FAHAD BAKITIAN

MONOLITHIC AND SEMI-MONOLITHIC TRANSLUCENT ZIRCONIUM-DIOXIDE RESTORATIONS

Aspects on design, material and strength

**MONOLITHIC AND SEMI-MONOLITHIC
TRANSLUCENT ZIRCONIUM-DIOXIDE
RESTORATIONS**

Malmö University
Faculty of Odontology Doctoral Dissertation 2020

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Malmö University, 2020
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Department of Materials Science & Technology
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To my first source of inspiration in life, my dear mother,
who passed away before the completion of my thesis

TABLE OF CONTENTS

THESIS AT A GLANCE	9
ABSTRACT	10
POPULÄRTVETENSKAPLIG SAMMANFATTNING	14
LIST OF PAPERS	18
Contribution by the respondent	19
ABBREVIATIONS AND NOMENCLATURE	20
INTRODUCTION	22
Background	22
Ceramics in dentistry	24
Ceramic materials for bilayered design	24
Ceramic materials for monolithic design	31
Potential deleterious effects on zirconia materials	34
Low temperature degradation concerns	34
Manufactural, clinical, and technical effects	35
Design challenges and clinical performance of zirconia	37
Bilayered and monolithic designs limitations	37
Connector design	40
Assessment of load-bearing capacity and clinical prediction for all-ceramic restorations	41
The need for continuous design improvement	42
AIMS	44
Specific Aims	44
MATERIALS AND METHODS	45
Laboratory procedures in study I, II, and III	45

Part 1: specimen preparation	45
Part 2: aging procedures and cementation	55
Part 3: load to fracture test and fracture mode analysis	58
Laboratory procedures in study IV	59
Part 1: creating 3D solid models.....	59
Part 2: generating and processing of FEA models	61
Statistical analysis	65
RESULTS.....	66
Study I	66
Load at fracture	66
Fracture mode	66
Study II.....	68
Load at fracture	68
Fracture mode	69
Study III.....	70
Load at fracture	70
Fracture mode	72
Study IV	75
DISCUSSION	80
The main findings of this thesis	80
Analysis of the load-bearing capacity	80
Analysis of fracture mode	88
Aspects of methodology in this thesis	89
Load to fracture test and artificial aging	89
Load to fracture test and tested specimens	91
Finite element method (FEM)	92
Methodological limitations	93
Clinical significance	94
Future perspectives.....	95
CONCLUSIONS	97
ACKNOLEDGMENTS.....	99
REFERENCES	102
PAPERS I-IV	115

THESIS AT A GLANCE

Study	Aim	Illustration	Key findings
(I) Fracture strength of veneered translucent zirconium dioxide crowns with different porcelain thicknesses	To evaluate fracture strength of veneered translucent zirconium dioxide crowns designed with different porcelain layer thicknesses		Translucent zirconia crowns can be veneered with minimal thickness layer of 0.5mm porcelain without showing significantly reduced fracture strength compared to traditionally veneered crowns. Fracture strength of micro-veneered crowns with a layer of porcelain (0.3 mm) is lower than that of traditionally veneered crowns but still within range of what may be considered clinically sufficient. Porcelain layers of 2.0mm or thicker should be used where expected loads are low only
(II) Effect of different semimonolithic designs on fracture resistance and fracture mode of translucent and high-translucent zirconia crowns	To describe different designs of semi-monolithic crowns made of translucent and high-translucent zirconia materials and to evaluate the effect on fracture resistance and fracture mode		Translucent and high-translucent zirconia crowns might be used in combination with 0.3 mm microcoating porcelain layer with semi-monolithic design to enhance the esthetic properties of restorations without significantly decreasing fracture resistance of crowns. If 0.5 mm porcelain layer is needed for semi-monolithic crown, wave design or cap design might be used to increase fracture resistance
(III) Load-bearing capacity of monolithic zirconia fixed dental prostheses fabricated with different connector designs and embrasure shaping methods	To investigate load-bearing capacity and failure mode of monolithic zirconia fixed dental prostheses (FDPs) with different connector designs gained by using different embrasure shaping methods		Sharp embrasures and interproximal separations made with diamond discs significantly decrease the load-bearing capacity of monolithic zirconia FDPs compared to FDPs made with blunt embrasures. Blunt embrasures in combination with localized porcelain build-up produce FDPs with high load-bearing capacity in relation to loads that might be expected under clinical use
(IV) Evaluation of stress distribution in tooth-supported fixed dental prostheses made of translucent zirconia with variations in framework designs: A three-dimensional finite element analysis (3D-FEA)	To evaluate influence of framework designs on the stress distribution within tooth-supported partially veneered FDPs made of translucent zirconia under simulated loads using 3D-FEA		Framework and veneer designs play a significant role in stress distribution of partially veneered zirconia FDPs under loading. The FDPs with zirconia frameworks with cap design minimize maximum principal tensile stress in veneering porcelain. The FDPs with 0.3mm-veneering porcelain show low maximum principal tensile stress in veneering porcelain, but highest maximum shear stress at zirconia-veneer interface. The FDPs with wave design of zirconia frameworks minimize maximum shear stress considerably

ABSTRACT

Several clinical reports have shown high rates of clinical success associated with fixed dental restorations made of traditional zirconium-dioxide (zirconia). Clinical complications related to the design of the restorations do however occur as some aspects of design for optimal aesthetics are in conflict with design for optimal strength. Previous clinical and laboratory studies indicated that veneering porcelain and the connector represent the weak parts of the fully-veneered zirconia restorations where failure may occur. Although the clinical performance of such restorations has recently been reported to be comparable to metal-ceramic restorations, further improvements in design are still required. Restorations with monolithic design made of modified translucent zirconia materials offer an excellent solution to these clinical problems. One of the advantages of monolithic restoration is that such restorations can be prepared without the weak veneering material. Thus, this restoration design has a much higher load-bearing capacity compared to the veneered restorations since it provides additional space for the high-strength zirconia material.

With regard to strength and aesthetics, the translucency of the former generation of monolithic translucent zirconia, which comes with equivalent mechanical properties to traditional zirconia, is insufficient. Recently, monolithic zirconia with high translucent properties was developed for highly aesthetic clinical uses. These new systems of translucent zirconia materials have limited capacity in terms of fracture strength and fracture toughness properties. Further, earlier studies have shown doubtful aging stability for these materials. Maintaining well-known strength properties of

zirconia restorations while providing a high aesthetic outcome is the ultimate goal for dental restorations such as single crowns (SCs) and fixed dental prostheses (FDPs). The optimum design for restorations made of the former generation of translucent zirconia could help prevent the risks associated with bilayered restorations and overcome the limitations of high-translucent monolithic restorations. Based on clinical needs and previous clinical observations, the overall aim of this thesis was to evaluate translucent and high-translucent zirconia restorations and the effect of design modifications, used to enhance the aesthetics, on fracture resistance.

In the first three studies, I, II, and III, all the SCs and the FDPs were artificially aged and loaded to fracture. Fracture mode analysis in the different studies was performed visually and microscopically. In study I, fracture strength and fracture mode of veneered translucent zirconia SCs designed with different porcelain layer thicknesses were evaluated. Study II described different designs of partially veneered monolithic (semi-monolithic) SCs made of translucent zirconia and evaluated the effect of those designs on fracture resistance and fracture mode of crowns made of two generations of translucent zirconia materials. Study III investigated the load-bearing capacity and failure mode of monolithic zirconia FDPs with different connector designs achieved by using different embrasure shaping methods. Finally, study IV aimed to evaluate the influence of the framework designs on the stress distribution within tooth-supported semi-monolithic FDPs made of translucent zirconia material under simulated loads using a three-dimensional finite element analysis (3D-FEA). Simplified 3D solid models of prepared abutment teeth with different 3-unit FDPs based on the designs of studies II and III, were created. The designs of 3-unit FDPs included monolithic zirconia, semi-monolithic zirconia with 0.3 mm veneer thickness, semi-monolithic zirconia with 0.5 mm veneer thickness, semi-monolithic zirconia with 0.5 mm veneer thickness supported with cap design, and semi-monolithic zirconia with 0.5 mm veneer thickness supported with wave design.

The outcomes of study I showed that translucent zirconia SCs can be veneered with a minimal thickness layer of 0.5 mm porcelain without showing significantly reduced fracture strength compared to traditionally veneered (1.0-2.0 mm) SCs. Fracture strength of micro-veneered SCs with a layer of porcelain (0.3 mm) is lower than that of traditionally veneered SCs but still within range of what may be considered clinically sufficient. Porcelain layers of 2.0 mm or thicker should be used only where the expected loads are low. All the SCs in groups 2.5 and 2.0 and more than 80% of the SCs in groups 1.0, 0.8 and 0.5 showed cohesive fracture mode. Conversely, there were significantly ($p \leq 0.05$) more complete fractures in group 0.3 compared to all other groups. The results of study II demonstrated that translucent and high-translucent zirconia SCs might be used in combination with a 0.3 mm micro-coating porcelain layer with semi-monolithic design to enhance the aesthetic properties of restorations without significantly decreasing fracture resistance of the SCs. If a 0.5 mm porcelain layer is needed for semi-monolithic SCs, wave design or cap design might be used to increase fracture resistance. The SCs made of translucent zirconia showed higher load at fracture compared to those made of high-translucent zirconia, regardless of design. All monolithic SCs, semi-monolithic SCs with 0.3 mm buccal veneer (100%), and all but one of semi-monolithic SCs with cap design (95%) showed complete fractures. Semi-monolithic SCs with wave design and semi-monolithic SCs with a 0.5 mm buccal veneer showed (70% and 55%, respectively) cohesive veneer fractures. The results of study III showed that sharp embrasures and interproximal separations made with diamond discs significantly decrease the load-bearing capacity of monolithic zirconia FDPs compared to monolithic FDPs made with blunt embrasures ($p < 0.001$). Blunt embrasures in combination with localized porcelain build-up produce monolithic FDPs with high load-bearing capacity in relation to loads that might be expected under clinical use. Fracture mode of the FDPs fabricated with sharp embrasures and interproximal disc separations differed significantly compared to the FDPs with no occlusal embrasures, the FDPs with blunt embrasures, and the FDPs with interproximal

porcelain separations ($p < 0.001$). The findings of study IV confirmed that framework and veneer designs play a significant role in the stress distribution of the partially veneered zirconia FDPs under loading. The FDPs with zirconia frameworks with cap design minimize the maximum principal tensile stress in the veneering porcelain. The FDPs with 0.3 mm-veneering porcelain show low maximum principal tensile stress in the veneering porcelain, but the highest maximum shear stress at the zirconia-veneer interface. The FDPs with wave design of zirconia frameworks minimize the maximum shear stress considerably.

POPULÄRTVETENSKAPLIG SAMMANFATTNING

Ett flertal kliniska studier har påvisat hög lyckandefrekvens för fasta protetiska konstruktioner gjorda av traditionell zirkoniumdioxid (zirkonia). Trots det uppstår emellanåt kliniska komplikationer relaterade till tandersättningarnas utformning i förhållande till optimal estetik, eftersom estetiken ofta står i konflikt med den utformning som ger bäst hållfasthet. Tidigare genomförda kliniska studier och laboratoriestudier har påvisat att ytporslin som skiktas utanpå en underkonstruktion utgör försvagningar i konstruktionen tillsammans med konnektorerna, som är andra ställen som kan utgöra försvagningar. Även om helkeramiska konstruktioner generellt visar goda kliniska resultat, väl jämförbara med traditionella metallkeramiska konstruktioner, behövs ytterligare förbättringar för att minska risken för komplikationer. Konstruktioner med monolitisk utformning, det vill säga utan ytporslin, framställda av modifierad translucent zirkoniumdioxid erbjuder en bra lösning för dessa kliniska problem. En avgörande fördel med monolitisk utformning är att tandersättningarna framställs utan ytporslin som i sig annars utgör en svag länk i konstruktionen. Genom att utelämna porslin frigörs även mer utrymme för den starkare zirkoniakärnan. Monolitisk utformning förväntas därmed ge högre lastbärande kapacitet och därmed bättre hållfasthet jämfört med konstruktioner som skiktats med ytporslin.

Avseende estetik är translucensen på tidigare generationers translucent zirkoniumdioxid, med motsvarande mekaniska egenskaper som traditionell zirkoniumdioxid, otillräcklig för att kunna efterlikna tandsubstans. Nyligen utvecklades därför högtranslucent

zirkoniumdioxid, med indikationer att användas monolitiskt där de estetiska kraven är högre. Dessa nya keramer har emellertid begränsade egenskaper när det gäller brottseghet och hållfasthet och studier har även påvisat oklarheter avseende negativ materialpåverkan vid åldrande av materialen. Att bibehålla de erkänt goda mekaniska egenskaperna hos zirkonia och samtidigt öka translucensen och därmed uppnå högestetiska resultat med kronor och broar, är det ultimata målet för tandersättningar utförda i translucent zirkonia. Att optimera utformningen av ersättningarna baserade på tidigare generationers starka translucenta zirkoniumdioxid och därmed eliminera de risker som är förknippade med porslinskiktningen och samtidigt övervinna de estetiska begränsningarna hos material som inte är lika translucenta som de högtranslucenta zirkoniamaterialen. Det övergripande syftet med föreliggande avhandling var därför att utvärdera translucenta och högtranslucenta tandersättningar med modifierad utformning baserad på kliniska behov och tidigare observationer, samt att undersöka hur dessa modifieringar påverkar tandersättningarnas mekaniska egenskaper.

I de tre första studierna, I, II, III, åldrades samtliga kronor och broar artificiellt och belastades till fraktur. Frakturanalyser genomfördes visuellt och i mikroskop. I studie I, utvärderades hållfastheten hos porslinsskiktade translucenta zirkoniakronor med olika tjocklek på porslinsskiktet. Studie II beskriver olika utformning av delvis skiktade monolitiska (semi-monolitiska) kronor gjorda av två olika generationers zirkoniumdioxid; translucent och högtranslucent zirkoniumdioxid. Därefter utvärderades effekten av dessa olika konstruktioners hållfasthet. I studie III undersöktes hållfasthet och frakturtyp för monolitiska zirkoniabroar med olika utformning på konnektorerna, erhållna genom användning av olika metoder för utformning av konnektorövergång och separeration mellan de olika leden. Slutligen, i studie IV var syftet att utvärdera hur olika broutformning påverkar spänningsutbredningen på tandstödda semi-monolitiska brokonstruktioner gjorda av translucent zirkonia under simulerade belastningar med hjälp av en tredimensionell finita-elementanalys (3D-FEA). Förenklade solida 3D-modeller skapades av stödtänder preparerade för 3-ledsbro med olika broutformning, enligt studie II och III skapades. Broutform-

ningen inkluderade monolitiska 3-leds broar gjorda av translucent zirkonia, semi-monolitisk translucent zirkonia med 0,3 mm samt 0,5 mm porslinstjocklek buckalt på konstruktionen, semi-monolitiska translucent zirkonia med 0,5 mm porslinstjocklek med stöd för porslinsskiktadfasad genom att utforma konstruktionen med kappa och semi-monolitisk translucent zirkonia med 0,5 mm porslinsfasad med stöd genom en vågformad utformning.

Resultaten av studie I visade att translucenta zirkoniumdioxid-kronor kan skiktas med minimal tjocklek med 0,5 mm porslin utan att uppvisa signifikant reducerad hållfasthet jämfört med traditionellt skiktade (porslinstjocklek 1,0 - 2,0 mm) kronor. Brottsegheten på mikroskiktade kronor, med 0,3 mm i porslinsskikt är lägre än för traditionellt skiktade kronor, men fortfarande inom vad som anses vara kliniskt tillräckligt. Porslinsskikt på 2,0 mm eller tjockare bör undvikas om inte de förväntade krafterna är låga. Alla kronor i grupperna med porslinstjocklek 2,5 och 2,0 mm samt mer än 80% av kronorna i grupperna 1,0 - 0,8 och 0,5 mm uppvisade kohesiva frakturer. I gruppen med 0,3 mm i porslinstjocklek skedde signifikant ($p \leq 0,05$) fler totalfrakturer jämfört med alla andra grupper. Resultaten av studie II visar att semi-monolitiska translucenta eller högtranslucenta kronor i kombination med ett 0,3 mm porslinsskikt kan användas för att förbättra de estetiska egenskaperna på konstruktionen, utan att signifikant minska brottsegheten. Om ett tjockare porslinsskikt behövs vid semi-monolitisk krondesign kan ett fasadurtag med vågform eller med kappa användas för att ge stöd åt porslinet och öka motståndet mot sprickpropagering. Kronor av translucent zirkoniumdioxid uppvisade högre hållfasthet jämfört med kronor av högtranslucent zirkoniumdioxid, oavsett utformning. Alla monolitiska kronor, semi-monolitiska kronor med 0,3 mm buckalt skiktat porslin (100%) och alla utom en av de semi-monolitiska kronorna med kappdesign (95%) visade totala frakturer. Semi-monolitiska kronor med vågdesign och semi-monolitiska kronor med 0,5 mm buckalt skiktat porslin visade (70% respektive 55%) kohesiva fasadfrakturer. Resultaten av studie III visade att skarpa övergångar och interproximala separationer gjorda med diamantttrissor minskar hållfastheten signifikant på monolitiska zirkoniabroar jämfört med monolitiska

broar gjorda med oskarp och mjukare utformning ($p < 0,001$). Mindre skarpa och mer avrundade övergångar i konnektordelen, i kombination med selektiv porslinsiktning ger monolitiska broar med hög hållfasthet jämfört med de belastningar som kan förväntas vid kliniskt bruk. Frakturtyperna för broar tillverkade med skarpa övergångar och interproximala separationer skiljde sig signifikant jämfört med broar utan ocklusala övergångar, broar med avrundad utformning och broar med interproximal separation ($p < 0,001$). Resultaten av studie IV bekräftar att broarnas utformning och ytporslinets utformning spelar en signifikant roll för spänningsutbredningen av de delvis skiktade zirkoniabroarna under belastning. Zirkoniabroarna med fasadurtag med kappdesign minimerar den maximala dragspänningen i det skiktade porslinet/ytporslinet. Broarna med 0,3 mm ytporslin uppvisar låg dragspänning i porslinet, men den högsta skjuvspänningen vid gränssnittet mellan zirkonia och fasadmaterial (ytporslinet). Broarna med vågdesign i fasadurtaget minimerar avsevärt skjuvspänningarna i gränssnittet.

LIST OF PAPERS

The present thesis was based on the following four scientific articles which are referred to in the main text by their Roman numerals I-IV. The published papers (I, II, and III) are reprinted with the permission of the publishers and included at the end of the thesis.

I. Fracture strength of veneered translucent zirconium dioxide crowns with different porcelain thicknesses.

Bakitian F, Seweryniak P, Papia E, Larsson C, Vult von Steyern P. *Acta Biomater Odontol Scand* 2017;3:74-83.

II. Effect of different semimonolithic designs on fracture resistance and fracture mode of translucent and high-translucent zirconia crowns.

Bakitian F, Seweryniak P, Papia E, Larsson C, Vult von Steyern P. *Clin Cosmet Investig Dent* 2018;10:51-60.

III. Load-bearing capacity of monolithic zirconia fixed dental prostheses fabricated with different connector designs and embrasure shaping methods.

Bakitian F, Seweryniak P, Papia E, Larsson C, Vult von Steyern P. *J Prosthodont* 2019;28:64-70.

IV. Evaluation of stress distribution in tooth-supported fixed dental prostheses made of translucent zirconia with variations in framework designs: A three-dimensional finite element analysis.

Bakitian F, Papia E, Larsson C, Vult von Steyern P. *J Prosthodont* (Accepted).

Contribution by the respondent

The respondent performed most of the work from planning, experimental work and the analysis of data. The respondent was also the main contributor to writing the manuscripts.

ABBREVIATIONS AND NOMENCLATURE

Zirconia	Zirconium dioxide
SCs	Single crowns
FDP	Fixed dental prosthesis
Y-TZP	Yttria-stabilized tetragonal zirconia polycrystals
CAD	Computer-aided design
CAM	Computer-aided manufacturing
MPa	Mega Pascal
ZTA	Zirconia toughened alumina
PSZ	Partially stabilized zirconia
CSZ	Cubic stabilized zirconia
m-ZrO₂	Monoclinic zirconium dioxide
t-ZrO₂	Tetragonal zirconium dioxide
c-ZrO₂	Cubic zirconium dioxide
Y₂O₃	Yttria, Yttrium oxide
Ce₂O₃	Ceria, Cerium oxide
MgO	Magnesia, Magnesium oxide
Al₂O₃	Aluminum oxide
ZLS	Zirconia-reinforced lithium silicate
LTD	Low-temperature degradation
(t)	Tetragonal-phase
(m)	Monoclinic-phase
(c)	Cubic phase
HIP	Hot isostatic pressing
CIP	Cold isostatic pressing
FEM	Finite elements method

FEA	Finite elements analysis
3D	Three dimensional
STL	Standard tessellation language
POM-C	Polyoxymethylene-copolymer
POM-C GF 25	Polyoxymethylene-copolymer glass-reinforced
MDP	10-Methacryloyloxydecyl dihydrogen phosphate
SEM	Scanning electron microscopy
rpm	Revolutions per minute
mm	Millimeter
GPa	Giga Pascal
N	Newton
Hz	Hertz
ANOVA	Analysis of variance
SPSS	Statistical Package for the Social Sciences
µm	Micrometer
SD	Standard Deviation
PDL	Periodontal Ligament

INTRODUCTION

Background

Loss of teeth is associated with a variety of complications that affect the oral functions and aesthetics.¹⁻³ Such complications consequently impair patients' self-esteem and social life, thus reducing their quality of life.⁴ Several treatment options for replacement of missing teeth can be indicated to alleviate these complications. Tooth and implant-supported single crowns (SCs) and multi-unit fixed dental prostheses (FDPs) can be successfully used to achieve excellent aesthetic and functional results, as well as improve the quality of life of patients requiring prosthodontic treatment.⁵⁻⁷ Although several steps are required for this kind of treatment with long and sensitive clinical procedures starting from teeth preparation or implant placement preparation, tissue management, impression making, and shade selection, it is still an excellent treatment option.⁵⁻⁷ Furthermore, the majority of patients prefer this treatment option compared to other removable treatment alternatives.^{8,9}

Several ceramic materials used for fixed dental appliances, such as SCs and FDPs, are available on the dental market.^{10,11} Recent developments in ceramic engineering and technology have made ceramic based restorations increasingly popular.¹²⁻¹⁴ Besides, the emergence of high-strength ceramic materials has resulted in a change in preference in the dental community from full metal and metal-ceramic restorations towards all-ceramic ones.¹²⁻¹⁴ Although ceramic materials have shared well-known biocompatibility compared to high-noble metal alloys, the optical and mechanical properties differ considerably among different ceramic materials.^{10,11,14}

Among the different dental ceramic materials available, yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) has been used and further developed since the mid-nineties. The Y-TZP material constitutes a promising high strength ceramic material for many prosthetic treatments.¹⁵⁻¹⁸ A growing interest in Y-TZP restorations has been seen over the last three decades. They are currently used on a routine basis for tooth/implant-supported SCs and FDPs.¹⁵⁻¹⁸ Their remarkable biocompatibility, strength, toughness, and optical properties compared to metal-based restorations as well as the latest advances in computer-aided design and computer-aided machine (CAD/CAM) technology have accelerated their usage.¹⁵⁻¹⁸

Despite the fact that many clinical reports have shown high rates of clinical success associated with Y-TZP restorations, clinical failure, that can be related to insufficient design, still occurs.¹⁹⁻²¹ The incidence of superficial chip-off fractures and connector fractures for porcelain-veneered Y-TZP restorations has been described.¹⁹⁻²¹ Fixed restorations with monolithic design that are made of translucent zirconia materials have been suggested as an alternative that might offer a solution to such problems.²²⁻²⁵ However, since its appearance in dentistry, different types of zirconia have been further developed with different chemical compositions, crystal configurations, mechanical, optical, and aging properties. From a material aspect, translucent zirconia materials are weaker than traditional zirconia. Taking into consideration that there are so many different types of zirconia materials with strength values ranging from 600 to 1400 MPa, the choice of material is extremely important.^{26,27} If the aesthetic aspect is highly required in certain cases, then more translucent zirconia material can be used. But if the clinical case requires FDPs with long spans, for instance, the strength of this translucent material is not sufficient. It is, therefore, likely that the use of a stronger material that comes with less translucent properties is required. The design then plays a significant role and becomes an essential alternative for achieving the required aesthetics using, for instance, partially veneered (semi-monolithic) designs.^{28,29} This design alternative can provide many opportunities for solving some of the clinical problems that are seen with materials such as metal-ceramic and veneered traditional

zirconium dioxide materials. Moreover, the establishment of innovative designs in combination with translucent zirconia may sometimes be a better alternative to the less strong high-translucent and ultrahigh-translucent monolithic zirconia, but still with comparable aesthetics. In this regard, the present thesis is focused on aspects of design modifications for SCs and FDPs made of translucent zirconia materials, stress formation, load-bearing capacity, and aesthetics for application in fixed prosthetic replacements.

