the water. Adopted from original publication II.

In this 'up and down" method, the specimens were tested sequentially. In this way, the first specimen was tested at the initial stress level detected from the preliminary data. The stress level for the next specimen was increased or decreased by a given interval depending on whether the first specimens survived or failed. This process was continued until all the specimens allocated for the experiment were tested. The magnitude of load by which the level was changed was 30 N. Data analysis was based on the failure versus no-failure events. The CFL and its standard deviation (S) were calculated according to the following equations (Bijelic-Donova et al., 2016; Ornaghi et al., 2014; Pollak et al., 2006):

$$CFL=X_0 + d\,(A/N \pm 1/2) \quad (5)$$

$$S=0.53\cdot d \quad (6),$$

where CFL is the compressive cyclic fatigue limit, X_0 is the lowest load level at which failure occurs, d is the fixed load increment (30 N) used in the sequential test, and S is the standard deviation. A and N are explained in Table 1 (Bijelic-Donova et al., 2016;

Ornaghi et al., 2014; Pollak et al., 2006). Specimens that did not break after the 10 000 loading cycles were subjected to a static loading to evaluate the flexural strength after fatigue testing, named here as post fatigue flexural strength (PFFS). A 95% confidence interval analysis was conducted for the CFL values of the tested groups. In addition, creep values were collected from the test machine and analyzed.

Table 1. Method for analyzing the staircase test data. Adopted from original publication II.

		Control group		
Load (L)	**Stress level (i)**	**Failures (N)=Σ ni ni**	**A=Σ i.ni i.ni**	***B =Σ i^2.ni i^2.ni***
130	0	0	0	0
160	1	3	3	3
190	2	3	6	12
		N= 6	**A= 9**	**B= 15**
		4L-A subgroup		
Load(L)	**Stress level (i)**	**Failures (N)=Σ ni ni**	**A=Σ i.ni i.ni**	***B =Σ i^2.ni i^2.ni***
200	0	0	0	0
230	1	2	2	2
260	2	4	8	16
		N= 6	**A= 10**	**B= 18**
		4L-N subgroup		
Load (L)	**Stress level (i)**	**Failures (N)=Σ ni ni**	**A=Σ i.ni i.ni**	***B =Σ i^2.ni i^2.ni***
170	0	0	0	0
200	1	5	5	5
230	2	1	2	4
		N= 6	**A= 7**	**B= 9**
		4L-A+4L-N subgroup		
Load (L)	**Stress level (i)**	**Failures (N)=Σ ni ni**	**A=Σ i.ni i.ni**	**B =Σ i^2.ni i^2.ni**
220	0	0	0	0
250	1	4	4	4
280	2	2	4	8
		N= 6	**A= 8**	**B= 12**

The PFFS values of the tested specimens were compared with the static flexural strength values of the similar groups previously evaluated in **study I**. **Study I** test groups were named as controlx group and 4Lx group which had 3 subgroups named as 4L-Ax, 4L-Nx, and 4L-Ax+4L-Nx.

4.2.3 Visual examination

After fatigue and post-fatigue testing procedures, the tested specimens were examined visually to detect any difference in failure modes.

4.2.4 SEM examination

Representative tested specimens were prepared in the same way as described before in study I and subjected to SEM examination (JSM 5500; JEOL Ltd)

4.3 Study III

4.3.1 Specimens fabricated for testing

Eighty overdenture-simulating specimens were fabricated from the same resin type and with the same dimensions as those fabricated in studies I and II. The used ball stud attachment consists of a metal matrix (3.1 mm in height × 3.6 mm in diameter) (Dalbo® Plus Female Part TE basic; Cendres + Métaux) and a ball abutment (2.25 mm wide × 2 mm high) (DYNA Octalock®; Dyna Dental Engineering). The resilient liner matrices were fabricated from vinyl polysiloxane soft lining material (RELINE™ II Soft; GC Corp) (Figure 7).

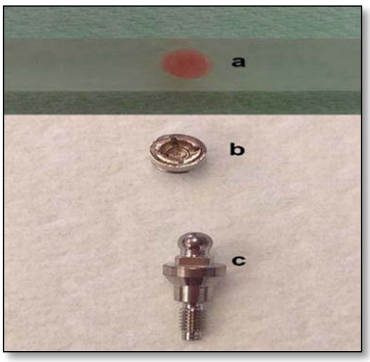

Figure 7. The attachment system and matrix used in testing procedures. a: silicone-based soft liner matrix; b: metallic matrix; c: ball abutment. Adopted from original publication III.

The fiber reinforcements used in this study were bidirectional SN (Stick Net; GC Corp) and unidirectional stick (Stick; GC Corp) E-glass fiber reinforcements. Both reinforcements are silanated E-glass fibers preimpregnated with porous PMMA (Figure 8).

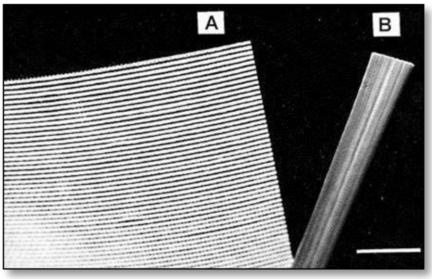

Figure 8. Woven SN reinforcement (A) and continuous unidirectional Stick fiber reinforcement (B). (Bar = 10 mm.) (Vallittu, 1999)

The test specimens were allocated into 4 groups. Group 1 had overdenture specimens with a metal matrix and without reinforcement (n=20). Group 2 had overdenture specimens with a silicone resilient liner matrix (n=20). Group 3 had overdenture specimens reinforced with one bundle of Stick glass fibers placed above the silicone matrix (n=20). Finally, group 4 had overdenture specimens reinforced with 4 layers of SN fiber weaves placed above the silicone matrix as recommended by previous studies (Gibreel et al., 2018, 2019) (n=20) (Figure 9).

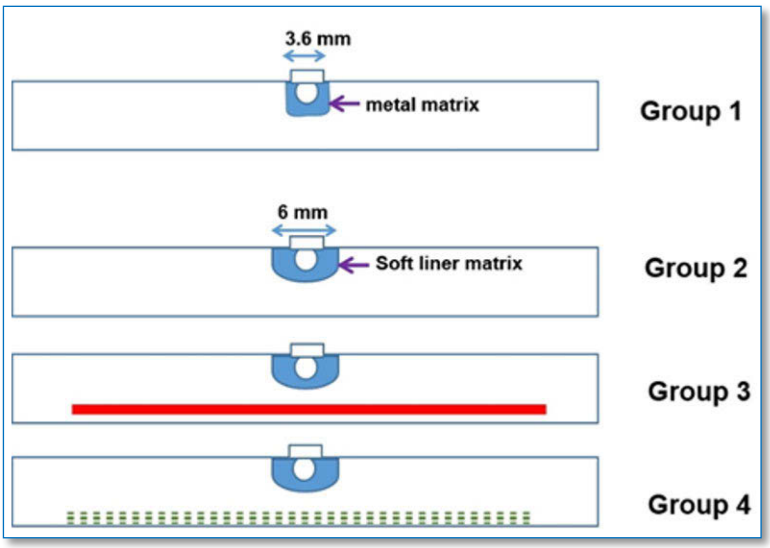

Figure 9. Test groups of ball-retained overdenture specimens according to the matrix material and reinforcement. The red line represents the unidirectional fiber bundle and the green lines represent the bidirectional fiber weaves. Adopted from original publication III.