Ceramics in dentistry

Several different classifications have been used for dental ceramic materials based on various factors. Chemical composition, microstructure, processing methods used for fabrication, level of strength and toughness, and clinical indications are among other factors outlining how dental ceramics are classified.^{10-14,30} In the present thesis, dental ceramic materials will be classified based on their chemical composition and by which functional design they appear in the clinic. Regarding design, dental ceramics that are used for all types of all-ceramic restorations can currently be found in two common designs: bilayered design made of two different ceramic materials or monolithic design made of a single ceramic material (Figure 1).

Ceramic materials for bilayered design

Demand for highly aesthetic dental restorations and high patient expectations has led to the replacement of full metal restorations with bilayered restorations made of metal alloys veneered with ceramic materials. This combination of materials has commonly been referred to as the gold standard for fixed dental restorations, and has been used widely since the late nineteen-fifties, taking advantage of both the superior mechanical properties of the high-noble metal alloys and the optimum aesthetics of porcelain.¹⁹⁻²¹ The aesthetic problems that follow the use of metals, such as the dark lines at the restoration margins, have been solved using all-ceramic restorations.³¹⁻³³ The bilayered design of all-ceramic restorations consists of two different ceramic materials forming both the veneer and the core/framework of the restoration. The concept of this

design is that high-aesthetic low-strength veneering ceramic material is supported and fused to low-aesthetic high-strength ceramic substructure materials.

Figure 1. Classification of dental ceramic materials based on design and chemical composition.

Veneering ceramic materials can be categorized according to their chemical composition into two main categories: feldspathic porcelain and glass-ceramic materials. Feldspathic porcelain is an amorphous (non-crystalline) aluminosilicate glass material that contains amounts of silica and aluminium as well as potassium and sodium. It is a highly aesthetic material with low fracture strength, ranging from 50 to 120 MPa and low fracture toughness approximately $1 \text{ MPa m}^{1/2}$.³⁴⁻³⁶ Glass-ceramics are produced by reinforcing the glass matrix of the aluminosilicate material with different crystal types such as leucite, lithium disilicate, or fluorapatite. The crystal reinforcement can be achieved by adding these crystals into the glass matrix (dispersion strengthening) or by heat-induced crystallization of glass leading to subsequent crystal growth in a process referred to as 'ceramizing'.³⁷ The reinforcement method aims to increase the low mechanical strength and toughness of glass material while maintaining its optical properties. Leucite-reinforced (IPS Empress, OPC) and lithium disilicate-reinforced (IPS Empress II, IPS Emax CAD and Press) glass ceramics are two common materials in this category. Leucite-reinforced glass-ceramic has a flexural strength ranging from 120 to 180 MPa with fracture toughness not exceeding $1.5 \text{ MPa m}^{1/2}$.^{38,39} Refining crystal size and increasing crystal number in lithium disilicate-reinforced glass ceramic improves the flexural strength, even more, ranging from 360 to 500 MPa with fracture toughness approximately $3 \text{ MPa m}^{1/2}$.⁴⁰ Feldspathic porcelain is mainly used in highly aesthetic clinical cases as veneering material in bilayered restoration. Another clinical use for this material and for leucite-reinforced glass-ceramic material is as monolithic restorations for anterior laminate veneers.⁴¹ Besides its main application as veneering material and monolithic laminate veneer, lithium disilicate ceramics can be successfully used as monolithic crowns in any region of the mouth as well as anterior three-unit bridge up to the second premolar.^{42,43}

High strength core ceramic materials can be categorized into glass-infiltrated alumina or zirconia-based ceramics (often defined as hybrid ceramics) and densely sintered oxide ceramics (polycrystalline ceramics). These materials are traditionally used for cores and frameworks, shaped to support veneering ceramic materials in

the bilayered design. Hybrid ceramics are made by fabricating a porous polycrystalline matrix that is infiltrated with lanthanum aluminosilicate glass. This category involves Vita In-Ceram® materials with flexural strengths ranging from 350 to 650 MPa.^{44,45} The complex and time-consuming manufacturing of the hybrid ceramics together with the sensitivity of the technique have resulted in significant reduction of the use of these materials compared to glass ceramics and oxide ceramics.⁴⁶ Pure polycrystalline oxide ceramics, on the other hand, are single-phase materials that are formed by fusing the crystals together in a sintering process forming a fully dense polycrystalline structure without any glass phase. Those materials have the highest strength properties among the dental ceramic materials. In clinical dentistry, two common high-strength oxide ceramics are available: aluminium oxide with flexural strength reaches to 600 MPa and fracture toughness of 6 MPa m^{1/2}, and zirconium dioxide with flexural strength ranging from 900 to 1400 MPa and fracture toughness of 8 to 10 MPa m^{1/2}.^{15-18,47} Because of its excellent mechanical properties in terms of fracture toughness and strength compared to aluminum oxide and hybrid ceramics, zirconium dioxide has become the dominant sub-structure material used in all-ceramic restorations with bilayered design since the nineties.¹⁵⁻¹⁸

Zirconium dioxide materials

There are different types of zirconium-dioxide (zirconia) materials available for dental applications. Those materials come with different chemical compositions, varying crystal configurations, several processing methods as well as significant differences in their main mechanical and optical properties. Figure 2 shows the successive generations of zirconia materials starting from using the zirconia crystals as reinforcement components in zirconia toughened alumina (ZTA) to partially stabilized zirconia (PSZ) and the traditional (3Y-TZP) until the emergence of different translucent zirconia materials such as cubic stabilized zirconia (CSZ).^{15-18,22-25,48}

Figure 2. Successive generations of zirconia materials developed for dental use.

Pure zirconia is a polymorphic metal oxide having three crystal configurations, exhibiting continuous phase transformations over a range of temperature changes (Figure 3).^{15,18} At room temperature and up to approximately 1170°C, zirconia exists in a monoclinic crystal configuration (m-ZrO₂). Above this temperature, it transforms into a tetragonal crystal configuration (t-ZrO₂) where it stays until it reaches melting temperature (2370°C) and transforms to cubic crystal configuration (c-ZrO₂). The transformation from t-ZrO₂ to m-ZrO₂ during cooling is associated with a 3-5 percent increase in volume, which in turn cause stress formation and initiates crack growth within the material.

Figure 3. Crystallographic phases of zirconia, temperature and hysteresis.

Inhibition of this detrimental microstructural transformation can be achieved by dissolving stabilizing oxides (yttria, ceria, or magnesia) within zirconia material to dope the material to become stable in the tetragonal crystal configuration at room temperature. This method results in metastable tetragonal material with unique property that can induce a remarkable increase in the fracture toughness of the material by preventing crack propagation in a process called “Transformation toughening” (Figure 4).^{49,50}

Traditional 3Y-TZP is the most common zirconia material, which is used as a substitution for the metal substructure in bi-layered dental restorations.^{15-18,51} At microstructure level, it only consists of one phase of tetragonal zirconia polycrystal, which is stabilized with 3 mol% yttria. The 3Y-TZP is considered to be the toughest zirconia material available in dentistry.⁵¹ When stress concentrates at the tip of an existing crack of a tetragonal structure, the crack starts to propagate. As a result, t-ZrO₂ crystals transform to m-ZrO₂ crystals, with a resulting increase in volume and in turn compressive stress in the crack tip area that effectively inhibits further crack propagation (Figure 4).^{49,50}

Figure 4. Transformation toughening mechanism.

ZTA is not a zirconia material itself but a hybrid ceramic consisting of two phases; aluminium-oxide matrix as the main phase that is reinforced with t-ZrO₂ crystals (stabilized with 12 mol% ceria).⁵² There is only one brand of this material on the market (In-Ceram® Zirconia). This material has relatively high toughness properties but low optical properties due to large pore content in the microstructure and high amount of alumina content.^{44,45} PSZ is another zirconia material that is composed of two phases; t-ZrO₂ mixed with c-ZrO₂ as the main phase. The material is partially stabilized with e.g. 10 mol% magnesia or 3 mol% yttria.⁵³ The two

materials (ZTA, PSZ) are not common anymore in everyday clinical use.

All previous traditional zirconia materials (3Y-TZP, ZTA, PSZ) were veneered with highly aesthetic porcelain or glass ceramic materials in the bilayered design because of their inferior optical properties in terms of colour and translucency. New generations of translucent zirconia materials (translucent 3Y-TZP, 4Y-TZP, 5Y-TZP, 6Y-CZP) with enhanced optical properties and with the possibility to be used without the veneer layer will be discussed in the next section.

Ceramic materials for monolithic design

Bilayered ceramic restorations show excellent aesthetic outcomes, but were prone to veneer fractures, so called chip-off fractures.¹⁹⁻²¹ Monolithic ceramic restorations, on the other hand, are not subject to this concern and have several advantages in comparison to bilayered ones. One of the advantages is related to the ability of monolithic reconstruction to restore the oral function and aesthetics with more conservative tooth preparation since the veneering layer is excluded.⁵⁴ The production of monolithic restorations is less time consuming and thus cheaper compared to bilayered ones where the veneer normally requires a skilful technician to build it up. Indications for monolithic restorations are thought to be increasing, and clinicians gradually seem to be substituting bilayered restorations for everyday clinical use with monolithic restorations.

All-ceramic restorations with monolithic design have been used in dentistry for more than 100-years. Already the old “jacket crown” made of feldspathic porcelain had a monolithic design.⁵⁵ In modern times, reinforced glass-ceramic materials, such as lithium-di-silicate, have been used successfully with a monolithic design for laminate veneers, single crowns and small FDPs in the anterior region.^{42,43} Recently, new ceramic material has been developed that is based on zirconia reinforced glass-ceramics; Zirconia-reinforced Lithium Silicate (ZLS). This is a fairly new material containing zirconia as a reinforcement of glass-ceramic matrix. The material is, in general, indicated clinically as lithium disilicate material (single crowns, FDPs up to three units anteriorly). Although ZLS has

slightly higher strength compared to lithium disilicate, this material comes with lower fracture toughness, and consequently with no significant difference in strength from a clinical point of view compared to lithium disilicate.⁵⁶⁻⁵⁸ ZLS might be a good alternative material to lithium disilicate in some clinical situations, but due to the lack of long term clinical knowledge of the material its use still limited.

The mechanical properties of glass-ceramic materials including ZLS are, however, not sufficient for monolithic FDPs longer than three units or placed in the posterior region. Which indicates that zirconia material might be a better alternative. The use of zirconia material for such monolithic restorations, however, has several challenges in regard enhancing the translucency while keeping adequate mechanical properties in terms of fracture strength and toughness.^{26,27,59} The need for highly aesthetic monolithic restorations with the properties of stabilized zirconia and, in particular, the unique transformation toughness led to the development of different translucent zirconia materials. Several successive generations of modified zirconia materials, with enhanced translucency have emerged on the market since 2008-2009 (Figure 2).

The translucency of zirconia depends on several factors; the grain size, the grain boundaries, and the density of the material being among the most significant ones.⁶⁰⁻⁶³ By changing the sintering temperature and the sintering time, those factors can be altered.⁶⁰⁻⁶³ In the early generation of translucent zirconia materials (translucent 3Y-TZP), it has been shown that using the optimal sintering temperature with short holding time results in a less porous material with small grain sizes, permitting higher translucency of the material.⁶⁰⁻⁶³ This translucent zirconia material consists of one phase of t-ZrO₂ like traditional 3Y-TZP. Monolithic restorations made of this early generation of translucent zirconia materials were indicated for clinical cases with less demanding aesthetics such as posterior restorations, since translucency may be insufficient if higher aesthetic result is needed. The mechanical properties, on the other hand, were comparable somewhat to the traditional 3Y-TZP as has been shown in previous studies.^{25,27}

The desire for even more translucent materials was born a few years ago. Obvious attempts at making monolithic translucent zirconia restorations used in highly aesthetic clinical cases were made. In 2014-2015, the limitation of translucency was overcome by the development of a new group of translucent zirconia materials, sometimes called Anterior Zirconia, High Translucent Zirconia, or Cubic stabilized Zirconia (CSZ). The translucency of these new materials is achieved by raising the yttria content from 3 to 6 mol%, resulting in more cubic phase formation at the expense of the metastable tetragonal phase in the microstructural level.⁶⁴⁻⁶⁹ This group of materials (such as the label DDCubeX²) is often considered to be the second generation of translucent zirconia materials. Zirconia in this group of materials is not only metastable in the tetragonal phase at room temperature as 3Y-TZP. It also contains up to 50 percent of the cubic phase crystals. The cubic crystals are larger and thus have less grain-boundaries per volume compared to the tetragonal ones. Scattered light gives the material a white opaque-looking colour, while less scattering in turn makes the material more translucent. Since substantial scattering takes place at the grain boundaries, fewer grain boundary passages give a more translucent material.⁶⁴⁻⁶⁹ Those cubic containing zirconia materials have better optical properties close to glass ceramics compared to the previous generation of translucent zirconia materials. However, from a material perspective, the high translucency achieved by raising the yttria content is at the same time lowering the inherent mechanical properties of the material to an ultimately severe extent.^{26,64-69} Consequently cubic containing zirconia materials have a flexural strength that is close to half the traditional 3Y-TZP approximately 600 MPa. Additionally, the ability for transformation toughening, which is a unique property for zirconia material, has been lost or reduced in these materials. Therefore, the low strength and toughness properties limit their use to SCs and small FDPs only anteriorly and with proper thickness.

In the last three years, what might be considered as the era of the latest generation of new multi-layered translucent zirconia materials have launched.^{70,71} These materials (for example: IPS e.max ZirCAD Prime, Katana Multi-Layer) come with two to four layers

of different translucencies within the same material. This is attained by a unique manufacturing technique that allows a combination of different zirconia materials (3Y-TZP and 5Y-TZP) in one innovative material. A Balance between strength and aesthetic properties has been achieved in these new materials by reducing the high cubic content of the previous generation to a balanced content with 3Y-TZP. Knowledge about these new zirconia materials is, however, still limited and manufacturers provide only limited information making an adequate evaluation of them very difficult.

Potential deleterious effects on zirconia materials

Low temperature degradation concerns

Failure of hip prostheses made of traditional zirconia material initially raised concerns about the effects of low-temperature degradation (LTD) on this material for dental applications.⁷²⁻⁷⁴ The LTD or aging of traditional zirconia is related to slow transformation of crystals from tetragonal to monoclinic phase (*t-m*) over time at low temperature and in a humid environment (Figure 5).⁷²⁻⁷⁴

LTD had been believed to be the responsible factor affecting negatively the mechanical properties of zirconia in the oral cavity over time.⁷²⁻⁷⁴ However, advances in dental biomaterials together with seven to ten years of clinical follow-up studies showed that LTD would probably not cause clinical failure of bilayered restorations made of traditional zirconia as frameworks/copings. Studies have failed to show signs of LTD, and on the contrary, the survival rates of zirconia frameworks have been shown as excellent.¹⁹⁻²¹ This might be related to the different environmental situation in which dental restorations are subjected to compared to the environment of hip prostheses. Additionally, in bilayered design, only small uncovered zirconia has a direct contact with oral fluids as it is virtually covered by veneering ceramics.

Figure 5. A diagram illustrating zirconia degradation process at low temperature in the presence of water. This image was reproduced from Chevalier, 2006.

On the other hand, emergence of modified translucent zirconia materials with monolithic design recently raised concerns about this problem again.^{24,69,75,76} Because of the continuous contact of oral fluids with the zirconia surface of monolithic restorations, the deleterious effect of LTD can be higher compared to restorations with bilayered design. Moreover, to enhance the translucency of zirconia material, some manufacturers reduce the alumina content because of its high reflective index which also makes the material look opaque.⁷⁷ Since one of the primary roles of alumina in the traditional zirconia is aging resistance, LTD of these new materials with a reduced alumina content can be having a higher impact as previous studies have shown.^{24,69,75-77} Because of doubtful aging resistance, low strength, and scarce clinical data, monolithic high translucent zirconia restorations need to be further investigated.

Manufactural, clinical, and technical effects

There are significant differences in strength and apparent translucency among the different zirconia materials present on the dental market. Some of the materials have higher translucency levels with a strength of 650 MPa, whereas others exhibit lower translucency levels with strength values over 1100 MPa.^{11,48,64} The reason for these differences is related to several manufactural factors that have potential influences on mechanical and optical properties of the material. These manufactural factors can only be controlled by the manufacturer, such as the chemical composition of zirconia

powder, sintering conditions, microstructural phase configurations, and presence of microstructural defects (for example; secondary phases, pores, and voids).^{60-63,70,78}

Regarding to the manufacturing process, fabrication of zirconia blanks or blocks undergoes several processing stages that play a significant role in determining the final properties. At the zirconia powder processing stage, presence and content of stabilizers (such as yttria) have effects on the microstructure phase, toughness, flexural strength, and translucency.^{48,64-69} The presence of other additives such as alumina has also potential effect on aging stability and low temperature degradation of zirconia material.^{69,76,77} At the sintering stage, both temperature and dwelling time significantly affect density (porosities), grain size, and the microstructure phase, leading to considerable variations in optical and mechanical properties of the final product.⁶⁰⁻⁶³

In general, fixed dental restorations (such as inlays, onlays, SCs, and FDPs) made of different zirconia materials can be fabricated through CAD/CAM technology.⁷⁹ Zirconia blanks or blocks that come with this technology can be fully sintered before milling using hot isostatic sintering procedure (HIP). By combining the sintering at high temperatures (between 1400 and 1500°C) and high pressure in the HIP procedure, zirconia with dense (less pores) microstructure, high values of toughness and translucency is produced.⁸⁰⁻⁸² However, fully sintered or HIP zirconia materials, at the milling stage, are harder and difficult to machine compared to pre-sintered zirconia (white stage) or non-HIP zirconia materials. Potential effects on the zirconia microstructure could occur due to the hard machining in terms of surface roughness, residual stresses, and initial monoclinic content.⁸³⁻⁸⁶ On the other hand, non-HIP zirconia materials are produced by cold isostatic pressing (CIP) or uniaxial pressing, and are partially sintered before milling. Most of the CAD/CAM systems on the dental market use the soft machining of partially sintered zirconia blanks or blocks to produce the dental appliances. After milling, zirconia is fully sintered at temperatures between 1350 and 1500°C.

Clinical and technical factors that can be controlled by the clinician or the technician also play main roles in determining the aes-

thetic and strength outcomes of zirconia restorations. Examples of these factors are crown thickness, cement material used, surface structure, and underlying tooth or core structures.⁸⁷⁻⁸⁹ Therefore, it is essential to choose the right zirconia material and to optimize the clinical and technical conditions in order to achieve maximum strength and aesthetics.

Design challenges and clinical performance of zirconia

Restoring oral function and a natural tooth appearance with favourable aesthetic and strength outcomes can be achieved using zirconia restorations with bilayered or monolithic design.^{19-23,90} Several investigations show that the chosen design plays an essential role in the high load-bearing capacity and expected high clinical success for all-ceramic restorations made of zirconia materials.⁹¹⁻⁹⁴ It is significant for any design to be able to maintain the well-known mechanical properties of zirconia while achieving optimal aesthetics, taking into account the common probable sites for clinical failures such as the veneer, the connector and the cervical margin.⁹¹⁻⁹⁴

Bilayered and monolithic designs limitations

Even with the toughening properties of traditional zirconia as a framework material for all-ceramic restorations, chip-off fracture has been reported as common clinical failure.^{19-21,90,95,96} Several investigations on tooth-supported zirconia restorations have made reference to veneer fracture with a rate of incidence ranging from 2 to 20%.^{19-21,90,95,96} When a restoration is supported by a dental implant, the veneering porcelain is even more prone to such issues.⁹⁷ Chipping, cohesive fracture within only the veneering porcelain, has been shown to be the main common fracture mode rather than the total complete fracture that extends to zirconia substructure or adhesive fracture where the porcelain fractures completely from the core material. The bilayered design with metal substructure made of high noble alloys, in contrast, showed less frequent chipping issues.^{19-21,95} This unstable behaviour of veneering porcelain upon traditional zirconia has been attributed to several possible factors discussed in the literature.^{95,98} Previous

studies pointed out that the causes of chipping might be related to the poor inherent material properties of the veneer and the sensitivity of veneering techniques.⁹⁹ It might also be associated with the improper design of the veneer and the zirconia framework, as other studies have shown.^{91,94,100-103} Different thermal properties of veneering porcelain and zirconia materials, such as thermal diffusivity and coefficients of thermal expansion, also play a significant role in generating the residual thermal stress during fast cooling leading to veneer fracture.^{102,104} Recently, some clinical studies have reported that the clinical success of veneered zirconia restorations is comparable to that observed for metal-ceramic ones, and chipping problems have been solved today.¹⁰⁵ The possible reason for these observations is the slow cooling protocols, that have been practiced which might play a role in avoiding the thermal stress formation after firing of the veneer.^{102,104,106} It might also be possible because of our improved understanding of traditional zirconia as a dental restorative material that has been available for several years. However, since the veneer chipping may be considered an unnoticed or minor problem for majority of patients, the claiming that problem has been solved only by clinical observations is not completely true. Additionally, due to several limitations of clinical studies regarding low power, bias, or poor design, the causes for chipping cannot be differentiated in the clinical studies, and the literature still lack more consistent clinical data to confirm the long-term reliability of veneered traditional zirconia restorations. Moreover, new translucent zirconia with various chemical compositions have emerged in the market, making the clinical performance of such materials with bilayered design still an open question. No studies have investigated which veneer thickness could be used without affecting the strength of translucent zirconia restoration.