Specimens for group 1 were prepared by pouring a mixture of acrylic resin (Palapress; Kulzer Gmbh) to fill a polyvinyl siloxane laboratory putty mold (Lab

Putty; Coltène) (5.2×10.2×65.2 mm) while a metal matrix is placed in the middle of it. For the other 3 groups, a silicone mold of the same size with a mid-projection (3.1 mm high × 6 mm wide) was used to keep a standardized space for the resilient liner matrix within the finished specimen. Group 3 specimens were fabricated using Stick fiber bundles cut into equal length (60 mm) and wet with a thin mixture of autopolymerizing acrylic resin (Palapress; Kulzer Gmbh) between 2 plastic sheets. They were then placed within the mold when it is filled with 4 mm of acrylic resin, and then the rest of the mixture was poured to fill the mold. For fabricating group 4 specimens, 4 layers of SN fiber weaves 60 mm in length and 9 mm in width were cut, wet, and placed inside the mold in the same way as the Stick fibers. That was followed by covering the specimens with glass plates. Finally, the specimens were polymerized, finished, and polished to the predetermined size as described before in studies I and II.

A primer (RELINE™ II; GC Corp) was applied to the mid-holes of the specimen of groups 2, 3, and 4, then, the holes were loaded with a soft liner mix. That was followed by placing each specimen over a ball abutment fixed to an acrylic model and allowing it to set for 5 minutes. Excess liner material was removed with a scalpel. The final specimen had a silicone soft liner matrix, which was 1.88 mm thick bilaterally on both sides of the ball abutment.

4.3.2 Mechanical testing

Half of the specimens from each group were stored in water at room temperature (23 ±1ºC) for 24 hours while the rest were stored in water at 37ºC for 30 days before testing. The flexural strength and modulus values of the test groups were measured using a static 3-point bending test carried out using a universal testing machine (Model LRX; Lloyds Instruments Ltd). The testing was performed in air at a speed of 5 mm/min using the implant with ball abutment for load application (Figure 10). The distance between the supports of the test specimens was 50 mm. The maximum load values exerted at failure were recorded in Newton (N). Elastic modulus values (GPa) were collected directly from the machine. Flexural strength (Fs) was then calculated from the following equation (Vallittu, 1999):

$$\text{Flexural strength (MPa)} = 3PL/2bd^2 \qquad (7),$$

where P=maximum load (N), L=span length (50 mm), b=specimen width (10 mm), and d=specimen thickness (5 mm).

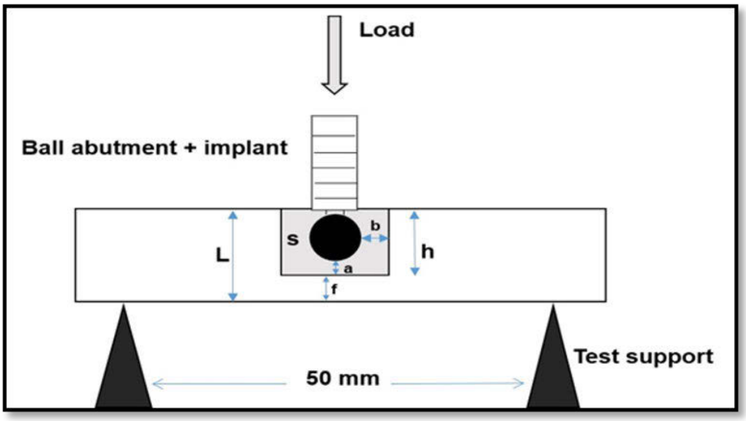

Figure 10. Three-point bending testing procedure. S: housing made of a silicone-based soft liner; L: 5 mm; f: 1.9 mm; h: 3.1 mm; a: 1 mm; b: 1.88 mm in thickness. Adopted from original publication III.

4.3.3 Visual examination

After mechanical testing, specimens were visually examined to detect any differences in failure modes.

4.3.4 SEM examination

Representative tested specimens were prepared in the same way as described before in studies I and II before being submitted to SEM examination. In addition, the fractured surfaces and bonding interphase between the soft liner and denture base resin before and after water storage were examined with SEM.

4.4 Study IV

4.4.1 Experimental overdentures fabrication

For this in vitro study, an experimental acrylic resin cast was obtained by duplicating a commercially available mandibular edentulous stone cast without undercuts. One dental implant 11.5×3.6 mm (Helix DC; Dyna) was attached in the mid-line area of the cast. To simulate the mucosal lining, a 2-mm-thick layer of autopolymerizing silicone resilient denture liner (RELINE^{TM} II Soft; GC Corp) was applied to the residual ridge of the cast. Metal matrices (3.1 mm in height and 3.6 mm in diameter) (Dalbo Plus Female Part TE basic; Cendres + Métaux) were embedded within the denture base's resin, while a ball abutment (2.25 mm in width and 2 mm in height) (Dyna) was threaded to the mid-line implant fixture (Figure 11).

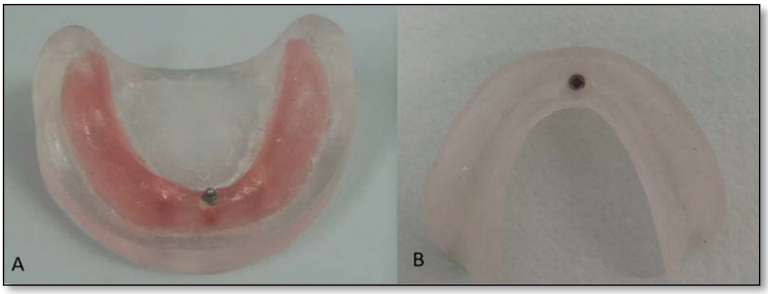

Figure 11. A, ball attachment inserted in the mid-line area of the cast. B, metal matrix in the mid-line area of the overdenture base. Adopted from original publication IV.

Twenty-four experimental overdentures were fabricated. One experimental mandibular overdenture was fabricated over the edentulous cast simulating the shape of a record rim. Then, it was duplicated to get the other overdentures. Four test groups were designed for testing. In group AP (n=6), the overdentures were made from autopolymerizing acrylic resin (Palapress; Kulzer GmbH) without fiber reinforcement. Group APF (n=6) had overdentures made from autopolymerizing acrylic resin and reinforced with unidirectional Stick fiber bundles (Stick; GC Corp) over the residual ridge and the ball matrix. Group HP (n=6) had overdentures made from heat-polymerized acrylic resin (Paladon 65; Kulzer GmbH) without fiber reinforcement. Finally, group HPF had overdentures made from heat-polymerized acrylic resin and reinforced with unidirectional Stick fiber bundles over the residual ridge and the ball matrix. One fiber bundle was used for each denture. The Stick fiber bundles (100 mm long) were wet with a thin powder-liquid mixture of autopolymerizing or heat-polymerized acrylic resin, placed in a mold that was filled with almost a 2 mm layer of acrylic resin, and finally covered with the rest of the resin material. The polymerization process of the autopolymerizing acrylic resin dentures was carried out in distilled water maintained at 55 ±2ºC under air pressure of 300 kPa for 15 minutes inside a pneumatic polymerization unit (Ivomat IP3; Ivoclar Vivadent AG) while the heat-polymerized resin dentures were cured in a hot water polymerization unit (Kulzer GmbH). The polymerization was accomplished using the rapid boiling method according to the manufacturer’s instructions where the dentures were placed in a water bath at 80 ºC for 15 minutes, boiled at 100 ºC for 20 minutes, and finally cooled in the water bath.

4.4.2 Strain measurement

Biaxial 90-degree rosette strain gauges (KFGS-1-120-D16-23L1M2S; Kyowa Electronic Instruments Co) (2 measuring directions, 0 and 90 degrees) were used for strain measurement. One gauge was cemented at the anterior mid-line of each overdenture above the ball attachment, connected to the sensor interfaces (PCD-300A; Kyowa Electronic Instruments Co), and controlled by a personal computer. Strain values were recorded from each channel separately (Figure 12) while at the same time, a 100-N vertical occlusal load was applied bilaterally through a metal bar (5×5×100 mm) positioned on the occlusal surface and running across the 2 sides in the first molar areas. A universal testing device (Model LRX; Lloyd Instruments Ltd) with a crosshead speed of 0.5 mm/min was used for load application (Figure 13).

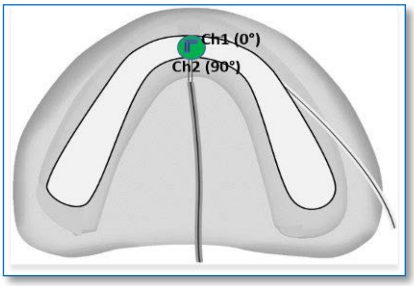

Figure 12. Schematic view showing the positioning of the biaxial strain gauge on the mid-line area of the overdenture during strain measurements. Adopted from original publication IV.

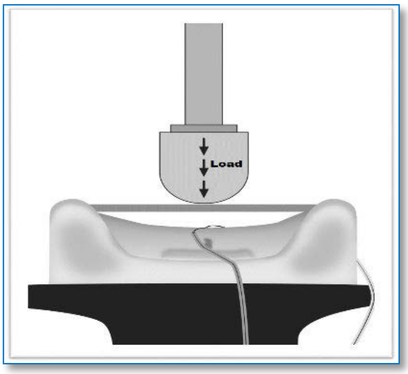

Figure 13. Schematic figure showing the experimental cast placed on the loading machine during testing procedures. Adopted from original publication IV.

The strain of each experimental denture was recorded for 2 minutes. The measurements were repeated 5 times for each denture with an intervening interval of 5 minutes for recovery. The strain values for both channels Ch1 and Ch2 (εa and εb) and the deflection values of the denture base were recorded and submitted for statistical analysis.

5 Statistical analysis

Data obtained from studies I-IV were analyzed with SPSS version 21 (studies I, II, and III) and version 26 (study IV) (Statistical Package for Social Science, SPSS Inc, Chicago, IL, USA). An analysis of variance (ANOVA) test at the significance level of $p=0.05$ was used followed by Tukey's post hoc analysis in all studies. Additionally, a 2-way ANOVA ($\alpha=0.05$) was used in study I to detect the effect of the number and position of fiber layers as the independent variables on the evaluated flexural properties. In study II, a 2-way ANOVA was used to detect the effect of fatigue cyclic loading, the position of the fiber layers, and their interaction on the flexural strength. Also, it was used in study IV to detect the effect of acrylic resin type and reinforcement as the independent variables on overdenture base strains and deflection. A 3-way ANOVA was conducted in study III to reveal if the matrix material, reinforcement, and duration of water storage affected the flexural strength and modulus. Failure modes were not statistically analyzed.

6 Results

6.1 Flexural properties of simulated Locator-retained overdenture system (study I)

The 2-way ANOVA revealed that the number of fiber layers significantly influenced the flexural strength (p=0.03) while the position did not affect it significantly (p=0.153). The interaction between the number of layers and location was not significant (P=0.203). Furthermore, the 2 variables did not have a significant effect on either the flexural modulus (p=0.940 and p=0.147 respectively) or strain (p=0.529 and p=0.309 respectively). The 1-way ANOVA revealed a statistically significant difference in the flexural strength values among the tested groups (p<0.001). No significant differences were found for flexural modulus (p=0.195) or strain values among the groups. The flexural strength of the control group was significantly lower than the 4L-A subgroup (P=0.001) and 4L-A+4L-N subgroup (p<0.001) (Tukey's post hoc analysis) (Table 2) (Figure 14). The rest of the subgroups were not statistically significant from the control (p>0.05). Figure 15 shows the load-deflection curves of the evaluated groups.

Table 2. Mean flexural strength (MPa) of the tested groups. Modified from original publication I.

Group	Subgroup	FS (MPa) Mean ± standard deviation (SD)
Control	__	92 ±14^{a}
2L (2 layers SN fiber)	2L-A (2 layers SN fiber above metal housing)	108 ±19^{abc}
	2L-N (2 layers SN fiber next to metal housing)	103 ±9^{abc}
	2L-A+2L-N (2 layers SN fiber above metal housing and 2 layers SN fiber next to metal housing)	105 ±10^{ab}
4L (4 layers SN fiber)	4L-A (4 layers SN fiber above metal housing)	116 ±7^{c}
	4L-N (4 layers SN fiber next to metal housing)	106 ±12^{abc}
	4L-A+4L-N (4 layers SN fiber above metal housing and 4 layers SN fiber next to metal housing)	117 ±6^{bc}
one-way ANOVA (P value)		<0.001

Key: p<0.05 significant. Same superscripted letters indicate a non-significant difference among different groups when compared by Tukey multiple comparisons post hoc analysis (p>0.05).

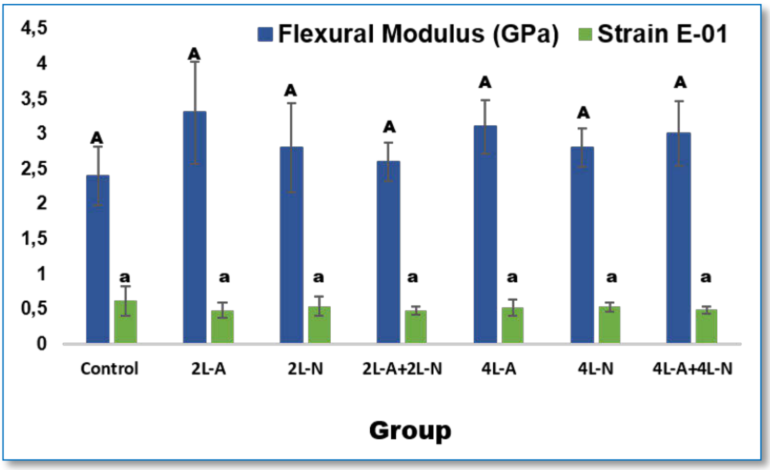

Figure 14. Flexural modulus and strain values of the evaluated groups. Same superscripted uppercase letters indicate a non-significant difference between flexural modulus values among different groups when compared by Tukey multiple comparisons post hoc analysis ($p>0.05$). The same superscripted lowercase letters indicate a non-significant difference between strain values among the different groups when compared by Tukey multiple comparisons post hoc analysis ($p>0.05$).

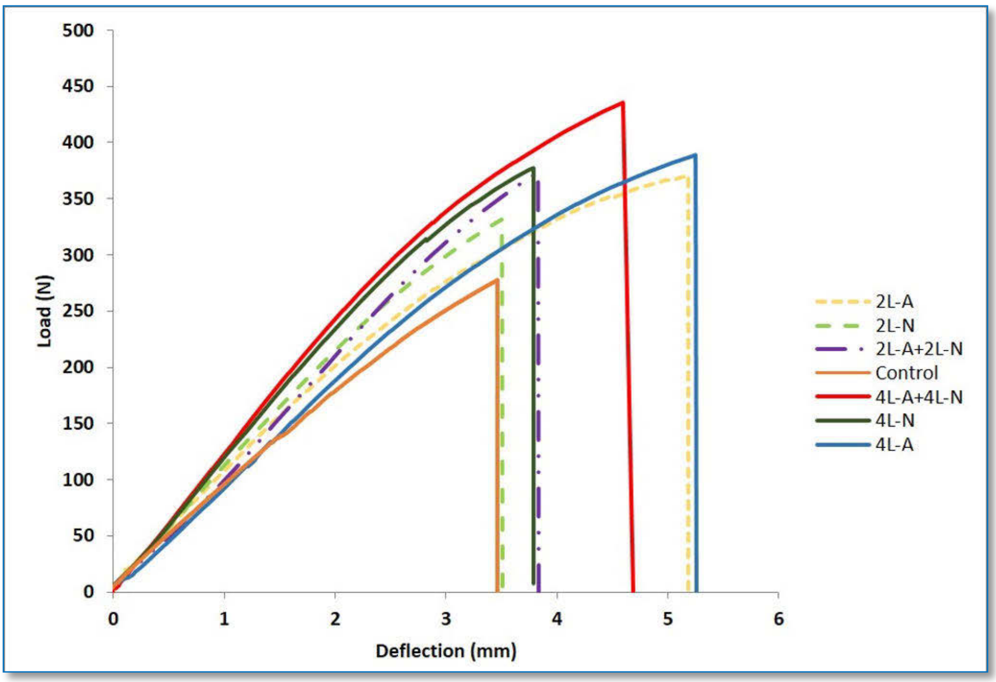

Figure 15. Load-deflection curves of the evaluated groups.