Several modified designs have been proposed as attempts for reducing the high incidence of veneer chipping in traditional zirconia restorations. Some studies show that restorations with pronounced anatomical substructure have a low rate of veneering porcelain fractures compared to ones with non-anatomical substructure.^{91,101} The significant factor was the capability of the anatomical sub-

structure design to provide an even veneer thickness influencing the concentration of thermal tensile stress during firing and cooling. Other studies also show that the added occlusal thickness of the core in order to support the porcelain veneer may significantly reduce veneer chipping and fracture of zirconia crowns.^{94,102} Extension of a zirconia core in the areas where high loads are expected, can be an effective method of reducing chipping as well.¹⁰⁷ Several studies have shown the significance of design, in terms of thickness for bilayered ceramic materials, on developing residual thermal stress during cooling.^{91,94,95,102,103,107} This residual thermal stress has been hypothesized to being the leading cause for the chipping problems of veneered traditional zirconia restorations. Since the veneering porcelain is the weak link of bilayered design, determining the optimum veneer thickness for translucent zirconia restorations would be beneficial in clinical treatments seeking long-term clinical performance with excellent aesthetics.

Excluding the porcelain veneer using monolithic design might be an excellent strategy to achieve better long-term clinical performance of zirconia restorations. There seems to be a preference in the dental community toward monolithic restorations made of translucent zirconia materials in both the posterior and anterior regions. Higher load-bearing capacity can thus be achieved by monolithic design made of translucent zirconia compared to monolithic restorations made of lithium disilicate glass ceramics and traditional veneered zirconia.^{25,108} Promising results from several clinical studies investigating monolithic zirconia restorations have recorded low fracture rates.^{22,23,109,110} The main clinical limitation of monolithic zirconia restorations was mainly related to the insufficient aesthetical outcomes of the early generation of translucent zirconia materials. However, the emergence of new generations of high-translucent zirconia materials seem to have overcome this problem, but with lower strength and toughening properties as a result. Therefore, modified designs for zirconia restorations with balanced strength and aesthetics would be good alternatives for those monolithic restorations made of high-translucent zirconia materials. For instance, using partially veneered monolithic designs by adding porcelain veneer to monolithic low translucent zirconia

designed with a cut-back might be beneficial.^{28,29} A proper combination of high strength and good aesthetics can be achieved using such designs. However, the studies that use this combination of materials with different properties should consider both the veneer thickness and supporting zirconia coping/framework design.

Connector design

Several studies pointed out the significance of selecting the proper connector design and dimensions for the successful performance of zirconia FDPs.^{92,94,111-115} As shown by these studies the total fracture of zirconia FDPs predominantly occurs at the connector site (the thinnest part of a prosthesis). In general, the dimensions of the connector design depend substantially on aesthetics, hygiene control and material limitations. In metal-based FDPs, by reducing the metal connector dimensions it is possible to enhance the aesthetics taking advantage of the high tough rigid properties of the metal frameworks.^{116,117} However, in cases of all-ceramic restorations, the brittleness of zirconia does not allow the use of this technique. Instead, increasing the connector dimensions is indicated for providing more bulk material in the connector area in order to improve the structural integrity of the restoration.¹¹¹⁻¹¹⁵

In everyday clinical use, the extensive indication for monolithic zirconia FDPs is related to the high strength and toughening properties of zirconia materials compared to other monolithic ceramic materials.^{15-18,51} However, designing monolithic zirconia FDPs has several challenges with regard to achieving the optimal natural appearance while keeping adequate long-term clinical performance. For instance, adding more material on the incisal embrasures in order to increase the strength of monolithic zirconia FDPs leads to artificial-appearing restorations. This eventually affects the aesthetics outcome significantly. To avoid this, dental technicians use different strategies to produce monolithic zirconia FDPs with acceptable aesthetics. They may use sharp milling or post-milling strategies in the connector area to separate the different units to enhance the natural-appearing facial embrasures of monolithic FDPs to achieve highly aesthetic results. It is not uncommon that improper shaping of the embrasures of the connectors affects the

aesthetics outcome by drawing attention to the artificial nature of the prosthesis resulting from the absence of the separation appearance between teeth. To improve the separation appearance, the incisal embrasure should be narrow and deep to provide the individual units of the FDPs with the optimum shadows enhancing the overall aesthetics. In another approach, the FDPs might be designed with no gingival embrasures in which the connector is over contoured gingivally. However, this approach affects the patient's ability for hygiene maintenance, affecting periodontal health as there will be no access to remove plaque. These aesthetic separating strategies at the connector area need to be investigated to show the effect on the strength of monolithic zirconia FDPs.

Assessment of load-bearing capacity and clinical prediction for all-ceramic restorations

Fixed dental restorations such as crowns and bridges are considered to be long-term replacements for missing teeth. In terms of cost and time efficiency, these treatment options are costly and time-consuming. They should, therefore, be indicated when they are expected to serve for an extended period in the patient's mouth. Well-conducted clinical studies are often considered to be the optimum means to evaluate any new material, system or design for fixed dental restorations. Clinical studies, however, require a long time to be conducted with costly resources.¹¹⁸ Furthermore, testing a new design or material in a standardized environment is very difficult in clinical settings. To give a new clinical indication or to expand an existing one for any material or design, conducting laboratory studies is an advantageous pre-step. This is to ensure patient safety and to save time and cost of unfavourable treatments. Laboratory studies, additionally, allow conducting tests in a more standardized manner using less time and at a lower cost. Therefore, dental restorations made of new materials, produced with novel designs or fabricated with new processing techniques should be assessed in a laboratory set-up before clinical use.

Laboratory evaluation of the load-bearing capacity of zirconia restorations made with different designs can offer valuable information regarding the expected clinical performance of these resto-

rations in a real oral environment. Various types of laboratory tests were performed in previous studies to estimate the limit in which a restoration fails and the distribution of fatal stress within the restoration during loading.^{119,120} Load to fracture mechanical test combined with artificial aging procedures are valuable testing methods for analysing the external fracture behaviour and for determining the fracture resistance of restorations.^{25,91,93,94,101,112,114} This kind of test is not only supposed to provide the limits of strength where a restoration fails, but is also crucial for providing expected behaviour during normal function, for understanding material-design relationships, and tooth response to stress and strain. A good laboratory test is one that mimics clinical failures of dental restorations. To achieve this, significant aspects such as artificial age simulation, the complexity of the oral environment, and shape of tested specimens have to be taken into consideration when setting up these mechanical tests.¹¹⁹

Other laboratory studies use computational (digital) testing methods for analysing and obtaining information about the internal behaviour of the tested specimens such as the finite element method (FEM).¹²⁰⁻¹²⁵ In contrast to experimental investigations on real *in vitro* or *in vivo* specimens, FEM has many benefits. It is a repeatable method that might be modified as it is needed, and there are no considerations for ethical approvals. The role of FEM for solving sophisticated design and structural problems has received increased attention across several disciplines such as industry, civil, and aeronautical engineering. In the field of dentistry, recent years have witnessed a growing academic interest in FEM as a research tool for analysing stress and deformations on structures of interest such as implants, bone, different kinds of teeth preparations, and post and core restorations.¹²⁰⁻¹²⁵ The FEM of simulated occlusal loads can also be used to optimize the shape of restorations by studying the internal stress within teeth and restorations with different dental materials and designs.

The need for continuous design improvement

Based on previous clinical observations, the general use of zirconia restorations with monolithic design has recently increased in eve-

ryday clinical practice. This is related to our previous knowledge of well-known toughening and strength properties of traditional zirconia materials as well as the recent enhancements in the translucency of the newly developed brands of zirconia materials. However, recent advances in the field of dental materials and technology have led to a renewed interest in the strength of zirconia restorations which is reduced due to the enhanced translucent properties. Proper design selection is one of the requirements to ensure excellent outcomes for patients who are seeking strong, long-term, fixed dental restorations with highly aesthetic results. Since this field is active and dynamic, and new brands of zirconia materials are continually emerging, clinicians' knowledge should entail the fundamental aspects of material and design for zirconia. They should be able to evaluate advancements in the field critically and make the right choice in the use of any new material or design, that allow the prosthesis to function adequately in the long term.

AIMS

The overall aim of this thesis was to evaluate translucent and high-translucent zirconia-based crowns and fixed dental prostheses regarding the effect of design modifications in order to enhance the aesthetics on fracture resistance.

Specific Aims

To evaluate fracture strength of veneered translucent zirconium-dioxide crowns designed with different porcelain-layer thicknesses. (*Study I*)

To describe different designs of partially veneered (semi-monolithic) crowns made of translucent and high-translucent zirconia materials and to evaluate the effect on fracture resistance and fracture mode. (*Study II*)

To investigate the load-bearing capacity and failure mode of monolithic zirconia FDPs fabricated with different connector designs and embrasure shaping methods. (*Study III*)

To evaluate the influence of the framework designs on the stress distribution within tooth-supported partially veneered FDPs made of translucent zirconia under simulated loads using a 3 dimensional-finite element analysis (3D-FEA). (*Study IV*)

MATERIALS AND METHODS

This thesis encloses three laboratory studies using clinically relevant load in fracture tests in combination with aging procedures involving mechanical preloads and thermal cycles. Additionally, the thesis involves one laboratory study utilized FEM.

Laboratory procedures in study I, II, and III

Part 1: specimen preparation

Study I

Sixty standardized SCs made of translucent zirconia, divided into six groups of 10, were used. The six groups were divided according to different thicknesses of the porcelain layer on translucent zirconium dioxide cores of equal thickness (Table 1).

Table 1. Porcelain layer thicknesses in the different groups.

Groups	Thickness of porcelain layer	Thickness of zirconia core	Overall thickness of crown (preparation depth)
Group 2.5	2.5 mm ^c	0.5 mm	3.0 mm
Group 2.0	2.0 mm ^b	0.5 mm	2.5 mm
Group 1.0	1.0 mm ^a	0.5 mm	1.5 mm
Group 0.8	0.8 mm ^b	0.5 mm	1.3 mm
Group 0.5	0.5 mm ^c	0.5 mm	1.0 mm
Group 0.3	0.3 mm ^c	0.5 mm	0.8 mm

NB: ^a Optimal thickness according to manufacturer’s recommendations, ^b thickness within the recommendations, and ^c thickness exceeding the recommendations.

The gradual reduction in porcelain thicknesses, from group 2.5 to group 0.3 (Figure 6), resulted in a corresponding decrease in preparation depths of the abutment dies without changing the outer dimensions of any of the crowns.

Figure 6. Abutment dies with different preparation depths for the different groups that have the same outer dimensions. NB: Groups with veneer thicknesses exceeding manufacture’s recommendations are indicated in red.

For specimen preparation, a metal die resembling a molar crown preparation was made for an all-ceramic crown with a 1.2 mm deep 120° cervical chamfer and 15° angle of convergence. The die was replicated using a silicone impression material (President®, Coltene AG, Altstätten, Switzerland). The impression was poured with die stone material (Vel-Mix™, Kerr Corporation, Orange,

CA) to create the master die. By using a double scan technique, the master die was scanned following scanning of a wax-up of a full anatomic crown with an occlusal thickness of 3.0 mm and an axial thickness of 3.0 mm. The scanning was carried out with a dental laboratory scanner (D900L, 3Shape, Copenhagen, Denmark) and the data were then transferred to a computer running CAD software (Autodesk, San Rafael, CA). By using the CAD software, the first crown/die complex was copied and adjusted by decreasing the preparation depths and correspondingly reducing the crown thickness toward the dimension settings of the different six groups (Figures 6,7). In this way, the outer dimensions of the crowns in all

Figure 7. An illustration showing the CAD files of the different groups with overall crown thicknesses (core/porcelain). All crowns had an equal core thickness but different porcelain thicknesses. The gradual reduction in crown thicknesses resulted in a corresponding decrease in preparation depths without changing the outer dimensions (the X-imaginary line) of any of the crowns. The red dotted line denotes the core/porcelain interface.

groups were identical, but the preparation depths of the abutment dies differed to match the different crown thicknesses. Six different CAD files (STL format) were thus produced and subsequently sent to a certified BruxZir milling center (Cosmodent AB, Malmo, Sweden) where they were used to produce 60 translucent zirconium dioxide cores of equal 0.5 mm-thickness (BruxZir® Solid zirconia, Glidewell dental laboratories, San Diego, CA). The CAD data were also used to produce 60 dies in six different preparation depths corresponding to the different six groups (Figure 6). All abutment dies were milled from a polymer material (POM, Nordbergs Tekniska AB, Stockholm, Sweden).

The zirconia cores were veneered with porcelain recommended by the manufacturer (IPS e.max Ceram®, Ivoclar Vivadent, Schaan, Liechtenstein). A custom-made knife was used to standardize the outer shape and dimension of the crowns (Figure 8). While applying the porcelain layers, the crown over the abutment was placed and fixated to the brass holder that was rotated manually, permitting the knife to carve the porcelain into the final shape.

Figure 8. A custom-made knife used for the porcelain build-up, with a zirconia core mounted on the abutment die.

Each crown underwent five porcelain firing cycles: liner, wash, dentin 1 and 2 and finally glaze firing. The firing cycles were carried out in a calibrated furnace (Ivoclar P 500, Ivoclar Vivadent, Schaan, Liechtenstein) according to the manufacturer's recommendations (Table 2).

Table 2. Firing parameters for IPS e.max Ceram® .

Firing program	Stand-by temperature [°C]	Closing time [min]	Heating rate [°C/min]	Firing temperature T ₁ [°C]	Holding time H ₁ [min]	Firing temperature T ₂ [°C]	Holding time H ₂ [min]	Vacuum 1 [°C]	Vacuum 2 [°C]
ZirLiner firing	403	4	40	960	1	-	-	450	959
Wash firing	403	4	90	650	-	730	2-3	400	650
1 st Dentin firing	403	4	90	650	-	730	2-3	400	650
2 nd Dentin firing	403	4	90	650	-	730	2-3	400	650
Glaze firing	403	6	60	710	1	-	-	450	709

Study II

One hundred standardized SCs with different designs were produced and divided into five groups (n=20): monolithic (M), partially veneered monolithic (semi-monolithic) with 0.3 mm micro coated buccal veneer (SM0.3), semi-monolithic with 0.5 mm buccal veneer (SM0.5), semi-monolithic with 0.5 mm buccal veneer supported by wave design (SMW), and semi-monolithic with 0.5 mm buccal veneer supported by occlusal cap design (SMC) (Figure 9). Each group was divided into two subgroups (n=10) according to the crown materials used, translucent zirconia (BruxZir® Solid zirconia) and high-translucent zirconia (Zirkon BioStar HT, Siladent, Goslar, Germany).

Figure 9. Different crown designs shown: (A) in the CAD software and (B) after milling and sintering.

The semi-monolithic SCs in this study were designed with specific dimensions through the CAD software (Figure 10).

Figure 10. Illustrations showing dimensions of the crown designs. The black line represents the interface between the core and the veneer. The arrows denote the local dimensions of the crowns.

The methods applied for study I were also used for preparing the crowns and abutments in study II (the preparation, double scan technique, milling centre, and veneering process).

Study III

Seventy standardized 4-unit zirconia FDPs (with two premolar pontics and two retainers) were fabricated. The FDPs were divided into seven groups (n=10) including the control; all the FDPs were produced with different embrasure shaping strategies for the connector (Table 3, Figure 11).

Table 3. Description of the study groups.

Connector design and embrasure shaping methods	FDPs design
Sharp embrasures	Monolithic 4-unit FDPs
Blunt embrasures	Monolithic 4-unit FDPs
Blunt embrasures with no occlusal embrasures	Monolithic 4-unit FDPs
Blunt embrasures and interproximal separations made with diamond discs at soft stage	Monolithic 4-unit FDPs
Blunt embrasures and interproximal separations made with diamond discs at fully-sintered stage	Monolithic 4-unit FDPs
Blunt embrasures and interproximal separations accentuated by localized porcelain build-up	Monolithic 4-unit FDPs
Control group with default milling settings	Veneered 4-unit FDPs

For design and production of the FDPs and abutment models, a plastic model of an upper jaw was used (KaVo YZ, KaVo Dental GmbH, Biberach, Germany), and two abutment preparations were made for the canine and the first molar. The two premolars were removed to provide space for pontics in a 4-unit FDP. The abutment preparations were designed to provide the space allowance for an all-ceramic restoration, with a 120° chamfer and 15° convergence angle.

Figure 11. The FDPs with different connector designs gained by different embrasure shaping methods. NB: *Two different groups of monolithic FDPs were fabricated with interproximal separations made with diamond discs at the soft stage and at the fully sintered stage.

A full arch impression (President®) was made and poured with die stone (Vel-Mix™) to produce a master cast. Using a double scan technique, the master cast was scanned first by using a laboratory scanner (D900L). Subsequently, two different wax-ups were scanned: full anatomy for monolithic groups and reduced anatomy (cut-back) for the veneered group.

After scanning of the master cast and the wax-ups, the data were transferred to a computer loaded with the CAD software. The connector dimensions of all the FDPs were set identically to 3 mm × 3 mm except for the FDPs designed without occlusal embrasure, to which 1 mm extra was added occlusally. In all groups, the minimum thickness of the occlusal core was set to 1 mm, and the minimum thickness of the axial walls was set to 0.8 mm. The FDPs in the control group were designed to be fully veneered with 1 mm porcelain according to manufacturer recommendations (IPS e.max Ceram®). In all groups, the gingival and occlusal radii of the em-

embrasures in the connector areas were set according to the default settings of the CAD program, except for monolithic FDPs with interproximal porcelain separation where a large radius cut-back was used to accommodate the buccal veneering material.

The CAD data were sent to a certified production centre (Cosmodent AB, Malmö, Sweden) where they were used to produce the FDPs: 60 monolithic translucent zirconia (BruxZir® HT, Glidewell dental laboratories, San Diego, CA) and 10 traditional zirconia frameworks (BruxZir® Solid zirconia). The FDPs with sharp embrasures were fabricated using a sharp milling tool with a 0.30 mm diameter (Zenotec, Wieland Dental/Technik GmbH, Pforzheim, Germany). The FDPs in the other groups were fabricated with a blunt milling tool with a 1 mm diameter followed by different post-milling embrasure shaping procedures on the same intermediate connector. No such shaping procedures were carried out in two groups (the FDPs with blunt embrasures and the FDPs designed with no occlusal embrasures). In two groups of the FDPs, the intermediate connector was separated manually using a separation disc (S-H 394F 220, Sunshine diamonds GmbH, Langenhagen, Germany) at both the soft stage and the fully sintered stage. Disc separation was standardized and performed using the same type of disc, with a rotational speed of 17 rpm for fifteen seconds. New discs were used for each FDP in these two groups. One operator, using the same hand pressure as far as possible in a manual process, did the disc separation. The connector area was examined by using a light microscope (Leica DFC 420, Leica Microsystems CMS GmbH, Wetzlar, Germany) to confirm the standardization of both the location and the depth of the notch area created during the disc separation step.

Monolithic FDPs with interproximal porcelain separation and the FDPs in control group were not veneered but instead underwent a veneer-simulating process in which the frameworks were subjected to heat treatment according to manufacturer recommendations to simulate the firing cycles of the veneering porcelain (IPS e.max Ceram®) (Table 2).

Seventy abutment models were produced from a polymer material (POM-C GF25, Simrishamns Mekaniska AB, Simrishamns, Sweden) with a modulus of elasticity close to dentin (9 GPa).

Part 2: aging procedures and cementation

All the SCs (study I, II) and the FDPs (study III) underwent artificial aging procedures before they were loaded to fracture. These procedures included thermocycling, storing in a 37°C moist environment, and mechanical preloading cycles.

Thermal cycles

In study I and II, all the SCs underwent 10,000 thermal cycles in two water baths of a specially built thermocycling device at temperatures of 5°C and 55°C. Each cycle lasted for 60 seconds, 20 seconds in each bath, and 10 seconds for transfer between the baths. The same thermocycling protocol but with a different device (THE-1100, SD Mecharonik GmbH, Feldkirchen-Westerham, Germany) was applied for all FDPs in study III (Figure 12).

Figure 12. Thermocycling devices used in study I, II (A), and study III (B).

Cementation

In study I and II, the crowns were cemented to the respective abutment dies using an MDP containing resin cement (Panavia F 2.0 Kuraray Medical Inc., Okayama, Japan). While, in study III, the FDPs were cemented onto the abutment models by using dual-cured resin (Panavia V5, Kuraray Medical Inc.) following manufacturer instructions.

The inner surfaces of the crowns and the FDPs were sandblasted using 50 μm aluminum oxide at a pressure of 2 bar, and then rinsed with water according to manufacturer's recommendations. During the cement setting, the crowns and the FDPs were loaded in the direction of insertion with a load of 15 N (Figure 13).

Figure 13. Tool used to standardize the seating load of 15 N during cementation.

Before cementation, the abutment models were roughened with air-abrasion (Basic Quattro IS, Renfert GmbH, Hilzingen, Germany) using 110 μm aluminum oxide at a pressure of 2 bar from a distance of 10 mm at a 90° angle to the abutment surface to increase the surface area.

In study I and II, they were steam cleaned and subsequently treated with ED PRIMER II A and B (Kuraray Medical Inc.). While in study III, the abutment models were treated with two primers (Tooth Primer, Clearfil Ceramic Primer; Kuraray Medical Inc.) according to manufacturer instructions.

Storing in an incubator

All tested specimens were stored in a plastic container inside an incubator (Mettmert Incubator, Mettmert GmbH, Schwabach, Germany) and were wrapped in wet paper at a temperature of 37°C for 60 days before they were loaded to fracture.

Preload mechanical cycles

In study I, II, III, a specially made preloading device was used (MTI Engineering AB/Pamaco AB, Malmö, Sweden) to subject the

specimens to 10,000 load cycles between 30 and 300 N, with a load frequency of 1Hz (Figure 14). The specimens were mounted submerged in distilled water at 10° inclination relative to the vertical plane in the preloading machine. A 0.2 mm thick plastic foil (PE-Baufolie, Probau, Bauhaus, Zug, Switzerland) was placed between the indenter and the specimens during preloading to distribute the loading forces evenly and to prevent cone-crack formation.

Figure 14. Test set-ups used for the cyclic preload and the load to fracture in study I, II, III. NB: The specimens were submerged in water, and a plastic foil was used to spread the load evenly over the loaded surfaces. The specimens were loaded at 10° of inclination in relation to the loading direction.

In study I and II, preload cycles were applied on the occlusal surface of the crowns by a stainless-steel ball with a diameter of 4 mm. In study II, loads were applied at the highest point of the inner buccal cusp inclination of the crowns (Figure 15).

Figure 15. Different indenters used for cyclic preload and load to fracture tests in study II: (a) a ball indenter (4 mm in diameter) used in cyclic preload and (b) a special indenter designed to prevent sliding movements during load to fracture test.

In study III, the preload cycles were applied centrally on the buccal cusp of the second premolar pontic in the control group by using a special indenter designed to prevent sliding movements during loading. The other groups in this study were loaded with a ball indenter (4 mm in diameter) on the mesial marginal fossa of the second premolar pontic.