Visual examination showed that the fracture path was arrested at the fiber layers surrounding the housing only in 4 specimens of the 4L-N subgroup and in all of the specimens from the 4L-A+4L-N subgroup (Figure 16 and 17). The rest of the specimens fractured completely into 2 pieces (Table 3).

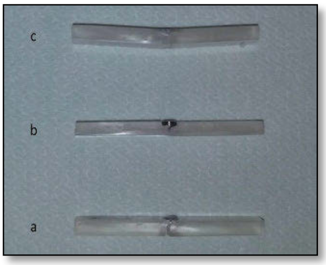

Figure 16. Fractured test specimens from group 4L: a. 4L-N subgroup, b. 4L-A subgroup, c. 4L-A+4L-N subgroup. Adopted from original publication I.

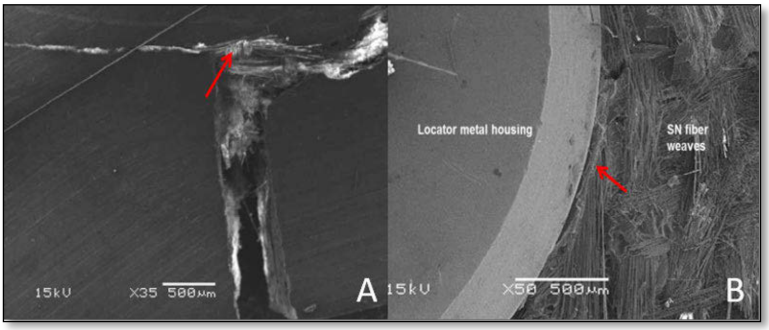

Figure 17. Scanning electron photomicrographs of subgroup 4L-N. A, Incomplete fracture with some intact fibers (red arrow) next to the metal housing. Original magnification ×35. B, Fiber weave location (red arrow) around the metal housing. Original magnification ×50.Adopted from original publication I.

Table 3. Fracture mode of test specimens for the investigated groups. Adopted from original publication I.

Group	Subgroup	Fracture mode	
		Fracture arrested at fibers	Specimens fractured into 2 pieces
Control	-	-	10/10
2L (2 layers SN fiber)	2L-A	-	10/10
	2L-N	-	10/10
	2L-A+2L-N	-	10/10
4L (4 layers SN fiber)	4L-A	-	10/10
	4L-N	4/10	6/10
	4L-A+4L-N	10/10	-

6.2 Fatigue resistance of a simulated single locator-retained overdenture system (study II)

The results of compressive cyclic fatigue limits showed an overlap only between the 95% confidence interval values of 4L-A and 4L-A+4L-N subgroups, meaning that they were statistically non-significant (Table 4). The rest were statistically different as shown in Table 4.

Table 4. CFL values in Newton (N) of the tested groups at 95 % confidence intervals. Adopted from original publication II.

Group	Subgroup	CFL	Standard deviation (SD)	Standard error (SE)	T-value	Confidence interval (CI)	Lower limit	Upper limit
Control	-	190	15.9	4.59	2.20	10.09	179.91	200.09
4L	4L-A	265	15.9	4.59	2.20	10.09	254.91	275.09
	4L-N	220	15.9	4.59	2.20	10.09	209.91	230.09
	4L-A+4L-N	275	15.9	4.59	2.20	10.09	264.91	285.09

The 1-way ANOVA revealed that the PFFS values were significantly different ($p<0.001$), while the difference in creep values was non-significant among the control group and reinforced subgroups ($p>0.05$) (Table 5). The post hoc Tukey HSD test revealed that the PFFS values of the 4L-A and 4L-A+4L-N subgroups were significantly higher than the control group ($p<0.001$) and the 4L-N subgroup ($p=0.004$ and $p=0.005$). Also, there was a non-significant difference in the PFFS values between the control group and the 4L-N subgroup ($p=0.828$) and between the 4L-A and 4L-A+4L-N subgroups ($p>0.05$).

Table 5. Mean flexural strength and creep values of tested groups. Adopted from original publication II.

Test condition	After 10^4 cycles				Without 10^4 cycles				One-way ANOVA (p value)
Group	Control		4L		Controlx		4Lx		
Subgroup	-	4L-A	4L-N	4L-A+4L-N	-	4L-Ax	4L-Nx	4L-A+4L-Nx	
FS (MPa) Mean ±SD	53 ±8^a	74. ±15^b	57 ±5^a	74 ±12^b	92.4 ±13.9^c	116 ±7.3^d	106 ±11.7^{dc}	117 ±6^d	<0.001
Creep (mm) Mean ±SD	0.7 ±0.1	0.7 ±0.2	0.8 ±0.1	0.8 ±0.2	-	-	-	-	0.192

Key: Same superscripted letters indicate a non-significant difference between flexural strength values among different groups when compared by Tukey multiple comparisons post hoc analysis ($p>0.05$).

The 1-way ANOVA showed a statistically significant difference ($p<0.001$) between the flexural strength values of the groups exposed to the fatigue loading cycles (PFFS) and those of the groups that were not exposed to the same cycles. The post hoc Tukey's HSD test showed that all the uncycled specimens had significantly higher flexural strength values than those exposed to cyclic loading before static testing ($p<0.001$) (Table 5). The 2-way ANOVA showed that both cyclic loading and fiber position affected the flexural strength significantly ($p<0.05$). However, the interaction between the 2 factors was not significant ($p=0.467$).

A difference in the fracture mode between groups was detected by visual examination. In the 4L-N and 4L-A+4L-N subgroups, all the specimens were partially fractured and the fracture path was arrested at the fiber layers placed around the metal housing (Table 6). The rest of the specimens were broken completely into 2 pieces (Figure 18 and 19).

Table 6. Fracture mode of test specimens for the investigated groups. Modified from original publication II.

Group	**Subgroup**	**Fracture behaviour**			
		Fracture arrested at fibers		**Complete fracture**	
		Post-fatigue static loading	**Cyclic loading**	**Post-fatigue static loading**	**Cyclic loading**
Control	--	--	--	6/12	6/12
4L (4 layers of SN fibers)	4L-A	--	--	6/12	6/12
	4L-N	6/12	6/12	--	--
	4L-A+4L-N	6/12	6/12	--	--

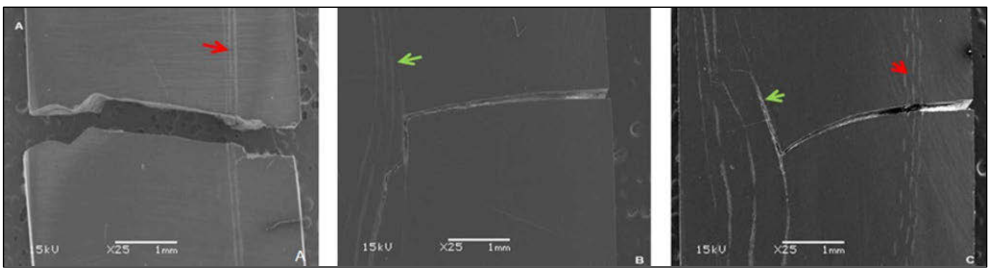

Figure 18. SEM micrograph of the fracture path for the 4L group demonstrating the fiber position (red and green arrows) within the specimen. A, Subgroup 4L-A with SN fiber above the metal housing (red arrow), B. subgroup 4L-N with SN fiber adjacent to the metal housing (green arrow), and C. Subgroup 4L-A+4L-N with SN fiber above the metal housing (red arrow) and SN fiber adjacent to the metal housing (green arrow). Adopted from original publication II.

Figure 19. Fractured specimen of the control group (top view). Adopted from original publication II.

6.3 Flexural strength and flexural modulus of fiber-reinforced, resilient liner-retained overdenture (study III)

The 3-way ANOVA revealed that the matrix material, water storage duration, and reinforcement had a significant effect on the flexural strength and flexural modulus values of the test groups ($p<0.05$). The interaction between the matrix material and storage was significant ($p=0.018$, $p=0.024$) while that between storage duration and reinforcement was not significant ($p=0.236$, $p=0.053$).

The flexural strength and flexural modulus values were significantly different among the tested groups ($p<0.001$) (Table 7). After 24 hours of water storage, the flexural strength and flexural modulus values of groups 1, 3, and 4 were not significantly different from each other ($p=0.788$, $p=0.312$) while they were significantly higher than group 2 ($p<0.05$) (Tukey's post hoc analysis).

After 30 days of water storage, the flexural strength of group 3 was significantly higher than group 1 and group 2 ($p<0.001$), while a non-significant difference was detected among their flexural modulus values ($p=0.065$). In addition, the flexural strength of group 4 was significantly higher than that of group 2 ($p=0.003$). Water storage for 30 days caused a significant reduction in the flexural strength and modulus values of group 1 only ($p<0.001$) while the other 3 groups were not significantly affected ($p>0.05$).

Table 7. Mean flexural strength and flexural modulus values of the tested groups. Adopted from original publication III.

Storage condition	Group	FS (MPa) Mean ±SD	FM (GPa) Mean ±SD
After water storage for 1 day	Group 1 (overdenture specimens with metal matrix)	100.8 ±16.7[a]	2.87 ±0.32[ad]
	Group 2 (overdenture specimens with silicone soft liner matrix)	75.4 ±10.1[bd]	2.16 ±0.22[b]
	Group 3 (overdenture specimens with one bundle of unidirectional Stick glass fibers above the silicone soft liner matrix)	98.7 ±12.3[ac]	3.13 ±0.59[acd]
	Group 4 (overdenture specimens with 4 layers of bidirectional SN glass fiber weaves above the silicone soft liner matrix)	93.5 ±10.1[ace]	2.69 ±0.25[a]
After water storage for 30 days	Group 1 (overdenture specimens with metal matrix)	70.5 ±6.5[bd]	2.51 ±0.28[bcd]
	Group 2 (overdenture specimens with silicone soft liner matrix)	62.2 ±9[b]	2.29 ±0.40[bc]
	Group 3 (overdenture specimens with one bundle of unidirectional Stick glass fibers above the silicone soft liner matrix)	92.1 ±6.8[ace]	2.76 ±0.29[acd]
	Group 4 (overdenture specimens with 4 layers of bidirectional SN glass fiber weaves above the silicone soft liner matrix)	81.3 ±9.7[de]	2.67 ±0.23[acd]
One-way ANOVA (p value)		**< 0.001**	**< 0.001**

Key: Same superscripted letters indicate a non-significant difference among different groups when compared by Tukey multiple comparisons post hoc analysis (P>0.05).

All specimens of groups 1, 2, and 4 fractured completely (Table 8) while only 3 specimens out of twenty from group 3 were partially fractured since the fracture path was arrested near the fibers (Figure 20). Fractured surfaces of groups 3 and 4 are shown in Figures 21 and 22. The bonding interphase between the soft liner and the denture base resin showed good adaptation of soft liner to the resin surface without separation (Figure 23).

Table 8. Fracture mode of test specimens for the investigated groups. Modified from original publication III.

Group	Fracture arrested at fibers (incomplete/partial fracture)		Complete fracture	
	1 day water storage	30 days water storage	1-day water storage	30 days water storage
1	--	--	10/10	10/10
2	--	--	10/10	10/10
3	1/10	2/10	9/10	8/10
4	--	--	10/10	10/10

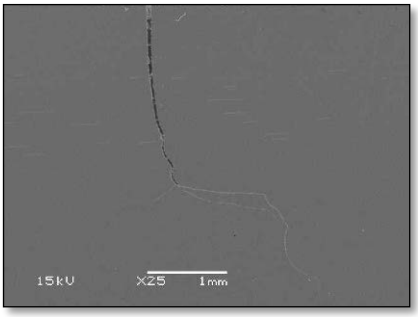

Figure 20. Scanning electron photomicrographs of the fracture line for an incompletely fractured test specimen from group 3. Adopted from original publication III.

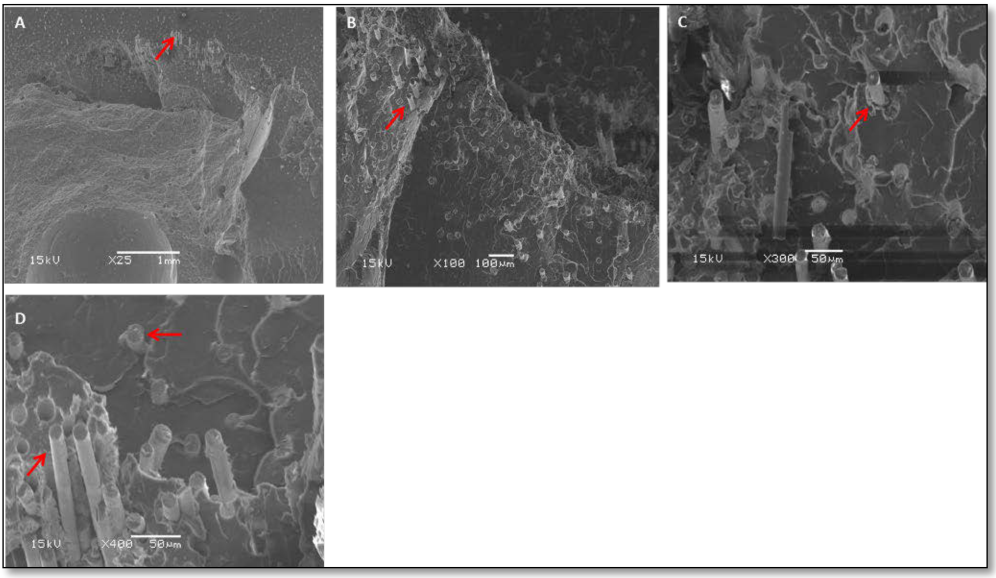

Figure 21. Scanning electron photomicrographs of the fractured surface of group 3 (A: x 25, B: x100, C: x 300, D: x 400) (red arrows represent fibers). Adopted from original publication III.