Part 3: load to fracture test and fracture mode analysis

Load to fracture test

After artificial aging procedures, in study I, II, and III, all specimens were mounted in a test jig at a 10° inclination, in the same way as during the preloading step. Specimens were loaded to fracture by using a universal testing machine (Instron 4465, Instron Corp, Norwood, MA). All specimens were submerged in water during loading with a plastic foil placed between the indenter and the specimens to distribute the loading forces evenly. The cross-head speed was set to 0.255 mm/min, and the load at fracture values were registered. The fracture was defined as either an apparent crack, a load drop in the graph or an acoustic event, whichever came first. The acoustic event was ascertained by hearing a cracking sound that is conveyed by the changes in the load-path graph on a computer that is connected to the testing machine.

Fracture mode analysis

Examination of the fracture surfaces and determination of fracture mode were performed by both gross visual and microscopic (Leica DFC 420) assessments by one investigator in study I, II, and III. In study I and II, the fracture surfaces of the crowns were classified into adhesive (denoting fracture in the interface between the veneer and the core), cohesive (denoting fracture within the buccal veneer, i.e., chipping), or complete (denoting the fracture of both the veneer and the core). Failure modes of the FDPs, in study III, were classified according to the starting points of the fracture and its propagation path.

Laboratory procedures in study IV

Part 1: creating 3D solid models

Simplified 3D solid models of prepared abutment teeth (first premolar and first molar) with different 3-unit FDPs based on the designs were created (Table 4, Figure 16). The different designs of this study were adopted from study III.

Table 4. Models with different FDP designs created in study IV.

Model	Design of 3-unit FDPs
Control	Monolithic zirconia
SM0.3	Semi-monolithic zirconia with 0.3 mm buccal veneer thickness
SM0.5	Semi-monolithic zirconia with 0.5 mm buccal veneer thickness
SMC	Semi-monolithic zirconia with 0.5 mm buccal veneer thickness supported with cap design
SMW	Semi-monolithic zirconia with 0.5 mm buccal veneer thickness supported with wave design

The teeth preparations, to receive all-ceramic FDP, were made on mandibular first premolar and first molar plastic models (KaVo YZ; KaVo Dental GmbH) using diamond burs (Two stripper, Abrasive Technology Inc., Lewis Center, USA). The second premolar plastic model was removed from the plastic mandibular arch to provide space for the pontic in a three-unit FDP. The teeth preparations were performed with a 15° angle of convergence between the axial walls. The height of preparations was set to 5 mm based on the average dimensions of prepared molars and premolars.¹²⁶ The marginal areas of the two abutments were prepared with a rounded 120° chamfer angle.

For simplicity of modelling, the occlusal surfaces of the preparations were flat. The marginal, axial, and occlusal surface reductions were 0.5 mm, 0.8 mm, and 1.5 mm respectively. The only exception for the axial surfaces was the buccal surface; it was reduced to 1.2 mm to accommodate the veneering porcelain in the

different semi-monolithic designs. All line angles of preparations were rounded to decrease stress development in the FDPs.

A full-arch impression (President; Coltene AG) was made and poured with dental stone (Vel-Mix; Kerr Corp) to produce a master cast. The FDP model was completed using a double-scan technique in which the master cast was scanned first by a laboratory scanner (D900L; 3Shape) followed by scanning the full-contour wax-up of the FDP. The scanning data of the FDP and the prepared teeth were transferred to a computer loaded with a CAD software (3Shape program) and were saved as an STL file. The solid geometries of the bone block, the prepared teeth, the FDPs with different components (veneer, zirconia, cement) were rebuilt from the STL file with a 3D modelling software (Creo Parametric 4.0, PTC, Needham, USA). For the convenience of the simulation process, the prepared teeth in the CAD software were fixated in a simplified block of alveolar bone with no separating PDL gap in between. Additionally, a uniform cement gap was included in the 3D model.

The FDPs have simplified anatomical occlusal surface with only two cusps (buccal and lingual). The connector dimensions (height, width) of all the FDP were set identically to 4 mm × 4 mm and were designed with round interproximal embrasures. The veneer-zirconia junction of all models was designed with rounded corners to avoid stress concentration. The total occlusal thickness, in all models, was set to 1.5 mm; including only zirconia material in the control model, and zirconia with veneer in semi-monolithic models.

Figure 16. CAD images with illustrations showing dimensions of all models with variations in zirconia framework design and veneer thickness. NB: In the 2D cross-section, the blue shade represents the veneer thickness layer.

Part 2: generating and processing of FEA models

The 3D solid models with different components (veneer, zirconia, cement, dentin, bone) were exported to the FEA program (ANSYS Mechanical Pro; Professional Version 19.2, Canonsburg, USA). After uploading the models, the meshing process was performed on the models with four-node linear tetrahedral elements, based on the convergence test. The criterion for convergence analysis was set to be less than 6 % changes in the highest von-Mises stress of bone and dentin between the element sizes.¹²⁷

For accurate analysis, the mesh size was set to 0.7 mm for bone and dentin components. For the zirconia framework and the veneer

components, it was set to 0.25 mm and 0.1 mm respectively. The total number of elements for each model was around 1,086,445 with a small range of difference ($\pm 101,232$) between the models because of the different designs. The mechanical properties in (Table 5) were applied for each element in 3D-FEA model.

Table 5. Material and tissue elastic properties of 3D-FEA models.

Tissue/Material	Poisson's ratio	Young's modulus (GPa)	Ultimate tensile strength (MPa)
Bone	0.31	16.3	44
Dentin	0.31	18.6	48
Cement	0.27	6.48	97
Translucent zirconia	0.30	210	1000
Porcelain	0.19	74	100

NB: All values were obtained from the published data for FEA.¹²⁸⁻¹³¹

The 3D-FEA models used in this study were based on the following assumptions: (1) All components of the models were linear elastic, isotropic, and homogenous. (2) There were no flaws in any component. (3) All components of the models were considered perfectly bonded. (4) There was a uniform 100 μ m cement space based on values from previous studies.^{132,133} (5) The effect of the PDL on generating the stresses within the models was negligible, and the gap junction between the bone and dentin was immovable with no translation or rotation in any direction. (6) The cementum layer that covers the root surface and the pulp chamber were negligible and were included in the dentin of the teeth since they are thin components with no or lesser effects on stress generation.¹³⁴

The simulation and the numerical analysis of 3D-FEA models were performed using the Ansys program to evaluate the different stress fields for the different models. The simulated loading forces of 300 N were applied occlusally over six different loading points (C1-C6). The loading forces were applied at a 10° oblique angle relative to the long axis of the clinical crown. The oblique loading

forces were directed toward the buccal cusps of the pontic and one of the retainers. (Figure 17). The loading points were simulated over the 1.0 mm^2 circular areas on the occlusal surface.

After the simulation, the maximum principal stress, the maximum shear stress at the zirconia-veneer interface, and minimum safety-factor levels were calculated. The analysis of maximum principal stress was used to differentiate areas of tensile (positive) and compressive (negative) stress in the veneer and zirconia components of all models.

Figure 17. Schematic illustration of loading points (C1-C6) and load direction. NB: C1, C2, and C3 indicate loading points at pontic where veneer is fully supported by zirconia, partially supported by zirconia, and partially supported by zirconia near connector respectively. C4, C5, and C6 indicate loading points at retainer where veneer is fully supported by zirconia, partially supported by zirconia, and partially supported by zirconia near proximal ridge respectively.

Statistical analysis

Several statistical tests were used in the current thesis for interpreting the results of the four studies. These tests were chosen in relation to the type of independent and dependent variables and the sample number. Parametric tests were applied to analyse the differences between the tested groups in the four studies. All statistical tests were performed with SPSS software (IBM SPSS Statistics, IBM Corp, Chicago, IL). A significant level of 0.05 was set in all tests ($p \leq 0.05$). All statistical analyses in the four studies were done by an experienced professional statistician.

In study I, II, and III power analyses to determine the sample numbers were based on previous studies where differences regarding the level of significance and standard deviation were detected among the zirconia-based specimens. The normal distributions of data were verified using the Shapiro–Wilk test. When we performed this test, we got a p-value of 0.196. This indicates that the assumption holds and no departure from a normal distribution and that ANOVA is a valid analysis. The numerical parameters obtained with the load to fracture mechanical tests in study I and III were analysed with a one-way ANOVA, followed by Tukey's test to examine if the mean differences of fracture loads among the groups were statistically significant. In study II, the differences in the load at fracture data among the tested groups with different designs and materials were analysed using a two-way ANOVA, followed by Tukey's post hoc test. Regarding fracture mode/failure mode, the differences in the way that the crowns (study I and II) and the prostheses fractures (study III) were analysed using Fishers' Exact probability test. The level of significance was set to p-values ≤ 0.05 . In study IV, no statistical analysis was conducted since the results gained by finite element stress analysis were variances that occur as a result of non-mathematical calculations.

RESULTS

Study I

Load at fracture

All specimens were visually intact through all steps of the artificial aging, and neither fractures nor crack formations were detected in the crowns, or the abutment dies. Load at fracture, levels of significance and fracture modes are summarized in (Table 6). The one-way ANOVA showed significant differences in load at fracture among the groups. Group 1.0 showed significantly ($p \leq 0.05$) highest load at fracture in comparison with groups 2.5, 2.0, and 0.3. There was no significant difference ($p > 0.05$) in load at fracture among groups 1.0, 0.8 and 0.5. Groups 2.0 and 2.5 showed significantly lower load at fracture in comparison with all other groups but with no significant differences between them ($p > 0.05$).

Fracture mode

There were significant differences in the way the SCs fractured. Two kinds of fracture mode were seen visually, either complete fracture involving both the veneer and the core or cohesive fracture denoting fracture within the veneer only (Figure 18). All the SCs in groups 2.5 and 2.0 and more than 80% of the SCs in groups 1.0, 0.8 and 0.5 showed cohesive fracture mode. Conversely, there were significantly ($p \leq 0.05$) more complete fractures in group 0.3 compared to all other groups. No adhesive fractures, involving the bonding area between the veneer and the core, were observed in the SCs under the light microscope.

Table 6. Load at fracture in Newton.

Groups	Group 2.5	Group 2.0	Group 1.0	Group 0.8	Group 0.5	Group 0.3
Specimen no.						
1	882	946	1402	1277	1138	1042
2	1064	1009	1355	1512	1664 Φ	1023 Φ
3	803	870	1727	1313	1483 Φ	1205 Φ
4	701	750	1256	1154	1137	1226 Φ
5	867	1203	1430	1050	1326	1154
6	848	805	1773	1748 Φ	1039	1397 Φ
7	863	969	1485	1076	1273	1086
8	903	816	1828	1297	1066	1297 Φ
9	843	910	2118 Φ	1639	1475	1310 Φ
10	732	822	1030	1063	1263	1276 Φ
Mean*	851 ^a	910 ^a	1540 ^b	1313 ^{b,c}	1286 ^{b,c}	1202 ^c
SD	98	131	318.9	247	203.7	123.9
Fracture mode (complete/cohesive)	0/10	0/10	1/9	1/9	2/8	7/3

NB: * Means with the same letters in superscript (denoted a, b, c) did not show any significant difference in load at fracture ($p>0.05$). Φ Complete fracture mode.

Figure 18. Example of the two fracture modes: (A) Complete fracture (B) Cohesive fracture.

Study II

Load at fracture

There were significant differences in load at fracture among the groups depending on the materials used and the different designs tested ($p \leq 0.05$) (Figure 19). Generally, the SCs made of translucent zirconia showed higher load at fracture compared to those made of high-translucent zirconia, regardless of design ($p < 0.001$).

Figure 19. Comparison of mean fracture load (Newton) among the crown designs with two different materials. Notes: *Mean with the same letters (denoted a, b, and c) did not show any significant difference in load at fracture ($P > 0.05$).

Regarding design, the semi-monolithic groups with 0.3 mm micro coated buccal veneer (SM0.3) showed the highest load at fracture of all the other semi-monolithic groups, with no significant difference compared to the M and SMW groups ($p>0.05$). The SCs in the SM0.5 group showed significantly lower load at fracture compared to the SCs in the SMW and SMC groups.

By comparing the designs in both materials used; the M, SM0.3, and SMW groups of translucent zirconia showed significantly higher load at fracture compared to all the groups of high-translucent zirconia. There was no significant difference in load at fracture between the M group of high-translucent zirconia compared to the SMC and SM0.5 groups of translucent zirconia.

Fracture mode

There were substantial differences among the groups in the way that the SCs fractured. Two fracture modes were seen visually and under the light microscope: complete fracture and cohesive veneer fracture (Figure 20).

Figure 20. Fracture modes of the crowns: (a) complete fracture in M, SM0.3, and SMC groups and (b) cohesive veneer fracture in SMW and SM0.5 groups.

All the SCs in the M and SM0.3 groups (100%), and all but one of the SCs in the SMC groups (95%) showed complete fractures with significant differences ($p\leq0.05$) in the fracture mode compared to the SCs in the other two groups (SMW and SM0.5). The SMW and SM0.5 groups showed (70 and 55%, respectively) cohesive veneer fractures. None of the tested SCs showed an adhesive fracture (Figure 21).

Figure 21. Comparison of fracture mode among the crown designs with two different materials.

Study III

Load at fracture

Load at fracture and levels of significance in the groups are summarized in (Table 7). There were significant differences in load at fracture among the groups based on the connector designs gained by different embrasure shaping methods used in the study ($p < 0.001$).

The FDPs fabricated with interproximal porcelain separation showed significantly the highest load at fracture of all groups ($p < 0.001$), with no significant difference compared to the FDPs fabricated with no occlusal embrasures ($p > 0.29$). The FDPs fabricated with blunt embrasures showed significantly higher load at fracture compared to the FDPs in the control group and the FDPs with sharp embrasures ($p < 0.001$). The FDPs fabricated with interproximal disc separation at soft stage had the lowest load values numerically, but with no significant differences compared to the FDPs fabricated with interproximal disc separation at fully-sintered stage ($p > 0.23$).

Table 7. Load at fracture (N)

Groups Specimen number	Sharp embrasure	Blunt embrasure	No occlusal embrasure	Separation with disc (soft)	Separation with disc (fully-sintered)	Separation with porcelain	Control
1	367	820	743	360	449	1014	694
2	394	745	855	422	380	959	705
3	463	800	1076	334	516	1011	771
4	464	774	988	386	332	947	691
5	374	1033	1220	330	442	1072	642
6	376	830	744	426	637	1006	708
7	477	945	1107	340	503	990	606
8	608	1099	736	300**	370	1177	688
9	347	858	1048	377	460	1185	561
10	300**	831	827	300**	578	1022	823
Mean* SD	417 ^a 87	873 ^b 115	934 ^{b,c} 175	357 ^a 45	467 ^a 94	1038 ^c 82	689 75

* Means with the same letters in superscript (denoted a, b, and c) did not show any significant difference in load at fracture (p>0.05). ** FDI's fractured at 30 to 300 N preload.

Fracture mode

All FDPs fractured in the intermediate connector of the two pontics, and no abutment fracture was observed. There were significant differences among the groups regarding failure modes ($p < 0.001$). Table 8 presents the distribution of failure modes of the FDPs among the tested groups. Failure mode of the FDPs with sharp embrasures and the FDPs with interproximal disc separations showed that all fractures (100%) started from the gingival side of the intermediate connector in the notch area created by either the disc or the sharp milling tool, propagating towards the loaded pontic (Figure 22). By contrast, all fractures (100%) in the FDPs of the other groups occurred at the distal corner of the intermediate connector on the gingival side, propagating towards the loading site (Figure 23).

Figure 22. Fracture mode of the FDPs with interproximal disc separation (lingual view). Fracture started from the gingival side of the intermediate connector in the notch area (triangle) created during post-milling disc separation and extended toward the loading site.

Figure 23. Fracture mode of the FDPs in the control group (buccal view). The fracture occurred at the distal corner (triangle) of the intermediate connector on the gingival side and propagated toward the loading site.

Table 8. Distribution of failure modes for each group.

Groups	Starting points of the fracture	
	Notch area in intermediate connector	Distal corner of intermediate connector
Sharp embrasure	10	0
Blunt embrasure	0	10
No occlusal embrasure	0	10
Separations with disc (soft)	10	0
Separations with disc (fully-sintered)	10	0
Separation with porcelain build-up	0	10
Control	0	10

Study IV

The maximum principal (tensile and compressive) stresses introduced to both veneering porcelain and zirconia framework of the different model designs are presented in (Table 9) and (Figures 24,25). The results demonstrated that the veneer thickness and the zirconia framework design had considerable effect on the maximum stress distributions within the simulated 3D-FEA models in the 10° oblique load condition.

Table 9. Maximum principal stress in veneering porcelain and zirconia frameworks of different model designs at 10° oblique load condition. All values in megapascal (MPa).

Model design	Veneering porcelain		Zirconia framework	
	Maximum principal tensile stress (positive values)	Maximum principal compressive stress (negative values)	Maximum principal tensile stress (positive values)	Maximum principal compressive stress (negative values)
Control	N/A*	N/A	53.4	100.5
SM0.3	61.6	100.7	53.6	27.5
SM0.5	70.1	113.8	52.2	13
SMC	21.2	1.6	52.9	157
SMW	58.5	121.4	42.3	10.4

*N/A: Not applicable

Concerning veneering porcelain in SM0.3 model with a 0.3 mm-veneer thickness, the maximum principal stress was lower compared to SM0.5 model with a 0.5 mm-veneer thickness. The SMC model resulted in the smallest maximum principal stress in the veneering porcelain compared to all other models (SM0.3, SM0.5, SMW). Regarding the stress in zirconia framework, all models had comparable results in maximum principal tensile stress, except SMW model had a lower value. The SMC model resulted in a larger maximum principal compressive stress compared to all other models.

Figure 24. Maximum principal stress distributions for 3D-FEA models with different FDPs designs.

Figure 25. Different views of 3D-FEA models; (A) FDP with cap design, (B) FDP with wave design, (C) FDP with 0.3 mm veneer thickness, and (D) FDP with 0.5 mm veneer thickness.

Table 10 shows the locations of the maximum principal stress in the different models, taking into consideration the six loading points (C1-C6) in this study. The results showed that the maximum principal tensile and compressive stress levels were located in the veneer component of SM0.3, SM0.5, SMW models with semi-monolithic designs. The only exception to this pattern was SMC model with cap design where the maximum principal tensile stress

levels were observed at the cervical areas of the zirconia framework. Similarly, when analysing the model with the monolithic design, the peak of tensile stress levels occurred at the cervical areas of the monolithic zirconia, in addition to the loading sites.

Table 10. Locations of maximum principal stress in the different models regarding the six loading points at 10° oblique loads.

Model design	Location of maximum principal stress*	
	Maximum tensile stress	Maximum compressive stress
Monolithic	M1	C4
Semi-monolithic zirconia (0.3 mm)	C5	C4
Semi-monolithic zirconia (0.5 mm)	C2	C4
Semi-monolithic zirconia (cap)	M2	C3
Semi-monolithic zirconia (wave)	C3	C4

*(C) indicates locations of maximum principal stress levels at the loading points; C2, C3 indicate the loading points at pontic, C4, C5 indicate the loading points at retainer. (M) indicates locations of maximum principal stress levels away from the loading points; M1 Marginal area of the non-loaded retainer (lingual surface), M2 Marginal area of the loaded retainer (buccal surface).

Regarding the maximum shear stress values at the zirconia-veneer interface, SM0.3 model showed the highest maximum shear stress, while SMW model had the lowest value (Figure 26).

Figure 26. Maximum shear stress at zirconia-veneer interface for each model design.

For the specific 10° load condition, the 3D-FEA models with different FDP designs showed different minimum safety-factor levels (Figure 27).

Figure 27. Minimum safety-factor levels for each model design at 10° oblique load condition.

DISCUSSION

The main findings of this thesis

The four papers included in this thesis can be considered among a series of investigations in literature that have shown the significance of design on the performance of all-ceramic restorations. The overall results emphasize the importance of the design of the core/framework, the connector area and the veneer on the load-bearing capacity as well as fracture mode of translucent zirconia restorations. The results of these *in vitro* studies can also be considered as a sound basis for future clinical studies.

Analysis of the load-bearing capacity

The clinical use of materials for dental restorations, such as SCs and FDPs, is determined by the strength and toughness properties that are considered among the most preferred material properties used in dental material science. Any dental restoration should meet the clinical requirements with adequate strength and toughness properties to gain high load-bearing capacity to withstand occlusal loads generated in the oral environment. Previous studies have evaluated the range of normal maximum occlusal loads in the anterior and posterior regions.¹³⁵⁻¹³⁷ The reported values range from 30 N to 1000 N depending on location, age, gender and occlusal status among other factors. In general, loads are higher posteriorly compared to anteriorly. Not only the occlusal loads need to be considered when fabricating a dental restoration, but also the load duration.^{119,138} Cyclic loads and fatigue loading should be considered rather than only the ultimate strength of the restoration in re-

lation to the maximum bite force values.^{119,138} Although paper I, II, and III of this thesis use a laboratory set-up with a specific protocol for simulating artificial aging, this kind of test is never the same as what the restorations endure during use under clinical conditions. In other words, the complexity of the oral environment compared to the simplicity of laboratory procedures make extrapolation of the results from these tests to the real clinical situation very difficult. It is critical to define the expected clinical performance of any specific design or material by comparing the load at fracture in relation to the maximum bite forces. Therefore, it is not correct to consider a particular load, for instance, 1000 N, as an exact number, claiming that at 990 N the design will fail or at 1100 N the design will succeed even better as it is unknown exactly what the success limit is. It is not a linear relationship. As regards the measurements (reference values) of the maximum occlusal loads, the loads in this thesis that were considered acceptable might act as guidelines and approximate values for attaining a large safety margin. This is particularly true for ceramics, as they have a wide range of strength scatter due to flaws (brittle sensitive flaw populations).¹³⁹ So, if the material should be able to withstand 800 N, 1000 N may be sufficient for certain types of restorations to allow a sufficient margin of safety.^{140,141}

Efficacy of veneer design

The papers included in this thesis (I, II, IV) show the significant influence of the veneer layer on the load-bearing capacity of translucent zirconia restorations. The optimum thickness of porcelain veneering material used in bilayered zirconia restorations ordinarily range from 1.0 to 1.5 mm according to the manufacturer instructions with maximum 2 mm and minimum 0.8 mm. This range of thickness takes into consideration both the ability of porcelain to mask the underneath opaque zirconia material and the low strength and toughness of porcelain. In bilayered zirconia restorations, the veneer is a highly translucent layer making the aesthetic outcomes of the restoration greatly influenced by the light reflectance from the inner whitish zirconia layer. Even though the unfavourable influence of the decreased translucency of traditional zir-

conia material is compensated by the possibility of decreasing the zirconia core thickness to 0.5 mm as previous studies have suggested,¹⁴² high strength restorations with aesthetical outcomes are still needed. Reducing the zirconia thickness in favour of the porcelain layer has shown to give a deteriorating effect on fracture resistance of the restoration by providing less support for the porcelain layer.¹³² To overcome this problem, translucent zirconia materials, such as used in papers I and II, can be used in combination with the veneering porcelain, thus allowing for sufficient dimensions (thickness) and design. From an optical point of view, the recent generations of zirconia with enhancements in translucent properties may need only thinner layers of porcelain to achieve the optimum aesthetic results. The results of papers I and II on veneer and semi-monolithic designs point out that zirconia crowns designed with thin layers of porcelain veneering material, such as 0.3 and 0.5 mm in thickness, have higher load-bearing capacity compared to the crowns designed with thicker layers such as 2 and 2.5 mm. Those results were comparable to previous studies that showed less stress formation within the thin bilayered specimens made of porcelain and zirconia materials under slow-cooling conditions compared to the thick ones.^{95,102-104} Beside that the results of paper IV on stress distributions within framework designs also confirmed that partially veneered FDP with 0.3 mm porcelain veneer thickness has low maximum tensile principle stress levels in comparison to the FDP with 0.5 mm veneer thickness.

Summarizing the results of the thesis studies concerning veneer design, reducing the porcelain veneer thickness to 0.5 mm or even to 0.3 mm on translucent zirconia core provides adequate fracture strength compared to what might be expected during the clinical performance. In contrast, use of porcelain thicknesses beyond manufacturer recommendations resulted in crowns with even more decreased fracture strength. Table 11 shows by percentage how the load-bearing capacity is significantly reduced by altering the veneer thicknesses among the groups in paper I.

Table 11. Effect of porcelain veneering material designed with different thicknesses on the load-bearing capacity of zirconia crowns in paper I.

Groups	Thickness of porcelain veneer	Thickness of zirconia core	Overall thickness of crown	
Group 0.3	0.3 mm	0.5 mm	0.8 mm	78%
Group 0.5	0.5 mm	0.5 mm	1.0 mm	84%
Group 0.8	0.8 mm	0.5 mm	1.3 mm	85%
Group 1.0	1.0 mm	0.5 mm	1.5 mm	100%
Group 2.0	2.0 mm	0.5 mm	2.5 mm	59%
Group 2.5	2.5 mm	0.5 mm	3.0 mm	57%

NB: *1 mm-porcelain thickness acts as a control group (100%). Percentages represent the decrease of load-bearing capacity compared to the control group. Red measurements represent thicknesses exceeding the manufacturer’s recommendations. Green measurements represent thicknesses within the recommendations. Black measurements represent optimal thicknesses according to recommendations.

Efficacy of coping/framework design

The results of the thesis showed that it was not only the veneer thickness that played a role in determining the load-bearing capacity of translucent zirconia restorations. Papers II and IV on different semi-monolithic designs showed that the design of the coping/framework has great influence on the support of the veneering porcelain of partially veneered (semi-monolithic) zirconia restorations. The SCs with semi-monolithic designs, such as wave and cap designs, provide adequate support for the 0.5 mm veneering porcelain compared to the ones with the same veneer thickness without these semi-monolithic designs. This support was clear from the high load-bearing capacity of the SCs with those designs in paper II, and by the low principle tensile stress developed within the veneer layer of semi-monolithic FDPs in paper IV. The defensive means of such designs are simply aiming to protect the weak veneer either using underneath strong zirconia waves extending within the veneer layer. Or by extending the zirconia into occlusal surface to cover the facial veneer.

In paper II, the SCs with different semi-monolithic designs were artificially aged and loaded to fracture using a mechanical test. The differences in load at fracture of the SCs were significant. The findings of paper II were also confirmed in paper IV where the FDPs with the same semi-monolithic designs were analysed using FEM. In paper IV, the stress distribution within the FDPs with cap or wave designs showed generally low levels of maximum principle tensile stress within the 0.5 mm-veneering layer. By comparing these two designs, however, the findings were not in agreement in paper II and IV. In paper IV, the FDP with cap design showed favourable outcomes in terms of tensile stress compared to the FDP with wave design. These results are not in agreement with the findings of study II where the crowns with wave design showed higher load-bearing capacity under mechanical load to fracture test. This could be related to the fact that the strength of wave design depends mainly on minimizing the residual thermal stress which is commonly generated during the stages of fabricating the real crowns as were used in paper II. Previous studies have shown that the low thermal diffusivity of porcelain and zirconia materials in the bilayered design is behind the unstable behaviour of veneered zirconia compared to metal-ceramic reconstruction.^{95,98} These inherent thermal properties of ceramic materials (zirconia and porcelain) lead to the development of residual thermal stress within the restoration during cooling, leading to fracture at low loads. However, by using slow cooling protocol, the problem can be decreased as previous studies have shown.^{105,106} Regarding paper IV, however, the residual thermal stress within the restorations are important factors that were not taken into consideration to show the efficiency of the wave design, and the reason possibly behind the conflicting results of papers II and IV. On the other hand, the FDP with wave framework design showed the lowest shear stress at the zirconia-veneer interface compared to all semi-monolithic designs (including the cap design) that were tested in paper IV. The low shear stress indicates the effectiveness of wave design in reducing the adhesive fracture in the interface between the veneer and zirconia. Even so a majority of studies have shown that this kind of fracture mode rarely occurs compared to cohesive fracture.^{19-21,95,96} Beside

that the different restoration types used (SCs vs. FDPs) might also behind these opposing results in both paper II and IV regarding wave and cap designs.

Efficacy of shaping strategies at the connector area

Shaping of the connector area of an FDP can be incredibly challenging for many dental technicians. Paper III in this thesis focuses on the shaping methods in the connector area which are used mainly in dental laboratories to achieve a natural looking appearance of monolithic zirconia FDPs. The outcomes of this paper show that monolithic zirconia FDPs fabricated with different connector designs gained via different embrasure shaping methods have great impact on determining the load-bearing capacity. The default settings of the connector height and width, in the CAD program, were set so as the cross-sectional area was equal between all groups. The results confirmed that the fracture strength of monolithic zirconia FDPs decreases substantially as the sharpness of embrasures increased. As can be seen from (Table 12), the FDPs with sharp embrasures (either fabricated by sharp milling tools or shaped by a disk) reported significantly low load at fracture values with nearly 30-50% decreasing in strength compared to the FDPs in the control group. On the other hand, higher fracture strength can be achieved with the same cross-sectional area of the connector by shaping blunt embrasures. This might be related to the magnitude of stress generated with sharp design compared to blunt design.

Table 12. Effect of embrasure shaping methods on load-bearing capacity by percentage.

Control	Sharp embrasure	Blunt embrasure	Separation with disk (soft)	Separation with disk (fully-sintered)	Separation with porcelain	No occlusal embrasure
689	417	873	357	467	1038	934
100%	60%	126%	52%	68%	150%	135%

NB: All groups are compared to the control group (100%). Percentages represent an increase or decrease of load-bearing capacity of the FDPs from the control group (100%).

The results also show that the increased cross-sectional dimension gained by excluding the occlusal embrasure significantly increases the overall strength. These findings agree with other studies that emphasize the significance of sufficient connector dimensions in all directions (the height and the width) on fracture resistance of the FDP.^{92,94,111-115} Even though the connector dimensions should be sufficient in the direction of the load (the height) rather than the width, but it is never known in what direction the main load is. As the chewing pattern and cusp angles in the clinical situation often produce main load directions that are closer to horizontal rather than vertical. It is also worth mentioning that monolithic FDPs without the occlusal-incisal embrasure might be indicated only in the posterior area where aesthetic is less sensitive, as this design results in less aesthetic outcomes in the anterior region. On the other hand, an approximately 50% increase in load-bearing capacity compared to the control group was achieved using interdental porcelain separation at the connector area of the FDPs (Table 12). This design can combine the needed aesthetics in some clinical cases with strong performance of monolithic zirconia FDPs.

Although the results were not statistically different, paper III also indicated that using a disc for separation after milling at the white stage results in a lower load-bearing capacity compared to when separation is carried out at the fully sintered stage. This may then be explained by the fact that the white stage of zirconia is soft, fragile and sensitive to surface modifications before the final sintering. This fact should be kept in mind during the entire working procedure. At this stage, grinding is also tricky, as the operator may cause multiple scratches on the material surface if not steady of the hand. Such scratches might not be closed in the final sintering and consequently act as fractural impressions. It might, therefore, be possible to exert better control when making disc separations in the hard, fully sintered material, even though disc separations should be avoided completely.

Efficacy of the chosen zirconia material

Zirconia materials have become very popular, and with all new brands of modern dental zirconia that have different chemical for-

mulations and produced with varying processing techniques, it is crucial to understand that all zirconia materials are not identical. In choosing the right zirconia material, several factors should be taken into consideration such as the reputation of the manufacturer, excellent communication with the technician, the favourable research outcomes, and individual factors for each clinical case. Since the various clinical conditions require different material properties, it is reasonable that there will never be one zirconia type that combines high strength, toughness, and excellent aesthetics.

Paper II of this thesis sheds some light on the importance of the chosen type of zirconia material on the load-bearing capacity of restorations. Several brands of translucent zirconia are available nowadays on the market. However, it was shown in paper II that SCs made of high translucent zirconia material fail at low loads compared with ones made of translucent zirconia. These are accepted results since previous studies have shown the negative effect of the improved translucent properties of zirconia on its mechanical properties.^{26,59,64-69} Moreover, other studies also show how aging resistance properties decreased in high translucent zirconia materials.^{24,69,75,76} These results that have a material aspect can also be analysed differently. For instance, from a design aspect, SCs made of translucent zirconia with semi-monolithic designs interestingly showed better results compared with monolithic SCs made of high translucent zirconia crowns. Using a monolithic design, that is assumed to have a higher strength property with the latest high translucent zirconia, is not always favourable, especially in areas with high expected loads. The balance of the design and the material in semi-monolithic translucent zirconia SCs can compensate for the less strong mechanical properties of high translucent zirconia simultaneously with achieving excellent aesthetics with a buccal veneer. Using the former generation of translucent zirconia with slight improved translucent properties is also advantageous in terms of tooth conservation.⁵⁴ The thickness of the restoration made of this former generation of zirconia can be reduced to 0.5 mm with a reasonable clinical safety margin due to sufficient mechanical properties.¹⁴² This possibility conserves the tooth structure and increases the translucency as well. Pros and Cons of the latest

generations of high translucent zirconia need to be critically investigated in future studies.

Analysis of fracture mode

Brittle ceramic materials, such as used in this thesis (porcelain, translucent and high translucent zirconia), behave differently under load compared to ductile metals.^{139,143} While metals under load show preceding plastic deformation prior to fracture, ceramic materials fail almost directly when exceeding the elastic capacity of the material. This catastrophic behaviour of ceramic materials virtually acts as a map allowing one to trace back to fracture origin.¹⁴⁴ Information regarding the causes of the fracture of ceramic restorations can be gained by detecting the fracture origin and the direction of the propagating cracks within the material.

The primary goals of the fracture mode analysis in this thesis was first simply to show an efficient method for determining the fracture mode in terms of location of the fracture starting points and the paths of crack propagation. Second, it aimed to evaluate the ability of the test used for mimicking the clinical failures of dental restorations. Notwithstanding the relatively limited fractography method used in the papers of this thesis, this work offers valuable insights into fracture mode results.

The main findings of paper I, II, and III regarding fracture mode were in agreement with previous studies that were examining the clinical failure of zirconia restorations.^{19-21,95,96} They demonstrated that the most common fracture mode is the one that occurs cohesively within the veneering porcelain. Results of paper I and II on veneered and semi-monolithic SCs showed that cohesive failure rather than complete failure is the common fracture mode for the majority of the SCs under test. Only the SCs with monolithic and cap designs in paper II, and micro-veneered SCs in paper I fractured completely with zirconia core involvement but at high fracture loads. This might be explained that those designs with high load-bearing capacity were able to make the high strength zirconia material to be the only determinant for the fracture limit of the tested restorations. So, with only very high loads, the restoration

completely fractures with the cervical margin being the weakest point where complete crown fractures often start.

Previous studies on fracture mode also pointed out that the common fracture origin causing the failure of the FDPs is located at the gingival side of the connector area.^{94,112,114,145} This is related to the development of stress concentrations in the connector areas when lateral and vertical loads are applied to the FDPs. In agreement with previous studies, all the FDPs with different connector designs gained with varying methods of shaping, in paper III, fractured in the connector area. In paper III, monolithic zirconia FDPs serve as a beam which is subjected to both cyclic fatigue loads and loads to fracture. These loads are transmitted from the FDPs to the artificial abutments, generating high tensile stress in the connector area where critical flaw acts as fractural impression from which crack propagation occurs until catastrophic failure.

Aspects of methodology in this thesis

The laboratory tests used in this thesis were loaded to fracture mechanical test accompanied with artificial aging procedures and FEA test. Load to fracture mechanical test was used in paper I, II, and III to evaluate the load-bearing capacity of clinically relevant shaped specimens made of translucent zirconia materials. The FEA test was used in paper IV to estimate stress distribution and location within FDPs having different framework designs during simulated loads. It is noteworthy that the combination of both mechanical and computational laboratory tests in the four papers included has a great advantage in analysing the overall results. The predictive power of the results of the first three papers using the mechanical tests can be increased by using FEA as previous studies have shown.¹¹⁹ However, FEA alone does not show results dependent on production factors as for instance detrimental thermal residual stress during porcelain cooling.

Load to fracture test and artificial aging

In papers I, II, and III, static load to fracture test was performed after subjecting the specimens to a specific protocol of artificial aging. Preloading mechanical cycles, thermocycles, and water storage

are proposed in previous *in vitro* studies as commonly used methods for artificially aging the ceramic materials.^{25,112,146-152} It must be considered, however, that there is no consensus about a standardized protocol in which these aging tests are set-up and performed. Several studies have been published using different aging protocols proposed for testing the ceramic materials.¹⁴⁶⁻¹⁵² For instance, some studies used combinations of mechanical cycles, long-term water storage, and thermocycles to produce the fatigue effect on the ceramic materials.^{25,91,112,149,150} Others utilized an autoclave for accelerated aging using high temperatures reaching approximately 134°C.^{151,152} The latter could not be considered clinically relevant to the real oral condition. The fundamental aim is to subject the ceramic material to fatigue tests comparable to the oral environment to evaluate their probable influence on the different properties of the material. In this way, the performance of these materials under real clinical situations can be predicted from these simulated laboratory tests. Research in this area shows no consensus on which artificial age protocol resembles aging in the clinical situation. Besides, there has been some controversy about the ability of these kinds of tests to simulate the real complex clinical conditions or if they have an actual effect since some studies failed to show their deterioration effect.¹⁵³⁻¹⁵⁵

In the papers included in the present thesis, however, a specific artificial aging protocol has been chosen. This particular protocol concerning the number of cycles for both thermocycles and the preloading procedures is congruent with previous other studies conducted by the same research groups, enabling comparisons to be made with those studies.^{25,91,94,112} It was also decided to standardize the aging protocols to reduce the heterogeneity of the results and to permit comparison across the different studies of this thesis. Many clinical conditions have been taken into consideration when setting up the simulating laboratory tests in papers I, II, and III. In an attempt to mimic the mechanical and hydrothermal degradation effects on translucent zirconia restorations, preloading of the specimens was performed in an aqueous environment to mimic the physical process of chewing cycles that the restorations would be subjected to. The reason for a moist environment was to show ma-

terial behaviour during load in relation to stress corrosion and slow crack growth in the ceramic material tested, as was noted in previous studies.⁷²⁻⁷⁴ The weakening of ceramic strength in the aqueous environment has been proven to be related to stress corrosion mechanisms. It is related to the chemical reaction between water and ceramic at the crack tip area under load. The specimens were also subjected to thermocycling that might build up the stress levels within the restorations.

Load to fracture test and tested specimens

Flexural strength of ceramic materials as a property can be evaluated using standard uniaxial and biaxial tests.¹⁵⁶⁻¹⁵⁹ With regard to the shape of the tested specimens in uniaxial tests, such as three and four-point bending flexural tests, the specimen is a beam with specific dimensions that is loaded on the upper surface while it is supported on the lower surface. In biaxial tests, however, the load is applied with a piston (ball or ring instead) in the disk-shaped specimen that is supported by usually three rigid balls underneath.

The shapes of the specimens in papers I, II, and III were, however, norm crowns and bridges supported with artificial abutments rather than beam or discs as the standard flexural tests. The reason for this approach was due to the overall aim of this thesis which was to evaluate zirconia restorations with different designs rather than only evaluating material properties. So, clinically relevant specimens were primarily to be used in this thesis. Another reason of using this approach was to achieve more clinically relevant results. The teeth preparation principles for receiving fixed dental restorations made of zirconia were also applied to the supporting abutment models (cervical preparation, convergence angle, anatomical occlusal preparation, axial preparation) following the manufacturer's recommendations.

Concerning the tested materials, there is no uniformity in the materials used for producing the supporting abutment among the different laboratory studies. It can be made of the dental origin (human, bovine) or can be made of materials having mechanical properties close to dentin. The latter has been used in several laboratory investigations.^{25,91,112} Evidence from previous studies has established that the chosen abutment material affects the fracture load values

significantly.^{160,161} It has been shown that ceramic-based restorations appear to be subjected to loads that exceed their load-bearing capacity when they are supported by stiff materials (high modulus of elasticity), resulting in high, unrealistic fracture loads. Therefore, it is advantageous that the elastic properties of the artificial abutments are close to those of dentin, ranging from 10 to 15 GPa, to get relevant clinical fracture data. The second advantage is to avoid fracture of the abutment models during mechanical tests. In paper III, polyoxymethylene, with mechanical properties close to dentin, was used to fabricate abutment-like models for supporting the FDPs during preload and load to fracture tests. However, in paper I and II, the used abutment materials have a modulus of elasticity slightly lower than the recommended range (approximately 6 GPa). Nevertheless, the dies remained intact during cyclic preloads and load to fracture tests.

Finite element method (FEM)

The FEM is primarily a numerical method used to solve structural or non-structural problems utilizing an array of mathematical-based calculations.^{120,121} The method name comes from the fact that this method breaks down a larger and more complicated structure into smaller and more simple ones; the latter ones are called finite elements. The mathematical calculations that model these elements are solved and reassembled into a more extensive system of calculations that model the entire problem. Several factors influence the accuracy of the results of these calculations, such as type, arrangement, and number of elements of structure of interest.¹²²⁻¹²⁵

In general, the initial steps of the FEM of simulated loads start with creating a finite element model with appropriate mesh density, followed by identifying proper material properties for each component. In later steps, by setting up the loading conditions and boundaries according to the proposed assumptions, the model can be accurately tested. Paper IV of this thesis raises some important points for discussion regarding the effect of framework design on stress distribution within partially veneered semi-monolithic FDPs using 3D-FEA.

Methodological limitations

Paper I, II, and III in the present thesis were subject to several potential methodological weaknesses. The indenter used for cyclic and static mechanical tests is made of steel with high modulus of elasticity that might have played a role in the magnitude of stress in the tested specimens. Thus, it is not possible to translate the resulting load at fracture values directly to the clinical situation, where different contact counterparts with varying moduli of elasticity (such as teeth, crown materials, fillings, and food) are present. Moreover, for examining the fractured surfaces of the restorations, in this thesis, the tools used were the naked eye and, most importantly, the light microscope. However, sophisticated fractography using a scanning electron microscope (SEM) with systemized methodology and higher magnifications would provide more detailed information regarding the distinguishing characteristics of the fracture surfaces. Another limitation in this thesis was related to the small sample size used in the included papers I, II, III. With this regard, caution must be applied, as the findings might not be transferable to a large sample size even if the power analysis was based on previous studies where differences regarding the level of significance and standard deviation were detected among the zirconia-based specimens.^{25,94,111,112} In spite of these limitations, when analysing the results of paper I, II, III, the main findings certainly offer valuable insights for predicting clinical performance of translucent zirconia restorations.

Several limitations also need to be noted regarding the FEM used in paper IV. The complex oral environment might not be mimicked entirely in a computerized in vitro study. By considering this fundamental limitation, it is not possible to implement the results from FEA directly into the clinical situation, and supplementation with further clinical evaluation is required. In paper IV, during loading, stress analysis within the restorations was performed under static load conditions, even this is so far from reality where cyclic dynamic load conditions are expected. Furthermore, the properties of materials for the different components (veneer, zirconia, teeth, cement, and bone) were set as linearly elastic, isotropic, and homogenous, which are not actually representing the real situ-

ation, especially concerning ceramic components. Besides, modelling the geometry for the restorations and boundary conditions were based on different simplified assumptions. Such assumptions may impact the results of the study. Notably, the study used some simplified anatomical characteristics that were incorporated in the FEM models, such as the flat occlusal surfaces of the preparations and the anatomical occlusal surface of the fixed partial dentures (FPDs) with only two cusps (buccal and lingual). Also, both abutments (molar and premolar) present similar dimensions. While the need for those anatomical simplifications in FEM models is quite essential for the simplicity of modelling and simulation avoiding complex calculations, such alterations can undoubtedly generate a different pattern of stress distribution, particularly in the case of FPDs, which may not represent a real clinical scenario.

Clinical significance

This thesis has provided a more in-depth insight into the significance of the chosen design on the expected clinical performance of translucent zirconia restorations. The main factors which determine the selection of material and design for the restoration are aesthetics and strength. To fulfil these two factors, the clinician should use restorations with effective designs that are safe under extreme conditions and perform the required functions efficiently, while achieving pleasing aesthetics. It is quite logical that some clinical situations need more strength while others need more aesthetics, and a material might be needed that combines both strength and aesthetics. However, the different zirconia materials are not as aesthetic, or perhaps not as strong, as is required in the different situations; therefore, based on the various clinical indications, it is necessary to match the right material with the right design. Increasing the aesthetic properties, either by manipulating the microstructure of the material or by technical adjustments in design at a dental laboratory, affects the load-bearing capacity of those restorations. It is essential to have a full understanding of how to design restorations made of these materials in order to have excellent clinical service. This thesis shows that restorations made of translucent zirconia must meet specific design criteria to enhance

aesthetics while maintaining strength. The use of aesthetic veneering materials with minimum thicknesses upon translucent zirconia is advantageous when restorations made of these materials are indicated. The advantages are not limited to only improving the aesthetics and strength but also offer more conservative alternative options. Partially veneered restorations with semi-monolithic design showed in this thesis the importance of supporting the veneering materials with appropriate translucent zirconia core design and thickness. Using such designs resulted in high load-bearing capacity with expected favourable clinical performance. The thesis also favours using the former generation of translucent zirconia materials with semi-monolithic designs rather than high or ultra-high translucent zirconia with monolithic design in areas where high loads are expected, and the aesthetic result is important. The thesis emphasizes the importance of the design of the connector area of monolithic FDPs made of translucent zirconia. The use of the appropriate interdental separation methods at the connector area for matching the natural-looking appearance, is expected to achieve optimal aesthetics with favourable long-term performance of monolithic zirconia FDPs under clinical use. Such restorations combine the strength of monolithic zirconia material with the aesthetic enhancement of veneering materials.