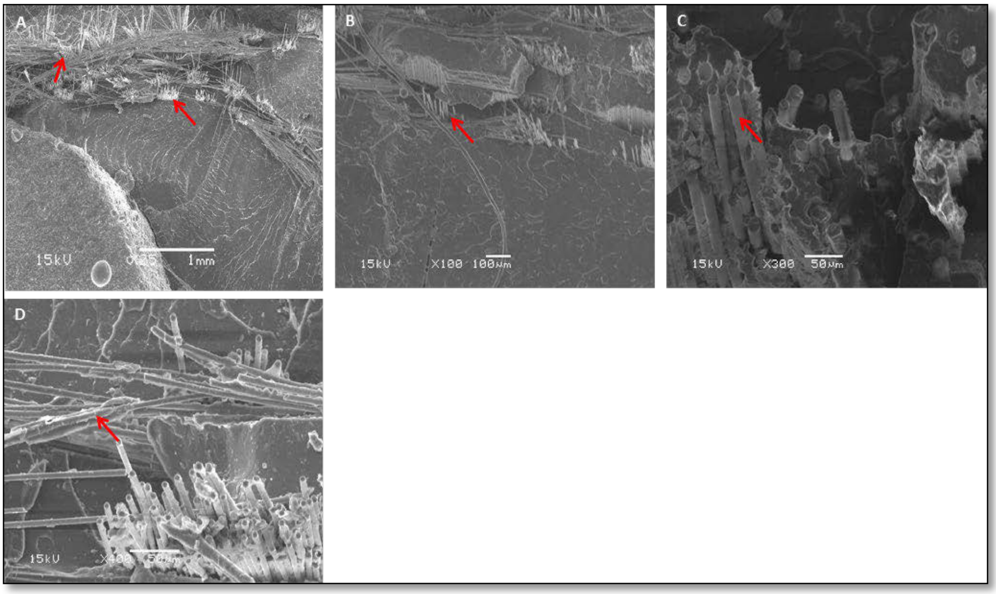

Figure 22. Scanning electron photomicrographs of the fractured surface of group 4 (A: x 25, B: x100, 300, D: x 400) (red arrows represent fibers). Adopted from original publication III.

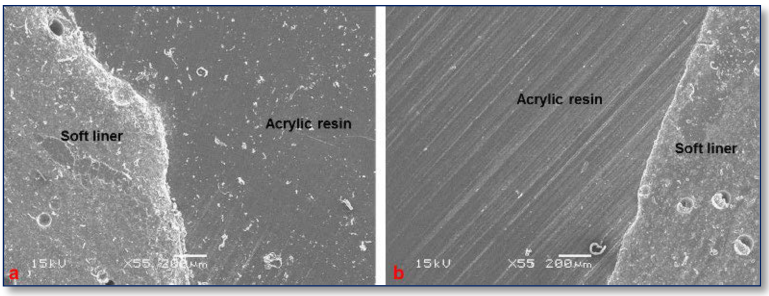

Figure 23. Scanning electron photomicrographs of bonding surface between acrylic resin and soft liner. a: after 1 day of water storage; b: after 30 days of water storage. Adopted from original publication III.

6.4 Midline denture base strains of glass fiber-reinforced single implant-supported overdentures (study IV)

The type of acrylic resin had a non-significant effect on the mean strain values among groups ($p=0.035$) while the type of reinforcement significantly affected them ($p<0.001$) as revealed by the 2-way ANOVA. The interaction between the 2 variables was statistically non-significant ($p=0.552$). For all tested groups, strain values recorded by Ch1 were tensile, while those recorded by Ch2 were compressive.

The tested groups showed a statistically significant difference in the mean strain values of Ch1 (εa) and Ch2 (εb) ($p<0.001$) (Table 9). Both groups APF and HPF had significantly lower strain values than groups AP and HP, by nearly 50% in both channels. ($p<0.05$). The strain values recorded in Ch1 and Ch2 for groups AP and HP were not statistically significant between the 2 groups (p= 0.365 and p=0.988, respectively). The difference in deflection mean values between the tested groups was non-significant (p=0.491) (Figure 24).

Table 9. Mean microstrain values of the tested groups. Modified from original publication IV.

Group	Microstrains Mean ±SD	
	Ch1 (ε_b)	Ch2 (ε_a)
AP (autopolymerizing acrylic resin without fibers)	523 ±85^a	-219 ±65^a
APF (autopolymerizing acrylic resin and unidirectional E-glass fiber reinforcement)	262 ±93^b	-129 ±34^b
HP (heat-polymerized acrylic resin without fibers)	545 ±51^a	-269 ±69^a
HPF (heat-polymerized acrylic resin and unidirectional E-glass fiber reinforcement)	246 ±99^b	-122 ±33^b
One-way ANOVA (p value)	<0.001	<0.001

Key: Same superscripted letters indicate a non-significant difference among different groups when compared by Tukey multiple comparisons post hoc analysis (p>0.05).

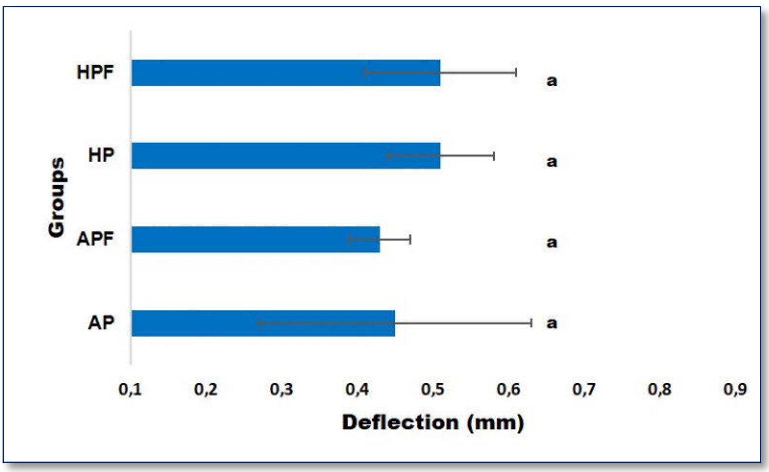

Figure 24. Deflection mean values of test groups. (Group AP=0.45 ±0.18, Group APF= 0.43 ±0.04, Group HP=0.51 ±0.07, Group HPF=0.51 ±0.1 mm). Modified from original publication IV. Same superscripted letters indicate a non-significant difference among different groups when compared by Tukey multiple comparisons post hoc analysis (p>0.05).

7 Discussion

7.1 Discussion of the studies I-IV

This series of studies was performed to investigate the effect of fiber reinforcement on the mechanical properties and strains of implant overdentures. The mechanical studies aimed to determine the suitable number and position of glass fiber layers for efficient overdenture reinforcement, as well as to investigate the behaviour of fiber-reinforced overdentures under static and dynamic loading conditions. The study of denture base strains aimed to evaluate the efficiency of glass fiber reinforcement in reducing denture base strains (deformation) and enhancing its rigidity.

Excessive denture base deformation can lead to denture base fracture complications. In addition, this deformation can transmit compressive stress to the bone causing residual ridge resorption, recurrent ulcer formation, and/or implant overloading (Kordatzis et al., 2003; Takahashi et al., 2018a). Excessive forces transmitted to the implant through the denture base during functional loading can cause a variety of complications such as attachment fracture and implant dislodgement or failure (Takahashi et al., 2015). Approximately 30% to 40% of the overall implant-supported overdenture loads were found to be supported by implants (Ando et al., 2011). Therefore, overdenture base reinforcement might be necessary not only to minimize denture base deformation but also to protect the supporting structures (implants and residual bone) from excessive harmful stress (Takahashi et al., 2015).

Two types of well-documented and frequently used attachments in clinical practice were used. The locator stud attachments were selected for studies I and II. Ball stud attachments with a metal cap or silicone resilient liner matrix were used for studies III and IV. When compared to locator attachments, ball attachments for implant-retained overdentures were linked with substantial mandibular denture base displacement over the implants (Elsyad et al., 2016). As a result, fractures are more likely to occur with ball attachment and denture base strengthening may be recommended to improve the base's fracture resistance. The type of strain developed within the implant overdenture base and its location can differ according to the attachment type and resiliency (Elsyad et al., 2016; Takahashi et al., 2018a), attachment height (Dong et al., 2006), and the number of supporting implants

(Takahashi et al., 2018a). Attachment resiliency can minimize implant loading (Takahashi et al., 2018b; Tanino et al., 2007). However, it can affect denture base deformation and longevity too, which is important for long-term stability (Takahashi et al., 2018a).

Stick Net fibers (woven bidirectional) were used in studies I, II, and III, while Stick (unidirectional) fibers were used in studies III and IV. Both are silane-treated and preimpregnated with porous PMMA. Therefore, they can bond with the resin matrix and provide strong adhesion to it, allowing for better load transfer from the weak polymer matrix to the strong fibers (Narva et al., 2001; Vallittu, 1999). They have different orientations and Krenchel's factor.

The hypothesis of study I was accepted since the number and position of SN fiber layers demonstrated a significant effect on the flexural strength of locator-retained overdenture. Study II confirmed that the fiber positioning significantly affected the CFL and PFFS of locator-retained overdenture. Moreover, it showed that cyclic loading had a significant effect on flexural strength. Those findings were in agreement with previous studies which reported that applying glass fiber reinforcement can improve the mechanical properties of acrylic resin denture bases (Narva et al., 2001; Narva et al., 2005a, 2005b). The fracture load values increased significantly when E-glass mesh fiber reinforcement was added above the abutments of a simulated implant-supported overdenture (Fajardo et al., 2011).

Study III compared the flexural strength and flexural modulus of soft liner-retained overdentures to ball and socket-retained overdentures. In the same study, the effect of using glass fiber as a reinforcement material for soft liner-retained overdentures on these mechanical properties was evaluated. In study IV, the effect of unidirectional glass fiber reinforcement on the midline strains of a mandibular single implant-supported overdenture was evaluated using biaxial strain gauges. The gauges were fixed at the mid-line of the overdenture base above the implant abutment. This area was reported to be the most common site of fracture incidence (de Paula et al., 2020; Gonda et al., 2010).

In studies I, II, and III, autopolymerizing acrylic resin was used instead of heat-polymerizing resin to avoid fiber displacement during compression molding procedures. In study IV, overdentures were fabricated from both autopolymerizing and heat-cured acrylic resin.

In study I, using 2 or 4 layers of bidirectional E-glass fiber weaves (SN) reinforcement caused an increase in the values of flexural strength. However, the values for the reinforced groups were not much higher than those of the control. Using 2 layers of SN fibers did not make a significant improvement on the flexural properties of locator-retained overdenture. This could be due to the low volume fraction of glass fibers in the high stress-bearing area of the specimen (Vallittu et al., 1994b). Another reason might be the insufficient bonding of the fibers or polymer matrix to the coupling agent (Solnit, 1991).

The results of the 4 included studies were in agreement with each other. The flexural strength (study I), CFL (study II), and PFFS (study II) of locator-retained overdenture increased significantly when 4 layers of SN fibers were placed either above the attachment matrix, or a combination of above and next to it. Study III found that the flexural strength and flexural modulus values increased significantly by placing Stick and SN fibers above the silicone-based liner after 1 and 30 days of water storage. In study IV, placing unidirectional Stick fibers above the residual ridge and the ball attachment resulted in a significant reduction in both compressive and tensile denture base strains up to almost 50%. This indicated that such reinforcement improved the denture base rigidity and distributed the stress by several thousand glass fibers to a wider area (Narva et al., 2001, 2005a). A sufficient amount of glass fiber reinforcements placed accurately within thin areas or areas of tensile stress within the denture bases can have a positive impact on their strain at fracture and improve toughness (Agha et al., 2016; Narva et al., 2005b; Vallittu, 1999). The highest stress concentration was detected in areas surrounding the attachment of a single-implant-retained mandibular overdenture model (Amaral et al., 2018). An implant overdenture with metal reinforcement above the copings recorded less strains within the denture base (Takahashi et al., 2013). Fracture load values (Fajardo et al., 2011) beside the static and dynamic strength (Rached et al., 2011) of simulated implant-supported overdenture increased with the addition of glass fibers above the abutments. A finite element analysis study (Berger et al., 2019) concluded that unidirectional glass fiber reinforcement placed in the middle region and over the top of the implants provided better load distribution within the denture base.

On the other side, placing the SN fiber reinforcement around the attachment housing significantly increased the CFL (study II) but did not cause a significant increase in the FS (study I) and PFFS (study II) values. This was in agreement with a previous study (Takahashi et al., 2015) where cast reinforcement placed on the sides of implant coping was not sufficient to minimize denture base strains and deformation significantly. Accordingly, since it reduced the deformation to a certain limit, the authors mentioned that it may be suitable in situations where the space between the abutment and denture teeth is lacking. Similarly, Rached et al. (2011) found that the strength potential of fibers placed on the compression side of an implant-supported overdenture simulating model was less than that of fibers placed at the middle section of the specimen. This can be explained by the effect of fiber layers positioning within the specimens during the onset of crack initiation and propagation (Agha et al., 2016; Narva et al., 2005b). Tensile stress enhances the onset of a crack. Placing the fibers closer to the tensile stress side makes them stretch and absorb more energy before fracture. So, when the fibers are positioned above the attachment matrix, they become more effective since they delay the onset of crack initiation and maximizing the force needed for crack growth. However, when they are located only around the housing near the side of compressive stress, the neutral axis is shifted closer to the fiber layer. As a result, the homogenous polymers are exposed to higher tensile stress and the test specimen will fracture at a lower load value (Figure 25)

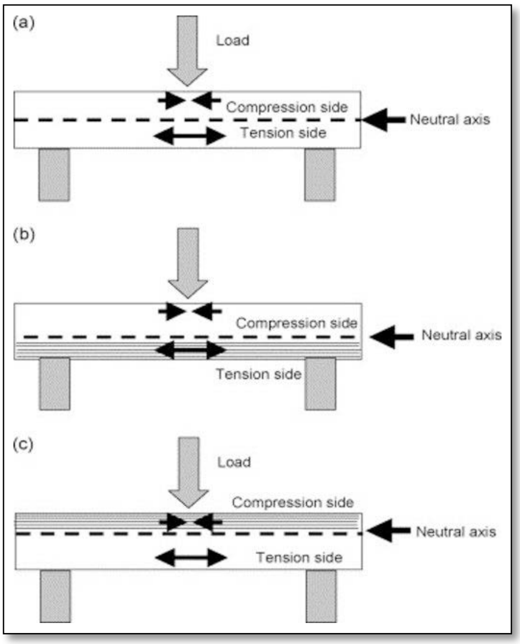

Figure 25. Schematic representation highlighting the areas of tensile and compression side. If specimen is homogenous without any reinforcement, neutral axis located in the middle of test specimen (a), but if the fiber reinforcement is placed in tension side of the test specimen (b), or in the compression side (c), neutral axis is moved towards fiber reinforcement layer (Narva et al., 2005b).

The non-significant increase in flexural modulus and strain values (study I) of the reinforced specimens could be due to the low fiber volume. Also, in study II, creep values of the reinforced groups did not show a significant reduction. A previous study (Dyer et al., 2005) reported an increase in the flexural modulus values when the fiber quantity increased. The used quantity of fiber may not be enough to make a significant increase in the fracture modulus (Y), which is directly proportional to the fiber concentration.