Future perspectives

The overall outcomes of this thesis are expected to provide a sound understanding of some design aspects that considerably influence the load-bearing capacity and aesthetics of translucent zirconia restorations, inspiring future research into their clinical performance. In this regard, large randomized controlled clinical trials could provide more definitive evidence of the effectiveness of the proposed designs in this thesis. In addition, there are still other factors that remain a question concerning the aging resistance, colour stability, wear resistance and mechanical strength of several new translucent zirconia materials. Since some or all the transformation toughening mechanism in these new materials has been lost, it should be considered that not only has strength been affected, but also toughness. And, since strength and toughness over time in

clinical use are two different things, further investigations are necessary. Many more studies on design are required, and the need to understand the zirconia materials both from design and material aspects is presented as an essential step in bridging the knowledge gap between basic and applied material sciences.

CONCLUSIONS

Based on the results gained from the four papers included in this thesis, the following conclusions on design of translucent zirconia restorations were withdrawn:

Translucent zirconium-dioxide crowns can be veneered with minimal thickness layer of 0.5 mm porcelain without showing significantly reduced fracture strength compared to traditionally veneered (1.0–2.0 mm) crowns. (*Study I*)

Fracture strength of micro-veneered crowns with a layer of porcelain (0.3 mm) is lower than that of traditionally veneered crowns but still within range of what may be considered clinically sufficient.
(*Study I*)

Porcelain layers of 2.0 mm or thicker should be used where expected loads are low only. (*Study I*)

Translucent and high-translucent zirconia crowns might be used in combination with 0.3 mm microcoating porcelain layer with semi-monolithic design to enhance the esthetic properties of restorations without significantly decreasing fracture resistance of the crowns.
(*Study II*)

If a 0.5 mm porcelain layer is needed for a semi-monolithic crown, wave design or cap design might be used to increase fracture

resistance. In both cases, fracture resistance gained is likely to be clinically sufficient as the registered fracture loads were high in relation to expected loads under clinical use. (*Study II*)

Sharp embrasures and interproximal separations made with diamond discs considerably decrease the load-bearing capacity of monolithic zirconia FDPs compared to FDPs made with blunt embrasures. (*Study III*)

Blunt embrasures in combination with localized porcelain build-up produce FDPs with even higher load-bearing capacity. (*Study III*)

Great care should be taken when performing interproximal separations to avoid decreasing the overall strength of monolithic zirconia FDPs. (*Study III*)

Framework and veneer designs play a significant role in the stress distribution of the partially veneered zirconia FDPs under loading. The FDP with cap design of zirconia framework minimize the maximum principle stress in the 0.5 mm veneering porcelain. (*Study IV*)

The FDP with a 0.3 mm veneering porcelain has low maximum principle tensile stress in the veneering porcelain but with high maximum shear stress at the zirconia-veneer interface. (*Study IV*)

The FDP with wave design of zirconia framework minimize the maximum shear stress. (*Study IV*)

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PAPERS I-IV

I

Fracture strength of veneered translucent zirconium dioxide crowns with different porcelain thicknesses

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ABSTRACT

Objective: To evaluate fracture strength of veneered translucent zirconium dioxide crowns designed with different porcelain layer thicknesses.

Materials and Methods: Sixty crowns, divided into six groups of 10, were used in this study. Groups were divided according to different thicknesses of porcelain veneer on translucent zirconium dioxide cores of equal thickness (0.5 mm). Porcelain thicknesses were 2.5, 2.0, 1.0, 0.8, 0.5 and 0.3 mm. Crowns were artificially aged before loaded to fracture. Determination of fracture mode was performed using light microscope.

Results: Group 1.0 mm showed significantly ($p \leq .05$) highest fracture loads (mean 1540 N) in comparison with groups 2.5, 2.0 and 0.3 mm (mean 851, 910 and 1202 N). There was no significant difference ($p > .05$) in fracture loads among groups 1.0, 0.8 and 0.5 mm (mean 1540, 1313 and 1286 N). There were significantly ($p \leq .05$) more complete fractures in group 0.3 mm compared to all other groups which presented mainly cohesive fractures.

Conclusions: Translucent zirconium dioxide crowns can be veneered with minimal thickness layer of 0.5 mm porcelain without showing significantly reduced fracture strength compared to traditionally veneered (1.0–2.0 mm) crowns. Fracture strength of micro-veneered crowns with a layer of porcelain (0.3 mm) is lower than that of traditionally veneered crowns but still within range of what may be considered clinically sufficient. Porcelain layers of 2.0 mm or thicker should be used where expected loads are low only.

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Introduction

Yttria-stabilized tetragonal zirconia polycrystal (Y-TZP), used as core material for all-ceramic dental restorations, is considered to be a reliable crown and bridge material for restoring patients' oral functions, including esthetics [1,2]. The unique transformation toughening properties of Y-TZP give the material high mechanical strength and toughness which is one of the reasons that the material is often described as the ideal substitute for metal alloys for cores and frameworks in veneered restorations [3,4].

Despite the excellent mechanical properties of Y-TZP, superficial chip-off fractures in the veneering porcelain have been reported as a common issue and, furthermore, the most common reason for the clinical failure of veneered Y-TZP [1,5]. The mechanisms behind the chipping of veneered Y-TZP have been discussed in the literature. Studies have suggested

various reasons, such as poor fracture toughness of the veneer materials [6], inadequate bond strength between the veneer ceramic and the Y-TZP core [7], inadequate core design in relation to veneer thickness [8–10], mismatch of coefficients of thermal expansion between the veneer ceramic and the Y-TZP core [7,11], and finally problems relating to the build-up of residual stresses in the veneer material during firing due to improper cooling [12,13].

Several methods for preventing chip-off fracture of veneered Y-TZP restorations have been described in the literature, such as improving technical procedures [14,15] and creating different stress breaking designs [8,9,16] in addition to producing materials with high standard processing and improved mechanical properties. A reliable way to avoid chip-off fractures is to fabricate monolithic restorations made entirely from translucent zirconium dioxide material with

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Table 1. Porcelain layer thicknesses in the different groups.

Groups	Thickness of porcelain layer	Thickness of zirconium dioxide core	Overall thickness of crown (preparation depth)
Group 2.5	2.5 mm ^c	0.5 mm	3.0 mm
Group 2.0	2.0 mm ^b	0.5 mm	2.5 mm
Group 1.0	1.0 mm ^a	0.5 mm	1.5 mm
Group 0.8	0.8 mm ^b	0.5 mm	1.3 mm
Group 0.5	0.5 mm ^c	0.5 mm	1.0 mm
Group 0.3	0.3 mm ^c	0.5 mm	0.8 mm

^aOptimal thickness according to the manufacturer's recommendations,

^bthickness within the recommendations and

^cthickness exceeding the recommendations.

equivalent strength compared to traditional Y-TZP [17]. However, this new direction away from bi-layered design toward monolithic restorations does not always completely eliminate the need to veneer translucent zirconium dioxide with porcelain to enhance the esthetic result when needed. By adding a thin layer of porcelain to translucent zirconium dioxide, it might be possible to improve the esthetic appearance of monolithic restorations [18,19]. This design, however, is not within the manufacturer's recommendations and there is no recommendation as to how thin the porcelain layer can be made without affecting fracture strength of such micro-veneered translucent zirconium dioxide restorations.

According to manufacturer's recommendations, the optimal thickness of veneering porcelain on a Y-TZP core should be 1.0–2.0 mm. In a mathematical analysis study, however, it was suggested that within a combination of two laminated materials, such as porcelain and Y-TZP, a reduced overall thickness of the combined materials lead to less detrimental stress formation in the porcelain after firing of the material [12]. The results were confirmed in another finite element analysis (FEA) study where spherical specimens were used [13]. Since the clinically weak point of veneered Y-TZP restorations is the porcelain layer [6], the effect of reduced porcelain thickness might play a major role in fracture strength of the crowns. With that in mind, and to investigate if those findings are applicable on fracture strength of porcelain-veneered translucent zirconium dioxide crowns, a study evaluating the design of anatomically shaped crowns with minimum porcelain layer thicknesses is motivated. The preparation depth of the abutment should then be considered, since the overall thickness of a crown calls for a corresponding tooth preparation. Consequently, a thinner crown allows for a less aggressive tooth-preparation, which might have an impact on the prognosis of the tooth-crown complex [20].

Therefore, the aim of this study was to investigate fracture strength of veneered translucent zirconium dioxide crowns designed with different porcelain

layer thicknesses. The null hypothesis was that the veneer thickness does not affect fracture strength of veneered translucent zirconium dioxide crowns.

Materials and methods

Experimental design

A total of 60 standardized translucent zirconium dioxide crowns, divided into six groups of 10, were used in this study (Table 1). The six groups were divided according to different thicknesses of the porcelain layer on translucent zirconium dioxide cores of equal thickness (0.5 mm). The thicknesses of porcelain veneer were 2.5, 2.0, 1.0, 0.8, 0.5 and 0.3 mm. The gradual reduction in porcelain thicknesses, from group 2.5 to group 0.3, resulted in a corresponding decrease in preparation depths of the abutment dies without changing the outer dimensions of any of the crowns (Figure 1).

Design and production of translucent zirconium dioxide cores and abutment dies

A metal die resembling a molar crown preparation was made for an all-ceramic crown with a 1.2 mm deep 120° cervical chamfer and 15° angle of convergence. The die was replicated using a silicone impression material (President[®], Coltene AG, Altstätten, Switzerland). The impression was poured with die stone material (Vel-Mix, Kerr Corporation, Orange, CA) to create the master die. By using a double scan technique, the master die was scanned following scanning of a wax-up of a full anatomic crown with an occlusal thickness of 3.0 mm and an axial thickness of 3.0 mm. The scanning was carried out with a dental laboratory scanner (D900L, 3Shape, Copenhagen, Denmark) and the data were then transferred to a computer running computer-aided design (CAD) software (Autodesk, San Rafael, CA). By using the CAD software, the first crown/die complex was copied and adjusted virtually by decreasing the preparation depths and correspondingly reducing the crown thickness

Figure 1. An illustration showing the CAD files of the different groups with overall crown thicknesses (core/porcelain). All crowns had an equal core thickness but different porcelain thicknesses. The gradual reduction in crown thicknesses resulted in a corresponding decrease in preparation depths without changing the outer dimensions (the X-imaginary line) of any of the crowns. The red dotted line denotes the core/porcelain interface.

toward the dimension settings of the different six groups. In this way, the outer dimensions of the crowns in all groups were identical, but the preparation depths of the abutment dies differed to match the different crown thicknesses (Figure 1).

Six different CAD files were thus produced and subsequently sent to a certified BruxZir milling center (Cosmodent AB, Malmö, Sweden) where they were used to produce 60 translucent zirconium dioxide cores of equal 0.5 mm-thickness (BruxZir[®] Solid zirconia, Glidewell dental laboratories, San Diego, CA). The CAD data were also used to produce 60 dies in six different preparation depths corresponding to the different six groups. All abutment dies were milled from a polymer material (Polyoxymethylene, Nordbergs Tekniska AB, Stockholm, Sweden).

Veneer build-up

The porcelain was built up by hand, using a technique that has been described in detail in previous

studies [21,22]. The zirconium dioxide cores were veneered with a porcelain recommended by the manufacturer (IPS e.max Ceram, Ivoclar Vivadent, Schaan, Liechtenstein). A custom-made knife (Figure 2) was used to standardize the outer shape and dimension of the crowns. Each crown underwent five porcelain firing cycles: liner, wash, dentin 1 and 2 and finally glaze firing. The firing cycles were carried out in a calibrated furnace (Ivoclar P 500, Ivoclar Vivadent, Schaan, Liechtenstein) according to the manufacturer's recommendations.

Artificial aging

All crowns underwent three stages of artificial aging: thermocycling, storing in a 37 °C moist environment and cyclic preload.

Thermocycling

In the first stage of artificial aging, all crowns underwent 10,000 thermocycles in two water baths of a

Figure 2. A custom-made knife used for the porcelain build-up, with a zirconia core mounted on the abutment die.

specially built thermocycling device at temperatures of 5 and 55 °C. Each cycle lasted for 60 s; 20 s in each bath and 10 s for transfer between the baths.

Cementation and storing in an incubator

The 60 crowns were cemented to the respective abutment dies using a methacryloyloxydecyl dihydrogen phosphate (MDP) containing resin cement Panavia F version 2.0 (Kuraray Medical Inc., Okayama, Japan) according to the manufacturer's recommendations. The inner surfaces of the crowns were sandblasted using 50 mm aluminum oxide at a pressure of 2 bar, and then they were rinsed with water according to the manufacturer's recommendations and as was suggested in a previous study by Kern et al. [23] Before crown cementation, the abutment dies were roughened with air-abrasion (Basic Quattro IS, Renfert GmbH, Hilzingen, Germany) using 110 mm aluminum oxide at a pressure of 2 bar from a distance of 1.0 cm at a 90° angle to the abutment surface. Then they were steam cleaned and subsequently treated with ED PRIMER II A and B (Kuraray Medical Inc., Okayama, Japan). The crowns were cemented to the abutment dies with a standardized seating load of 15 N in the direction of insertion.

After cementation, the specimens were stored in a plastic container inside an incubator (Memmert Incubator, Memmert, GmbH, Schwabach, Germany) at a temperature of 37 °C for 60 d. All specimens were wrapped in a wet paper to create a moist environment like the oral cavity and to prevent desiccation of the luting cement.

Cyclic preload

In the final stage of artificial aging, cyclic preload, a specially made preloading device was used

Figure 3. An illustration of the test set-up used for both the cyclic preload and the load to fracture. NB: The crowns were submerged in water, and a plastic foil was used to spread the load evenly over the loaded surfaces.

(MTI Engineering AB/Pamaco AB, Malmö, Sweden) to subject the specimens to 10,000 load cycles between 30 and 300 N, with a load frequency of 1 Hz and with the specimens submerged in distilled water at 10° of inclination toward the tooth axis (Figure 3). Preload was applied with a 4.0 mm Ø stainless steel ball placed on the occlusal surface of the crowns. A 0.2 mm thick plastic foil (PE-Baufolie, Probau, Bauhaus, Zug, Switzerland) was placed between the ball and the crown to distribute the load evenly and to prevent cone-crack formation as was suggested by Kelly JR [24].

Load to fracture

All specimens were mounted in a test jig at 10° inclination, in the same way as during preloading, and as was used in previous laboratory studies [9,17,21]. Specimens were loaded to fracture using a universal testing machine (Instron 4465, Instron Co. Ltd, Norwood, MA). The load was applied on the occlusal surface of the crowns with a 4.0 mm Ø stainless steel ball and a 0.2 mm thick plastic foil inserted between the ball and the crown. The crosshead speed was 0.255 mm/min, and the fracture was defined as a visible crack, load drop or an acoustic event, whichever occurred first. The acoustic event was determined by hearing a cracking sound that is accompanied by the changes in the load-path graph on a computer that is connected to the testing machine.

Analysis of fracture mode

Examination of the fracture surfaces and determination of fracture mode (complete, cohesive or adhesive fractures) were performed by a gross visual and microscopic (Leica DFC 420, Leica Microsystems

Table 2. Load at fracture in Newton.

Groups	Group 2.5	Group 2.0	Group 1.0	Group 0.8	Group 0.5	Group 0.3
Specimen no.						
1	882	946	1402	1277	1138	1042
2	1064	1009	1355	1512	1664 [✓]	1023 [✓]
3	803	870	1727	1313	1483 [✓]	1205 [✓]
4	701	750	1256	1154	1137	1226 [✓]
5	867	1203	1430	1050	1326	1154 [✓]
6	848	805	1773	1748 [✓]	1039	1397 [✓]
7	863	969	1485	1076	1273	1086
8	903	816	1828	1297	1066	1297 [✓]
9	843	910	2118 [✓]	1639	1475	1310 [✓]
10	732	822	1030	1063	1263	1276 [✓]
Mean*	851 ^a	910 ^a	1540 ^b	1313 ^{b,c}	1286 ^{b,c}	1202 ^c
Std. deviation	98	131	318.9	247	203.7	123.9
Fracture mode (complete/cohesive)	0/10	0/10	1/9	1/9	2/8	7/3

*Means with the same letters in superscript (denoted a, b, c) did not show any significant difference in fracture load ($p > .05$). [✓]Complete fracture mode.

CMS GmbH, Wetzlar, Germany) assessment by one investigator.

Statistical analysis

Loads at fracture were recorded and one-way ANOVA, Tukey's test was used to analyze differences in fracture loads among the groups, IBM SPSS Statistics version 22 (SPSS Inc., Chicago, IL). The differences in fracture mode were analyzed using Fishers' Exact Probability test. The level of significance was set to p values $\leq .05$. All statistical analysis was done by an experienced professional statistician.

Results

All specimens were visually intact through all steps of the artificial aging, and neither fractures nor crack formations were detected in the crowns or the abutment dies. Loads at fracture, levels of significance and fracture modes are summarized in (Table 2).

Fracture load

The one-way ANOVA showed significant differences in fracture loads among the groups. Group 1.0 showed significantly ($p \leq .05$) highest fracture loads (mean 1540 N) in comparison with groups 2.5, 2.0 and 0.3 (mean 851, 910 and 1202 N).

There was no significant difference ($p > .05$) in fracture loads among groups 1.0, 0.8 and 0.5 (mean 1540, 1313 and 1286 N). Groups 2.0 and 2.5 showed significantly lower fracture loads in comparison with all other groups but with no significant differences between them ($p > .05$).

Fracture mode

There were significant differences in the way the crowns fractured. Two kinds of fracture mode were

seen visually, either complete fracture involving both the veneer and the core or cohesive fracture denoting fracture within the veneer only, i.e. chipping fracture (Figure 4). No adhesive fractures, involving the bonding area between the veneer and the core, were observed in the crowns under the light microscope. All crowns in groups 2.5 and 2.0 and more than 80% of the crowns in groups 1.0, 0.8 and 0.5 showed cohesive fracture mode. Conversely, there were significantly ($p \leq .05$) more complete fractures in group 0.3 compared to all other groups.

Discussion

Despite the fact that recently developed zirconium dioxide materials are more translucent than traditional Y-TZP, the improvements may be esthetically insufficient for use without adding porcelain. From a mechanical point of view, however, it has been suggested that the veneering porcelain should be kept to a minimum to avoid chip-off fractures commonly seen on traditionally veneered reconstructions based on Y-TZP [1,5]. The optimal design for such translucent reconstructions, however, is still an open question. The results of this study suggest that changes in the thickness of the porcelain layer have a significant influence on fracture strength of translucent zirconium dioxide crowns. Thus, the null hypothesis of the study was rejected.

The porcelain veneer thickness in the range of 1.0–2.0 mm on the traditional Y-TZP core is commonly recommended by the manufacturers to protect porcelain against fracturing and to achieve a clinically acceptable esthetic appearance by masking the underlying white opaque Y-TZP. The results of this study showed that the reduction of the porcelain layer, from 1.0 to 0.5 mm, lead to numerically but not significantly decreased loads at fracture, still being within

Figure 4. Example of the two fracture modes: (A) Complete fracture (B) Cohesive fracture.

the range of what might be considered clinically sufficient. Of 1000 N or more have been suggested to be an adequate safety limit for long-term durability of the crowns for most patients in all areas of the mouth [24–26]. At porcelain thickness of 0.3 mm only, the difference in fracture loads became significant, but the loads the crowns endure before fracture are still within what is considered clinically sufficient. Consequently, all crowns with porcelain thickness thinner than recommended passed the safety limit and showed satisfactory results regarding fracture load. Most crowns with a porcelain thickness of 2.5 or 2.0 mm, on the other hand, showed fracture loads less than 1000 N. One explanation may be that the strength of a ceramic material is greatly influenced by the volume of material under stress. The increased volume presents more flaw populations including different flaw sizes and shapes, with a higher probability that a critical flaw would be present where the highest stresses are concentrated during loading. Consequently, this results in lower strength. Another possible explanation is that thicker layers of porcelain receive less support from the core than thinner ones, for geometrical reasons, and consequently become subjected to greater tensile forces that are especially detrimental to the brittle porcelain. Furthermore, all crowns of these two groups 2.5, 2.0 showed cohesive fractures in the porcelain, leaving the zirconia core visually intact. This result is consistent with studies showing an increased risk of cohesive porcelain fractures where the zirconia core is die-shaped rather than anatomically shaped, leaving the porcelain unsupported in thick layers [8,9,16].

The results of this laboratory study showed that the most common fracture mode was similar to those seen when investigating clinical failures, cohesive fractures involving fracture of thin flakes of porcelain within the surface of the veneer only [1,5]. This fracture mode is quite different from so-called cone cracks, sometimes seen in laboratory studies. This implies that the loading of the anatomically shaped specimens was adequate. Further investigations using fractography, however, might give detailed information on fracture mode.

Previous studies have attempted to classify cohesive fractures by their size and how they affect function and what kind of intervention is needed if any [27,28]. In this study, further analysis of the severity of cohesive fractures was not possible as the observation of how a chip affects the function and the decision on whether an intervention is needed or not, depends for instance on neighbor teeth, antagonist as well as patient expectations.

The crowns with 0.3 mm porcelain showed, however, a majority of complete fractures, suggesting that since the porcelain layer was so thin, no tensile stresses are built up within this thin layer. Thus, the full impact of the load is transferred to the core, making it the limiting factor for the strength of the crowns rather than the porcelain. In this study, the zirconia cores were only 0.5 mm thick, which is the thinnest dimension recommended for this type of crown according to the manufacturers. This recommendation was further supported in a study by Nakamura et al. [29]. If the core had been thicker, the loads at fracture might have been substantially higher,

even for the groups with the thinnest porcelain. As was reported by Wakabayashi et al., fracture resistance of veneered alumina disks increases and fracture sites shifts from the veneer to the core as the core/veneer ratio increases [30]. Although that study was investigating a different core material with a different methodology compared to that of this study, the results of this study support the findings of that previous study. With that in mind and to gain increased overall strength of porcelain-veneered translucent zirconium dioxide crowns, it would be interesting to study the combination of thicker cores and a 0.3 mm micro-veneering layer of porcelain in future research projects.

Previous studies have investigated the role of the overall thickness (core/veneer) on residual stress formation in porcelain-veneered zirconia using mathematical analysis of a bilayer model and FEA of spherically shaped specimens [12,13]. This study confirmed previous results but also added information based on clinically complex shaped crowns on tooth-like abutments. It was shown in previous studies that a thinner overall thickness of laminated specimens results in lower levels of stress formation during cooling of the material after firing. Since a thinner overall thickness in this study also was followed by a thinner porcelain layer, the higher fracture loads were probably dependent on lower stress formations in the thin porcelain compared to thicker porcelain. During the cooling phase after firing, the surface portion of a thicker porcelain layer will be cooled faster than the bulk material close to the zirconia core, since the zirconium dioxide material has extremely low thermal conductivity compared to a metal coping [12]. If the surface portion of the porcelain is solidified before the bulk material, then stress formations will occur when the bulk continues to shrink after solidification of the surface layer, resulting in a residual stress zone prone to cohesive chip-off fracture under the surface layer of the porcelain. The probability of such residual stresses might be lower the thinner the porcelain is. Although slow-cooling protocols are used today to avoid the problem with residual stresses, thinner layers might contribute to lowering these residual stresses during cooling. Further studies, using FEA for instance, are needed to be performed to confirm this explanation.