In study II, the fatigue test was performed at a low frequency in a water bath at 37ºC to avoid the effect of heat generation that may affect the results (Narva et al., 2005a). Since the type of load can affect the failure behaviour and mode (Coelho et al., 2016), the reinforcing effect of fibers was more notable in the case of dynamic load application as seen in study II.

The incomplete fracture of specimens with reinforcement placed next to the metal housing (studies I, II) may be due to the fiber weaves that engage the undercuts of the housing with some friction and at the same time bond chemically with the acrylic resin. Therefore, they may enhance the bond between the metal and the denture resin material. Another explanation might be that the fibers were able to absorb the low crack energy and stop it. In study III, the difference in fiber content between SN and Stick fiber reinforcement may explain the incomplete fracture of some Stick fiber-reinforced specimens. The Stick reinforcement has a thickness of 0.9 mm while the thickness of woven SN reinforcement is 0.06 mm (Vallittu, 1999).

A silicone-based soft lining material was selected as a retainer matrix material for ball attachment in study III since they are more resistant to aging when compared with acrylic resin-based ones. Moreover, they are more retentive, durable, and respond to load application and removal very quickly (Abe et al., 2009; Jepson et al., 1993; Murata et al., 2008).

In study III, water storage did not have a significant effect on the mechanical properties of glass fiber-reinforced overdenture specimens. This agrees well with a previous study (Yoshida et al., 2016) where the flexural strength of reinforced denture base resins was not affected after 180 days of water immersion in contrast to the bulk denture base resin. That was explained by the leaching of soluble components such as unreacted monomers and plasticizers that causes the formation of micro voids filled with water by inward diffusion (Arima et al., 1995; Vallittu et al., 1995). The voids can facilitate the movement of polymer chains resulting in a reduction in polymer strength (Takahashi et al., 1999). On the other side, the incorporation of glass fiber can reduce water sorption (Cal et al., 2000).

The material of the matrix significantly affected the tested mechanical properties in study III as shown by the 3-way ANOVA. This would be due to the difference in the elastic modulus between metal and silicone resilient liner which can alter stress distribution (Abe et al., 2009; Elsyad, 2012; Tanino et al., 2007). Another cause might be the diameter of the silicone soft liner matrix, which was 2.4 mm bigger than that of the metal matrix. However, after 30 days of water storage, the flexural strength was not significant between metal and silicone matrices. The liner matrix was designed to be 1.88 mm thick bilaterally on both sides of the ball abutment based on a previous study (Kanazawa et al., 2007) which concluded that a bilaterally 2 mm thick layer of a silicone-based soft liner with hardness less than 90 was the effective liner thickness in distributing and reducing stress transmission from the denture base to the implant supporting structures.

In study III, only the mechanical properties of overdentures with metal matrix were significantly affected by the 1-month water storage. Materials with different resiliencies can affect the load transmission and stress distribution (Tanino et al., 2007). In addition, the polymerization shrinkage of acrylic resin at the resin-metal interface tends to generate residual stresses in this area causing the resin to pull away from the rigid metal and increasing the risk of bond failure. Furthermore, it enables the influx of oral fluids which causes material degradation (Ikeda et al., 2006). While silicone soft liners are viscoelastic materials, behave as a cushion, and can compensate for the volumetric contraction. Additionally, they maintain adequate bonding and contact with denture base resin (Hashem, 2015) thanks to their viscoelastic properties which can remain unaffected even after 3 years of water storage (Murat et al., 2000). Also, it might be the bond strength between the tested matrix and the denture base (Domingo et al., 2013; Ozkir and Yilmaz, 2017). For the metal matrix, bonding was mechanical while in the case of silicone, the applied primer created a chemical bond with the denture base (Rodrigues et al., 2013). The high filler content (37%) of the used GC reline soft liner (McCabe, 1998) can reduce its water sorption and solubility (Abe et al., 2009). This was supported by the SEM evaluation (Figure 23). Water storage seems to be not affecting the bond strength of silicone-based soft liners to the denture base resin (El-Hadary and Drummond, 2000).

In study IV, deflection values were not significantly different among the evaluated groups. A previous study (Kostoulas et al., 2008) showed that the use of fiber reinforcement did not have a significant effect on repaired denture base resin deflection.

7.2 Clinical implications and future investigations

The findings of the studies demonstrated that preimpregnated glass fibers are suitable for reinforcing implant overdentures and can reduce the incidence of fracture complications. The necessity of proper positioning of fiber reinforcement in enhancing the flexural properties of IODs was emphasized in these investigations. Placing a suitable amount of such reinforcement in the position of high stress is effective. This position was found to be in the area above the attachment. The mechanical properties, namely, the flexural strength, cyclic fatigue limits, and post fatigue flexural strength of implant-retained denture base increased with glass fiber reinforcement. In addition, the denture base strains and deformation were reduced by 50% when adding unidirectional glass fibers above the attachment of SIO, meaning that that kind of reinforcement can enhance the overdenture base rigidity and distribute stress evenly rather than its concentration within the weak areas around the attachment.

Subtractive milling and additive manufacturing using 3-dimensional (3D) printing are two digital manufacturing technologies that have emerged as viable alternatives to the traditional method of denture base production. However, the mechanical performance of the 3D printed items is inferior to that of parts manufactured using traditional polymer processing technologies such as compression moulding. The addition of fibres to a polymer matrix to produce a composite can improve the structural strength of printed polymer objects significantly. Short fibers can be added to resin filaments. Also, in the literature, continuous fibre printing has been achieved using either "In-situ fusion" or "ex-situ prepreg". However, some limitations, such as weak interfacial and interlayer bonding at the fiber-polymer interface, still require more study and development. On the other hand, pre-polymerized PMMA-based discs designed for fabricating milled denture bases with the aid of computer-aided design/computer-aided manufacturing (CAD/CAM) technologies are equivalent or superior to conventional resins in terms of many properties. Due to their improved strength, CAD/CAM milled fiber-reinforced composites have been employed in the manufacture of telescopic crowns and indirect composite restorations. It would be of interest to investigate their use in the fabrication of IODs.

Clinical studies investigating the effect of glass fiber reinforcement on the prosthetic complications and deformation of implant-retained overdentures are recommended to confirm the findings of these studies.

8 Conclusions

This series of studies aimed to investigate the effect of fiber reinforcement on the mechanical properties and stress distribution of implant overdentures. The main findings and conclusion are:

1. Placing 4 layers of bidirectional woven preimpregnated E-glass fiber reinforcement above the attachment’s metal housing can improve the load bearing capacity of locator-retained overdentures.
2. The application of 4 layers of bidirectional woven preimpregnated E-glass fiber reinforcement above the attachment’s metal housing can increase the fatigue resistance of locator-retained overdentures. Bidirectional woven E-glass fiber placed adjacent to/around the attachment's housing does not significantly strengthen locator-retained overdentures before or after cyclic loading. Furthermore, the allocation of fiber reinforcement adjacent to the metal housing has no significant effect on the mechanical properties of overdenture specimens already reinforced with fibers above it.
3. Flexural strength and modulus of a ball-retained overdenture with a metal matrix are not significantly different from those with a silicone resilient liner matrix after water storage for 1 month. In addition, 1 month of water storage has no impact on the flexural strength and modulus of an overdenture with a silicone resilient liner matrix and fiber reinforcement. Placing polymer-preimpregnated unidirectional and bidirectional E-glass fiber reinforcement above the resilient liner matrix can improve the mechanical properties of soft liner-retained overdentures.
4. Adding polymer-preimpregnated unidirectional E-glass fiber reinforcement over the residual ridge and implant attachment can minimize the mid-line denture base strains and deformation of an SIO by almost 50%.

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Turku, 8 July, 2021
Mona Fathy Gibreel
Mona Gibreel

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