It was, however, shown in a previous study that fracture strength of laminated dental ceramics is not sensitive to the core/veneer thickness ratio [31]. That study suggested instead increasing the crown strength by adding thicker porcelain layers, keeping the core thickness as thin as possible. Those conflicting results

might be explained by that a different core material was used in that study (lithium disilicate) and the total thickness for the specimens tested was 1.5 mm, with the thickest porcelain layer thickness of 1 mm only. Hence, the core material used in that study is not associated with the cooling problems seen with zirconium dioxide, and the veneer thickness was not exceeding the optimal porcelain thickness of 1 mm.

The preparation of abutment dies in this study was made with an anatomical occlusal shape to mimic the clinical situation. The preparation design used, with a cervical chamfer, was the same for all groups in accordance with the manufacturer's recommendations. To obtain mechanical support comparable to that in real clinical situations, all crowns were cemented onto abutment dies made of a material with mechanical properties and a modulus of elasticity close to dentin. In that way, the risk of unrealistically high fracture loads of the crowns or fracture of abutments would be reduced as was investigated in previous studies [32–35]. Previous studies have shown the deleterious impact of the laboratory thermocycling on the integrity of bonding between restorative materials and cement [36,37]. Therefore, the crowns were cemented using a recommended MDP containing resin cement after thermocycling stage to prevent partly loose crowns during the pre-load and load to fracture tests.

In this study, the proper thickness of porcelain veneer was built-up manually on the cores by one operator. It is well known that manually built-up veneers are never completely identical between the crowns because the ratio of porcelain powder and liquid plus the compaction process leads to different shrinkage and flaw populations. Consequently, the strength will always differ to some extent between individual specimens. The over-press veneering technique might provide more standardization in the veneering process in comparison with manually build-up technique. However, using manually build-up veneering technique is the dominating technique in dental labs compared to the over-press technique, the latter uses quite different ceramic materials (glass or hybrid) than commonly used feldspathic porcelain. So, manually build-up veneering technique is a preferable method based on the used materials in this study. Furthermore, there is no absolute consensus on the reliability of the over-press technique [38–40]. When the veneers are fired under controlled laboratory conditions, in using manually build-up technique, any differences will have only minor effects on fracture strength of the crowns. In the study, the proportions of porcelain powder and liquid were measured and standardized within the crowns of

the same group and between different groups according to the manufacturer's recommendations.

Different protocols for the artificial aging procedures have been suggested in previous studies [41,42]. Fatigue protocols aim to simulate the various complex forces ceramic restorations would be subject to in a real clinical situation. It is well known that all-ceramic restorations are prone to continuous slow crack growth as a result of low continuous cyclic loads in a wet environment, resulting in stress corrosion at the crack tips of critical flaws. Thermocycling was used to mimic the influence of stresses that develop during the function, while the possible effects of the physical process of chewing were produced by cyclic preload with the specimens submerged in water. Nevertheless, there is no clear evidence on what aging protocol that resembles the aging in the clinical situation. Furthermore, there is no consensus regarding the effectiveness of these aging tests as some fail to show a direct relationship between the fatigue procedures and fracture loads of dental ceramics [43]. In this study, however, thermocycling and preload tests were adopted from previous studies conducted by the same research group to allow for comparison of the results of these studies [9,17,32].

In this study, static load to fracture test was used to assess fracture strength of the crowns. The crowns were mounted with a 10° of inclination relative to the load direction. This angle of inclination has been used in many previous studies and was initially suggested by Yoshinari and Derand [9,17,21]. A more eccentric loading angle might have been favorable to evaluate fracture strength of the crowns. However, the test set up was adopted from previous studies to make comparisons possible.

There are always limitations in all laboratory studies. A full clinical scenario in respect of the performance of restorations can be seen in a clinical setting only. It is hard to extrapolate clinical results from the static loading test alone. Therefore, the results of this study are needed to be confirmed clinically.

Conclusions

Within the limitations of this laboratory study, the following conclusions can be drawn: translucent zirconia crowns can be veneered with minimal thickness layer of 0.5 mm porcelain without showing significantly reduced fracture strength compared to traditionally veneered (1.0–2.0 mm) crowns. Fracture strength of micro-veneered crowns with a layer of porcelain (0.3 mm) is lower than that of traditionally veneered crowns but still within range of what may

be considered clinically sufficient. Porcelain layers of 2.0 mm or thicker on translucent zirconium dioxide result in significantly decreased fracture strength of the crowns compared to 1.0 mm or thinner and should, therefore, be used where the expected loads are low only.

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Disclosure statement

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II

Effect of different semimonolithic designs on fracture resistance and fracture mode of translucent and high-translucent zirconia crowns

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Purpose: The aim of this study was to describe different designs of semimonolithic crowns made of translucent and high-translucent zirconia materials and to evaluate the effect on fracture resistance and fracture mode.

Methods: One hundred crowns with different designs were produced and divided into five groups (n=20): monolithic (M), partially veneered monolithic (semimonolithic) with 0.3 mm buccal veneer (SM0.3), semimonolithic with 0.5 mm buccal veneer (SM0.5), semimonolithic with 0.5 mm buccal veneer supported by wave design (SMW), and semimonolithic with 0.5 mm buccal veneer supported by occlusal cap design (SMC). Each group was divided into two subgroups (n=10) according to the materials used, translucent and high-translucent zirconia. All crowns underwent artificial aging before loading until fracture. Fracture mode analysis was performed. Fracture loads and fracture modes were analyzed using two-way ANOVA and Fisher's exact probability tests ($P \leq 0.05$).

Results: SM0.3 design showed highest fracture loads with no significant difference compared to M and SMW designs ($P > 0.05$). SM0.5 design showed lower fracture loads compared to SMW and SWC designs. Crowns made of translucent zirconia showed higher fracture loads compared to those made of high-translucent zirconia. M, SM0.3, and all but one of the SMC crowns showed complete fractures with significant differences in fracture mode compared to SMW and SM0.5 crowns with cohesive veneer fractures ($P \leq 0.05$).

Conclusion: Translucent and high-translucent zirconia crowns might be used in combination with 0.3 mm microcoating porcelain layer with semimonolithic design to enhance the esthetic properties of restorations without significantly decreasing fracture resistance of the crowns. If 0.5 mm porcelain layer is needed for a semimonolithic crown, wave design or cap design might be used to increase fracture resistance. In both cases, fracture resistance gained is likely to be clinically sufficient as the registered fracture loads were high in relation to expected loads under clinical use.

Keywords: monolithic zirconia, porcelain veneer, Y-TZP

Introduction

Yttrium oxide-stabilized tetragonal zirconia polycrystal material (Y-TZP) has become accepted over the past decades as a biocompatible restorative material,¹ often considered to be the material of choice for many conditions in prosthetic dentistry.² Generally, all ceramic systems based on Y-TZP can be found in the following two major designs: traditional bi-layered design and a more recent monolithic design.

Bi-layered Y-TZP design is relatively well documented with respect to both clinical follow-up studies and laboratory studies.³⁻⁶ However, several clinical studies have

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51

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shown that despite the excellent mechanical properties of Y-TZP, superficial chip-off fracture in the veneering ceramic is a common clinical problem for this type of design.³⁻⁵ The reason for veneering Y-TZP restorations, in the traditional design, is that the optical properties of the material itself are poor,^{2,7} and the material needs to be veneered with a highly esthetic ceramic material to achieve an acceptable resemblance to natural tooth structure. Low-strength vitreous ceramics, such as porcelain, are then fused to the high strength Y-TZP core material to combine the strength with better esthetic outcomes.⁷

Monolithic design, on the other hand, has been increasingly popular over the last decade since it is believed to withstand high loads during function without the risk of chip-off fractures of veneering materials. To overcome the poor optical properties of the traditional Y-TZP material, translucent zirconia materials were developed for monolithic restorations,⁸ often referred to as monolithic translucent zirconia. The first generation of translucent zirconia materials had mechanical properties comparable to the traditional Y-TZP with promising laboratory and short-term clinical studies.⁹⁻¹²

Recently, there were notable attempts for making high-translucent monolithic zirconia to be used in highly esthetic clinical cases. Nevertheless, the translucency that can be gained without tampering with the strength of the material is very limited, and it is not possible to gain translucency above a certain level without losing the mechanical properties of the material to some extent.^{8,13,14} This new generation of high-translucent zirconia materials with scarce clinical data might be considered to be used with caution due to their low mechanical properties and doubtful aging stability.^{8,13,14} Therefore, it would be desirable to maintain the strength of the translucent zirconia materials and still gain better esthetic outcomes without the risks that are associated with full veneering or the increased translucency of high-translucent zirconia.

Partially veneered monolithic (semimonolithic) crown design with porcelain coverage on the buccal area only might be a feasible method to overcome both the issues with chip-off fractures and the poor esthetics of full monolithic zirconia crowns. In this way, it might be possible to exclude the porcelain on surfaces subjected to wear and high loads such as the occlusal tables and the marginal ridges. Such a design has become common today but is not always sanctioned in the manufacturers' recommendations for the materials and is consequently used without the support from manufacturers' instructions.^{15,16} Scientific studies on this subject are lacking.

Different alternative semimonolithic crown designs are, therefore, proposed and investigated in this study. They can be described as microcoating design, reduced veneer design, wave support design, and occlusal cap-like projection design; all are suggested to enhance the esthetics of the restoration without decreasing the overall strength. It has not yet been evaluated, however, which of the proposed designs that are to be recommended, or if the designs add strength to the crowns compared to monolithic ones.

Hence, the aim of the present study was to describe different semimonolithic crown designs for translucent and high-translucent zirconia materials and to evaluate their fracture resistance and fracture mode under the null hypothesis that all designs for both materials included would perform equally.

Methods

Study design

A total of 100 standardized crowns with different designs were produced and divided into five groups ($n=20$) (Figure 1): monolithic (M), partially veneered monolithic (semimonolithic) with 0.3 mm microcoated buccal veneer (SM0.3), semimonolithic with 0.5 mm buccal veneer (SM0.5), semimonolithic with 0.5 mm buccal veneer supported by wave design (SMW), and semimonolithic with 0.5 mm buccal veneer supported by occlusal cap design (SMC).

Each group was divided into two subgroups ($n=10$) according to the crown materials used, translucent zirconia (BruxZir® Solid Zirconia; Glidewell Dental Laboratories, Newport Beach, CA, USA) and high-translucent zirconia (Zirkon BioStar HT; Siladent, Goslar, Germany).

Production of crowns and abutment dies

A metal master die in the shape of a prepared molar, designed for an all-ceramic crown, was replicated by using a silicone impression material (President®; Coltene AG, Altstätten, Switzerland). The impression was poured with die stone (Vel-Mix; Kerr Corporation, Orange, CA, USA) to produce a master die. A wax-up of a full anatomical crown was fabricated on the master die. A double scan technique was then performed with a laboratory scanner (D900L; 3Shape, Copenhagen, Denmark) to create CAD files for both the wax-up crown and the master die. The semimonolithic crowns were designed with specific dimensions through the CAD software (Figure 1). The CAD data were sent to a certified milling center (Cosmodent AB, Malmö, Sweden) for the production of 100 zirconia crowns with different designs (Figure 2). In the same production step, 100 abutment dies

Figure 1 Illustrations showing dimensions of the crown designs.

Notes: The black line represents the interface between the core and the veneer. The arrows denote the local dimensions of the crowns. M, monolithic; SM0.3, semimonolithic with 0.3 mm buccal veneer; SM0.5, semimonolithic with 0.5 mm buccal veneer; SMW, semimonolithic with 0.5 mm buccal veneer supported by wave design; SMC, semimonolithic with 0.5 mm buccal veneer supported by occlusal cap design.

were milled from a polymer material (Polyoxymethylene; Nordbergs Tekniska AB, Stockholm, Sweden).

The crowns were veneered buccally with a compatible porcelain recommended by the manufacturer (IPS e.max Ceram; Ivoclar Vivadent, Schaan, Liechtenstein). A specially made knife was used to standardize the dimensions of the crowns in the different groups using a method that is used in previous studies.^{17,18} The porcelain was applied using specific brushes in several layers (liner, wash, dentin 1, dentin 2, glaze) according to the manufacturer's recommendations. While applying the porcelain layers, the crown over the abutment was placed and fixated to the brass holder that was rotated manually, permitting the knife to carve the porcelain into the final shape (Figure 3). Each layer was separately fired in a calibrated furnace (Ivoclar P 500; Ivoclar Vivadent) at specific temperatures, again according to the manufacturer's recommendations.

Artificial aging and cementation

All crowns underwent three steps of artificial aging: thermocycling, storing in a 37°C moist environment and cyclic preload. The crowns were first subjected to 10,000 thermocycles in a specially built thermocycling device containing two water baths at different temperatures, 5 and 55°C. Each

cycle lasted for 60 seconds, 20 seconds in each bath, and 10 seconds for transfer between the baths. In the present study, the crowns were subjected to thermocycling before cementation to prevent partly loose crowns during the preload and load to fracture tests since previous studies have shown the deleterious impact of the laboratory thermocycling on the integrity of bonding between the crown and cement.¹⁹

Before cementation, the abutment dies were air-abraded with 110 µm aluminum oxide at a pressure of 2 bar for 15 seconds, at a distance of 10 mm perpendicular to the die surface to increase the surface area. The inner surfaces of the crowns were sandblasted using 50 µm aluminum oxide at a pressure of 2 bar, and then, they were rinsed with water according to the manufacturer's recommendations as was suggested in a previous study by Shahin and Kern.²⁰ After that, the crowns were cemented to the abutment dies with a compatible resin cement according to the manufacturers' recommendation (Panavia F 2.0; Kuraray Medical Inc, Okayama, Japan). During the cement setting, all crowns were loaded in the direction of insertion with a load of 15 N for a period of 60 seconds.

After cementation, the specimens were stored in a plastic container inside an incubator (Memmert Incubator; Memmert, GmbH, Schwabach, Germany) at a temperature of 37°C for 60 days. All specimens were wrapped in a wet paper to

Figure 2 Different crown designs shown: (A) in the CAD software and (B) after milling and sintering.

Notes: M, monolithic; SM0.3, semimonolithic with 0.3 mm buccal veneer; SM0.5, semimonolithic with 0.5 mm buccal veneer; SMW, semimonolithic with 0.5 mm buccal veneer supported by wave design; SMC, semimonolithic with 0.5 mm buccal veneer supported by occlusal cap design.

Figure 3 Porcelain build-up step.

Note: The crown with abutment placed on the brass holder, and the specially made knife was used to standardize dimensions of semimonolithic crowns.

create a moist environment like the oral cavity and to prevent desiccation of the luting cement.

In the final stage of artificial aging, the specimens underwent 10,000 cycles of preload at 30–300 N in the form of a sine wave of 1 Hz. The specimens were mounted submerged in water at 10° inclination relative to the vertical

plane in preloading machine (MTI Engineering AB, Lund, Sweden/Pamaco AB, Malmö, Sweden) as was suggested in previous laboratory studies.^{17,18} The load was applied at the highest point of the inner buccal cusp inclination by a stainless steel ball with a diameter of 4 mm (Figure 4). A 0.2-mm thick plastic foil (PE-Baufolie, Probau; Bauhaus, Zug, Switzerland) was placed between the ball and the crown

Figure 4 Test set-ups used for: (A) cyclic preload and (B) load to fracture.

Note: During the test, the crowns were submerged in water, and a plastic foil was used to spread the load evenly over loaded surfaces.

during preloading to distribute the loading forces evenly as was suggested by Kelly.²¹

In the present study, artificial aging was used to mimic the aging procedure in the clinical situation to evaluate its effect on the fracture resistance of the crowns. There is no consensus about what constitutes the optimal artificial aging procedure in respect of the number of cycles, temperature range, or storage time.^{22,23} In this study, thermocycling and preload tests were adopted from previous studies conducted by the same research group to enable comparisons.^{24,25}

Load to fracture

After artificial aging, all specimens were loaded to fracture by using a universal testing machine (Instron 4465; Instron Co. Ltd, Norwood, MA, USA). The same test setup as during cyclic preload was used for this step, but the load was applied at the highest point of the inner buccal cusp inclination of the crown, with a special indenter designed to accommodate this inclination to prevent sliding movement at load (Figure 4). The crosshead speed was 0.255 mm/min. Load at fracture was registered, and the fracture was defined as an apparent crack in the crown, a load drop, or an acoustic event, whichever came first. The acoustic event was ascertained by hearing a cracking sound that is conveyed by the changes in the load-path graph on a computer that is connected to the testing machine.

Analysis of fracture mode

The fracture surfaces of the crowns were assessed by one examiner. A gross visual and microscopic assessments (Wild M3; Wild Heerbrugg, Heerbrugg, Switzerland) were performed to classify the fracture mode into: adhesive (denoting fracture in the interface between the veneer and the core), cohesive (denoting fracture within the buccal veneer, ie, chipping), or complete (denoting the fracture of both the veneer and the core).

Statistical analysis

The differences in fracture load data among the tested groups were analyzed using a two-way ANOVA, followed by Tukey's post hoc test (IBM SPSS Statistics 22; SPSS Inc., Chicago, IL, USA). Fracture mode was analyzed using Fisher's exact probability test. Results were considered statistically significant at 5% level of significance ($P \leq 0.05$).

Results

Fracture load

There were significant differences in fracture load among the groups depending on the materials used and the different

designs tested ($P \leq 0.05$) (Figure 5). Generally, crowns made of translucent zirconia showed higher fracture loads compared to those made of high-translucent zirconia, regardless of design ($P < 0.001$). Regarding design, the semimonolithic groups with 0.3 mm microcoated buccal veneer (SM0.3) showed the highest fracture loads of all the other semimonolithic groups, with no significant difference compared to the M and SMW groups ($P > 0.05$). The crowns in the SM0.5 group showed significantly lower fracture loads compared to the crowns in the SMW and SMC groups. By comparing the designs in both materials used: the M, SM0.3, and SMW groups of translucent zirconia showed significantly higher fracture loads compared to all the groups of high-translucent zirconia. There was no significant difference between the M group of high-translucent zirconia compared to the SMC and SM0.5 groups of translucent zirconia.

Fracture mode

There were substantial differences among the groups, based on the designs, in the way that the crowns fractured. Two fracture modes were seen visually and under the light microscope: complete fracture and cohesive veneer fracture (Figure 6). All crowns in the M and SM0.3 groups (100%), and all but one of the crowns in the SMC groups (95%) showed complete fractures with significant differences ($P \leq 0.05$) in the fracture mode compared to the crowns in the other two groups (SMW and SM0.5). The SMW and SM0.5 groups showed (70 and 55%, respectively) cohesive veneer fractures (Figure 7). None of the tested crowns showed an adhesive fracture.

Discussion

The null hypothesis assuming that crowns with different semimonolithic designs of both translucent and high-translucent zirconia materials would have equal fracture loads was rejected. The results showed that the semimonolithic crowns with a microcoating porcelain layer withstood the highest fracture loads among the semimonolithic crowns tested, close to the loads seen in the monolithic crowns with no statistical differences in these two crown designs. The results in the groups with semimonolithic wave design showed fracture loads at approximately the same level of the groups with the semimonolithic microcoating design. The fracture loads of the semimonolithic crowns veneered with porcelain 0.5 mm thick were, however, significantly lower compared to the other designs studied. This might imply that a thicker porcelain layer results in lower strength values, reaching those seen with traditionally veneered Y-TZP crowns where the veneered porcelain layer is usually within the 1–1.5 mm

Figure 5 Comparison of mean fracture load (Newton) among the crown designs with two different materials.

Notes: *Mean with the same letters (denoted a, b, and c) did not show any significant difference in fracture load ($P > 0.05$). M, monolithic; SM0.3, semimonolithic with a 0.3 mm buccal veneer; SM0.5, semimonolithic with a 0.5 mm buccal veneer; SMW, semimonolithic with a 0.5 mm buccal veneer supported by wave design; SMC, semimonolithic with a 0.5 mm buccal veneer supported by occlusal cap design.

Figure 6 Fracture modes of the crowns: (A) complete fracture in M, SM0.3, and SMC groups and (B) cohesive veneer fracture in SMW and SM0.5 groups.

Notes: M, monolithic; SM0.3, semimonolithic with 0.3 mm buccal veneer; SM0.5, semimonolithic with 0.5 mm buccal veneer; SMW, semimonolithic with 0.5 mm buccal veneer supported by wave design; SMC, semimonolithic with 0.5 mm buccal veneer supported by occlusal cap design.

range in thickness.^{8,13} This assumption, however, is not confirmed in the present study since fully veneered crowns were not included. One postulate in the study design was that the overall thickness of the crowns should be identical in all the groups, simulating a standardized preparation depth regardless of the design to be tested. This means that the thicker the veneering porcelain, the thinner the supporting core material

underneath became, which probably played a significant role in the results of the study. This is worth considering when analyzing the results. It is always a limiting factor in the clinical situation where preparation that is too aggressive removes important, often healthy, tooth structure, thus affecting the prognosis for the tooth vitality.²⁶ Hence, one indication for using monolithic zirconia is that it enables tooth tissue to

Figure 7 Comparison of fracture mode among the crown designs with two different materials.

Notes: M, monolithic; SM0.3, semimonolithic with a 0.3 mm buccal veneer; SM0.5, semimonolithic with a 0.5 mm buccal veneer; SMW, semimonolithic with a 0.5 mm buccal veneer supported by wave design; SMC, semimonolithic with a 0.5 mm buccal veneer supported by occlusal cap design.

be saved due to a material strength that makes a restoration durable even when it is designed as a thin (0.5 mm) crown coping.¹⁰ Thus, it might not be considered clinically relevant to test crown specimens where the overall thickness increases by increasing the thickness of the veneer portion since this would have necessitated a different preparation depth for the different groups in the study. It was, therefore, decided that the same clinical situation with respect to the preparation depth should be represented in all groups. Consequently, increasing the porcelain thickness resulted in a thinner core, providing less support for the 0.5 mm porcelain layer than for the crowns with a microcoating layer. Furthermore, the larger the volume of veneering porcelain, the greater the probability of flaw populations with a critical flaw existing in the material portions under stress during loading.^{27,28} This might be another reason for the lower fracture loads in the

groups with 0.5 mm veneer thickness compared to the groups with 0.3 mm veneer thickness.

Previous studies have suggested the minimum fracture strength for a dental restoration, allowing it to be used in all areas of the mouth.^{29,30} These suggestions were proposed in line with the data relating to average maximum bite forces, which vary in different areas in the mouth and range from 90 to 900 N.^{29,30} Maximum bite forces differ from patient to patient, and extreme loads have been registered in some isolated individuals. The highest measured load known to the authors was 4,000 N measured in a single individual, whereas the same source reported 2,000 N as the second highest load measured in another single person.³⁰ It is not reasonable, however, to aim for such high bite forces when designing restorations since the vast majority of patients seldom reach or exceed 1,000 N, and normal chewing forces are significantly

lower. With that in mind, all the groups in the present study performed well and with a high margin of safety in respect of the expected load under function in the oral environment.

Some cases require the use of thicker porcelain layers, ≥ 0.5 mm, for a better esthetic outcome. In such situations, it might be advantageous to protect the veneer by using other means of core support such as the designs proposed in this study. The crowns with a 0.5 mm buccal porcelain thickness, supported by wave-like projections in the zirconia core, showed significant improvement in fracture resistance compared to the crowns with the same thickness but without the addition of wave-like projections. This might be explained by the zirconia waves, which may prevent shear forces from developing along the buccal porcelain–zirconia interface. If that is the case, shear stress is interrupted by the wave projections, which act as support along the buccal interface. However, further studies, eg, using finite element analysis are needed to analyze the different developing stresses of this design to confirm this explanation.

Improvement in fracture resistance was also seen when the buccal porcelain was protected by a cap-like projection protecting the porcelain from high occlusal loads. These semimonolithic designs, however, are more complicated for the dental technician to prepare, and additional time is thus needed for the CAD-up compared to monolithic and traditional bi-layered designs where default settings can be used.

When comparing translucent zirconia with high-translucent zirconia materials in the present study, it was seen that the design was able to compensate for unfavorable mechanical properties of the weaker high-translucent material to some extent, by using a microcoating or wave design instead of one of the other designs investigated. In the present study, two zirconia materials were used each with a different degree of translucency. The first represents one of the early generations of translucent zirconia, and the second represents a more recent one with improved optical properties. There are different ways to gain increased translucency in densely sintered zirconia. One way is to alter the grain sizes and indirectly the proportional distribution of grain boundaries.^{8,31} Some producers also use materials with a higher content of cubic crystals at the expense of the tetragonal ones.¹³ The toughness of zirconia is one of the properties that make the material durable compared to all other dental ceramics.^{32,33} When a crack is formed during loading and stress concentrations develop at a critical flaw or in the crack-tip area in the material, phase transformations occur that close the crack tip, leaving it in residual compression, which prevents propagation of the crack, so-called transformation toughening.^{32,33}

This advantageous toughening property depends on both grain size and what crystal phases are present in the material. Hence, for the transformation toughening mechanism to be effective, both grain size and phase types present must be balanced. Larger grains tend to transform spontaneously, and grains that are too small do not transform at all, leading in both cases to a less tough material. In both cases, strength and toughness will decrease proportionately in the material.²⁷ If the traditional, optically dense Y-TZP type of zirconia is considered to represent the type of zirconia with optimal composition and grain size, it might be expected that strength and toughness decrease the greater the changes from this ultimate composition. If this is the case, it might explain the results of the present study where the material with the higher translucency showed lower fracture resistance regardless of crown design.

Some studies, however, do not show substantial differences between translucent zirconia and traditional Y-TZP in respect of fracture resistance.⁹ Such results may imply that different brands of translucent zirconia perform differently compared to each other, probably because translucency is gained by using a variety of compositions and production methods, which results in different properties. This leads to contradictory results and makes it difficult to compare different studies as they were testing different brands of translucent zirconia.

Analysis of the failure type of the crowns shows two modes of fracture among the tested designs. The crowns that were microcoated with a 0.3 mm veneer showed the same fracture pattern as the monolithic crowns (complete fractures). Since the porcelain layer was thin and the supporting core material was relatively thick, the porcelain was probably subjected to mainly compressive forces during loading, which in general, ceramics are very resistant to compared to shear and bending forces.²⁷ Consequently, the core material became the limiting factor, resulting in complete fractures. Almost the same result was seen in the cap design where the buccal veneer was protected by the occlusal cap. This crown design was close to the monolithic design, which was probably the reason for a similar fracture pattern.

Cohesive veneer fractures, on the other hand, were more common in designs with a 0.5 mm buccal veneer with a wave-protected configuration but only with the translucent and not with the high translucent material. The reason for this might be that the tensile stress building up in the core material exceeds the materials' load-bearing capacity before or at the same time as this limit is reached in the porcelain. The stronger translucent material resists the load that exceeds

the porcelain capacity, resulting in a cohesive fracture in the porcelain only. This is the same fracture behavior as detected in clinical and laboratory studies on fully veneered Y-TZP restorations with a comparable thickness of veneer layers.^{3,4,6} Thick layers of porcelain upon zirconia, regardless of coping thickness or design, therefore appear to be more liable to cohesive veneer fracture due to the low fracture strength of the porcelain material.

In vitro studies have shown that the modulus of elasticity of the supporting die material has an influence on the fracture outcomes of tested materials.^{25,34} It is recommended that the mechanical properties of dies are close to those of the dentin, ranging from 10 to 15 GP, to prevent them from fracturing during loading tests and to prevent the generation of irrelevant fracture data of the crowns. In the present study, however, the die material used has a modulus of elasticity slightly lower than the recommended range. Nevertheless, all the dies remained intact during all the artificial aging steps and load to fracture tests. Fracture loads were also comparable to previous laboratory studies using dies with elastic properties within the desired range.^{6,9,25} In addition, the die preparation design and the resin cement used in this study were standardized for all the groups tested and were used according to the manufacturers' instructions.

One of the limitations of laboratory studies is the difficulty to choose which aging procedures would produce comparable clinical results. Previous studies have shown the significance of combining the mechanical tests with water and thermocycles to produce the fatigue effect of zirconia.^{22–25} Other studies performed accelerated aging protocol in autoclave for aging zirconia using high temperatures reaching 134°C, which could not be considered representative for the oral environment.^{35,36} The specific aging protocol of the present study was performed in attempt to mimic the real clinical environment that the crowns would be subjected to it and to make comparisons with previous studies with similar methodologies.^{6,9,24,25}

There are always limitations in relation to the methodology of laboratory studies, and the results of the present study may not fully represent the clinical situation. The oral environment is far more complex and is affected by multiple factors that it is impossible to fully mimic, and it is difficult to extrapolate results from the mechanical loading test alone. One such limitation in the present study is the fact that the steel contact counterpart in the load to fracture apparatus has a different modulus of elasticity than the clinical contact counterpart (food, teeth), which influences on the stress magnitude suffered by the material. Therefore, the results

of the present study need to be confirmed by results from clinical studies.

Conclusion

Within the limitations of the study, the following can be concluded:

- Translucent and high-translucent zirconia crowns might be used in combination with 0.3 mm microcoating porcelain layer with semimonolithic design to enhance the esthetic properties of restorations without significantly decreasing the fracture resistance of the crown.
- If 0.5 mm porcelain layer is needed for a semimonolithic crown, wave design or cap design might be used to increase the fracture resistance.

In both cases, the fracture resistance gained is likely to be clinically sufficient as the registered fracture loads were high in relation to expected loads under clinical use. Clinical studies are, however, needed to confirm those findings.

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Author contributions

FB and PVvS contributed to study conception, design, data acquisition, analysis, and interpretation and drafted and critically revised the article. PS, EP, and CL contributed to study conception and design and drafted and critically revised the article. All authors gave final approval of the version of the article to be published.

Disclosure

The authors report no conflicts of interest in this work.

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III

Load-Bearing Capacity of Monolithic Zirconia Fixed Dental Prostheses Fabricated with Different Connector Designs and Embrasure Shaping Methods

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Keywords

All-ceramic restorations; connector design; translucent zirconia; zirconia; Y-TZP.

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Abstract

Purpose: To investigate the load-bearing capacity and failure mode of monolithic zirconia fixed dental prostheses (FDPs) fabricated with different connector designs and embrasure shaping methods.

Materials and Methods: Seventy four-unit zirconia FDPs (with two premolar pontics) were fabricated and divided into seven groups ($n = 10$) according to the different connector designs gained by using different embrasure shaping methods. The groups were as follows: monolithic FDPs fabricated with sharp embrasures, monolithic FDPs fabricated with blunt embrasures, monolithic FDPs fabricated with blunt embrasures and no occlusal embrasures, two groups of monolithic FDPs fabricated with blunt embrasures and interproximal separations made with diamond discs at the soft stage and at the fully sintered stage, and monolithic FDPs fabricated with blunt embrasures and interproximal separation accentuated by localized porcelain build-up. A final group was used as a control group, where fully veneered traditional zirconia FDPs were fabricated with default milling settings. The FDPs were artificially aged and loaded to fracture. Load to fracture and failure modes were analyzed by one-way ANOVA, Tukey's post hoc test, and Fisher exact test ($\alpha = 0.05$).

Results: The FDPs fabricated with interproximal porcelain separation showed significantly the highest load to fracture ($1038 \text{ N} \pm 82$) of all groups ($p < 0.001$), with no significant difference compared to the FDPs with no occlusal embrasures ($934 \text{ N} \pm 175$; $p > 0.29$). The FDPs fabricated with blunt embrasures showed significantly higher load to fracture ($873 \text{ N} \pm 115$) compared to the FDPs in the control group ($689 \text{ N} \pm 75$) and the FDPs with sharp embrasures ($417 \text{ N} \pm 87$; $p < 0.001$). There were no significant differences between the FDPs with sharp embrasures ($417 \text{ N} \pm 87$) and the FDPs with interproximal disc separations ($467 \text{ N} \pm 94$; $p > 0.23$). Failure mode of the FDPs fabricated with sharp embrasures and interproximal disc separations differed significantly compared to the FDPs in the other groups ($p < 0.001$).

Conclusions: Sharp embrasures and interproximal separations made with diamond discs significantly decrease the load-bearing capacity of monolithic zirconia FDPs compared to FDPs made with blunt embrasures. Blunt embrasures in combination with localized porcelain build-up produce FDPs with high load-bearing capacity in relation to loads that might be expected under clinical use.

Yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) has been used for all-ceramic fixed dental prostheses (FDPs) since the mid-1990s, and is today accepted as a high-strength material with unique transformation toughening properties.^{1,2} The conventional way of using Y-TZP for dental purposes is as a framework combined with veneering porcelain. Since Y-TZP

has high light-scattering properties,¹ it appears almost opaque if not covered with a veneering material, such as specially developed porcelain, a material combination that has now been used for several years.^{3,4}

An inherent weakness in such veneered Y-TZP restorations is the veneering material.^{5,6} Chip-off fractures of the veneer

porcelain have been commonly and more frequently reported than with metal-ceramic restorations.^{7,8} This problem might have diminished today, since slow cooling protocols have been introduced at dental labs to avoid subsurface stress formation during sintering of the veneer.^{9,10} Still, the veneer and the interface of the two different materials constitute the weak link of veneered Y-TZP restorations.

Monolithic translucent zirconium dioxide (zirconia) restorations, designed without veneering material, were recently developed and have become a popular choice for replacing missing teeth.¹¹⁻¹³ One advantage of monolithic zirconia FDPs is that such restorations can be prepared without the need for space for the veneering material. With that in mind, it is easy to believe that the extra space gained, allowing for the strength of the zirconia material, will give the restoration a much higher load-bearing capacity compared to the veneered restorations with less space left for the strong framework. Previous studies, however, have pointed out that the design of the connector area is the most influential factor in the clinical failures of Y-TZP FDPs, being second only to chip-off fracture.^{7,8,14-16}

Esthetic considerations play an important role in designing the connector area of an FDP. This applies to monolithic FDPs as well, although the framework shape differs, and when using veneering material, the connector should preferably be designed with U-shaped gingival embrasures. The radius of the embrasure is important because a larger radius increases the load-bearing capacity of the restoration, whereas a smaller radius acts as a stress concentrator for crack propagation. Thus, connector design plays an important role in the overall strength of all-ceramic FDPs.¹⁷⁻²⁰

Monolithic zirconia FDPs are prepared in such a way to allow pontics to appear as individual teeth. One way of doing this is to use a thin diamond disc to accentuate the separations. Interproximal porcelain build-up is another way of enhancing the perception of separations after milling. Both are common methods used in dental laboratories to shape interproximal embrasures of monolithic FDPs so that they appear natural, thus improving the esthetic appearance; however, there is always a risk that the connector design will become sharp and V-shaped rather than U-shaped and thus structurally weaken the restoration. Additionally, esthetic separation procedures carried out on a fully sintered restoration may cause damage to the zirconia surface that initiates phase transformations in the monoclinic microstructure of the material.^{1,21}

The aim of this study was to investigate the load-bearing capacity and failure mode of monolithic zirconia FDPs with different connector designs gained by using different embrasure shaping methods. The null hypothesis was that there would be no difference in the load-bearing capacity and failure modes among the groups under study.

Materials and methods

Seventy four-unit zirconia FDPs (with two premolar pontics and two retainers) were fabricated. The FDPs were divided into seven groups ($n = 10$; Table 1) according to the different connector designs gained by using different embrasure shaping methods performed at the intermediate connector of the FDPs (Fig 1).

A plastic model of a maxillary jaw was used (KaVo YZ; KaVo Dental GmbH, Biberach, Germany), and two abutment preparations were made on the canine and the first molar. The two premolars were removed to provide space for pontics in a four-unit FDP. The abutment preparations were designed to provide the space allowance for an all-ceramic restoration, with a 120° chamfer and 15° convergence angle. A full-arch impression (President; Coltene AG, Altstätten, Switzerland) was made and poured with die stone (Vel-Mix; Kerr Corp, Orange, CA) to produce a master cast. Using a double-scan technique, the master cast was scanned first by using a laboratory scanner (D900L; 3Shape, Copenhagen, Denmark). Subsequently, two different wax-ups were scanned: full anatomy for monolithic groups and reduced anatomy (cut-back) for the veneered group.

After the master cast and the wax-ups were scanned, the data were transferred to a computer loaded with computer-aided design (CAD) software. The connector dimensions of all the FDPs were set identically to 3 mm × 3 mm, except for the FDPs designed without occlusal embrasure, to which 1 mm extra was added occlusally. In all groups, the minimum thickness of the occlusal core was set to 1 mm, and the minimum thickness of the axial walls was set to 0.8 mm. The FDPs in the control group were designed to be fully veneered with 1 mm porcelain according to the manufacturer's recommendations (IPS e.max Ceram; Ivoclar Vivadent, Schaan, Liechtenstein).

In all groups, the gingival and occlusal radii of the embrasures in the connector areas were set according to the default settings of the CAD program, except for monolithic FDPs with interproximal porcelain separation where a large radius cut-back was used to accommodate the buccal veneering material. The CAD data were sent to a certified production center (Cosmodent AB, Malmö, Sweden) where they were used to produce the FDPs: 60 monolithic, translucent zirconia (BruxZir HT; Glidewell Dental Laboratories, Newport Beach, CA) and 10 traditional zirconia frameworks (BruxZir Solid Zirconia; Glidewell Dental Laboratories). The FDPs with sharp embrasures were fabricated using a sharp milling tool (0.30 mm diameter, Zenotec; Wieland Dental/Technik GmbH, Pforzheim, Germany). The FDPs in the other groups were fabricated with a blunt milling tool with a 1 mm diameter followed by different post-milling embrasure shaping procedures on the same intermediate connector. No such shaping procedures were carried out in two groups (the FDPs with blunt embrasures and the FDPs designed with no occlusal embrasures). In two FDP groups, the intermediate connector was separated manually using a separation disc (S-H 394F 220; Sunshine Diamonds GmbH, Langenhagen, Germany) at both the soft stage and the fully sintered stage.

Disc separation was standardized and performed using the same type of disc, with a rotational speed of 17 rpm for 15 seconds. New discs were used for each FDP in these two groups. One operator, using the same hand pressure as far as possible in a manual process, did the disc separation. The connector area was examined by using a light microscope (Leica DFC 420; Leica Microsystems CMS GmbH, Wetzlar, Germany) to confirm the standardization of both the location and the depth of the notch area created during the disc separation step. Monolithic FDPs with interproximal porcelain separation and the FDPs in the control group were not veneered but instead underwent

Table 1 Description of study groups

Connector design and embrasure shaping method	FDP design
Sharp embrasures	Monolithic four-unit FDPs
Blunt embrasures	Monolithic four-unit FDPs
Blunt embrasures with no occlusal embrasures	Monolithic four-unit FDPs
Blunt embrasures and interproximal separations made with diamond discs at soft stage	Monolithic four-unit FDPs
Blunt embrasures and interproximal separations made with diamond discs at fully sintered stage	Monolithic four-unit FDPs
Blunt embrasures and interproximal separations accentuated by localized porcelain build-up	Monolithic four-unit FDPs
Control group with default milling settings	Veneered four-unit FDPs

Figure 1 The FDPs with different connector designs gained by different embrasure shaping methods. *Two different groups of monolithic FDPs were fabricated with interproximal separations made with diamond discs at the soft stage and at the fully sintered stage.

a veneer-simulating process in which the frameworks were subjected to heat treatment according to the manufacturer’s recommendations to simulate the firing cycles of the veneering porcelain (IPS e.max Ceram). Seventy abutment models were produced from a polymer material (POM-C GF25; Simrishamns Mekaniska AB, Simrishamns, Sweden) with a modulus of elasticity comparably close to dentin (9 GPa).

All FDPs underwent 10,000 thermal cycles in two water baths at temperatures of 5°C and 55°C (THE-I100; SD Mecharonik GmbH, Feldkirchen-Westerham, Germany). The FDPs were cemented onto the abutment models using dual-polymerized resin (Panavia V5; Kuraray Medical Inc., Okayama, Japan) following the manufacturer’s instructions. Before cementation, an air abrasion device (Basic Quattro IS; Renfert GmbH, Hilzingen, Germany) was used to air-abrade the abutment models with 110 µm aluminum oxide at a pressure of 2 bars, from a distance of 10 mm and at an angle of 90° to the

abutment surface. The abutment models were also treated with two primers (Tooth Primer, Clearfil Ceramic Primer; Kuraray Medical Inc.) according to manufacturer’s instructions. After cementation, the specimens were subjected to 10,000 cycles of preload at 30 to 300 N at a frequency of 1 Hz in the preloading machine (MTI Engineering AB; Lund, Sweden/Pamaco AB, Malmö, Sweden). The specimens were mounted submerged in water, at 10° of inclination relative to the vertical plane. The loads were applied centrally on the buccal cusp of the second premolar pontic in the control group using a special indenter designed to prevent sliding movements during loading. The other groups were loaded with a ball indenter (4 mm diameter) on the mesial marginal fossa of the second premolar pontic (Fig 2). All specimens were stored in a humid environment at 37°C for 60 days before they were loaded to fracture.

After aging, all specimens were mounted in a testing jig at 10° of inclination and loaded to fracture by using a universal

Figure 2 Setup of preload and load-to-fracture tests. In both tests, specimens were loaded at 10° of inclination (red line) in relation to the loading direction (black dotted line). All specimens were submerged in water with plastic foil (yellow area) placed between indenter and FDPs. [Color figure can be viewed at wileyonlinelibrary.com]

testing machine (Instron 4465; Instron Corp, Norwood, MA). All specimens were submerged in water during loading with 1 mm thick plastic foil placed between the indenter and the FDPs to distribute the loading forces evenly (Fig 2). The crosshead speed was set to 0.255 mm/min, and the loads were registered. The fracture was defined as either an apparent crack, a load drop in the graph, or an acoustic event, whichever came first. The acoustic event was determined by a crack sound accompanied by changes in the stress and strain graph on the computer connected to the testing machine. Failure modes of the FDPs in the different groups were identified visually and by microscopic assessment (Leica DFC 420; Leica Microsystems CMS GmbH) and were classified according to the starting points of the fracture and its propagation path.

Power analysis was based on previous studies where differences regarding the level of significance and standard deviation were detected among the zirconia-based specimens.²⁰⁻²³ The differences in load to fracture of the groups were analyzed using one-way ANOVA, followed by Tukey's post hoc test (IBM

SPSS Statistics v24; IBM Corp, Chicago, IL). Failure mode data were analyzed using Fisher exact test. Results were considered statistically significant if $p \leq 0.05$.

Results

Load to fracture and levels of significance of the groups are detailed in Table 2. There were significant differences in load to fracture among the groups based on the connector designs gained by different embrasure shaping methods used in the study, $F(6, 63) = 69.623$, $p < 0.001$. The FDPs fabricated with interproximal porcelain separation showed significantly the highest load to fracture ($1038 \text{ N} \pm 82$) of all groups ($p < 0.001$), with no significant difference compared to the FDPs fabricated with no occlusal embrasures ($934 \text{ N} \pm 175$; $p > 0.29$). The FDPs fabricated with blunt embrasures showed significantly higher load to fracture ($873 \text{ N} \pm 115$) compared to the FDPs in the control group ($689 \text{ N} \pm 75$) and the FDPs with sharp embrasures ($417 \text{ N} \pm 87$; $p < 0.001$). The FDPs fabricated with interproximal disc separation at soft stage had the lowest load values numerically ($357 \text{ N} \pm 45$), but with no significant differences compared to the FDPs fabricated with interproximal disc separation at fully sintered stage ($467 \text{ N} \pm 94$; $p > 0.23$).

All FDPs fractured in the intermediate connector of the two pontics, and no abutment fracture was observed. There were significant differences among the groups with regard to failure modes ($p < 0.001$). Table 3 presents the distribution of failure modes of the FDPs among the tested groups. Failure mode of the FDPs with sharp embrasures and the FDPs with interproximal disc separations showed that all fractures (100%) started from the gingival side of the intermediate connector in the notch area created by either the disc or the sharp milling tool, propagating toward the loaded pontic (Fig 3). By contrast, all fractures (100%) in the FDPs of the other groups occurred at the distal corner of the intermediate connector on the gingival side, propagating toward the loading site (Fig 4).

Table 2 Load to fracture (N)

Specimen number	Sharp embrasure	Blunt embrasure	No occlusal embrasure	Separation with disc (soft)	Separation with disc (fully sintered)	Separation with porcelain	Control
1	367	820	743	360	449	1014	694
2	394	745	855	422	380	959	705
3	463	800	1076	334	516	1011	771
4	464	774	988	386	332	947	691
5	374	1033	1220	330	442	1072	642
6	376	830	744	426	637	1006	708
7	477	945	1107	340	503	990	606
8	608	1099	736	300**	370	1177	688
9	347	858	1048	377	460	1185	561
10	300**	831	827	300**	578	1022	823
Mean*	417 ^a	873 ^{b,c}	934 ^{b,c}	357 ^a	467 ^a	1038 ^c	689
SD	87	115	175	45	94	82	75

*Means with the same letters in superscript (denoted a, b, and c) did not show any significant difference in load to fracture ($p > 0.05$).

**FDPs fractured at 30 to 300 N preload.

Table 3 Distribution of failure modes for each group

Groups	Starting points of the fracture	
	Notch area in intermediate connector	Distal corner of intermediate connector
Sharp embrasure	10	0
Blunt embrasure	0	10
No occlusal embrasure	0	10
Separations with disc (soft)	10	0
Separations with disc (fully sintered)	10	0
Separation with porcelain build-up	0	10
Control	0	10

Figure 3 Fracture mode of the FDPs with interproximal disc separation (lingual view). Fracture started from the gingival side of the intermediate connector in the notch area (triangle) created during postmilling disc separation and extended toward the loading site.

Figure 4 Fracture mode of the FDPs in the control group (buccal view). The fracture occurred at the distal corner (triangle) of the intermediate connector on the gingival side and propagated toward the loading site.

Discussion

There were substantial differences in load-bearing capacity and failure modes among the groups, and consequently the null hypothesis was rejected. The FDPs in the study were divided

into seven groups depending on the connector designs gained by different embrasure shaping methods used. The embrasure shaping methods used in this study were at the CAD/CAM (computer-aided manufacturing) milling step (using sharp and blunt milling tools) and after the milling step (using discs and porcelain separations). The results show that different connector designs gained by varying embrasure shaping methods play a significant role in the load-bearing capacity and failure mode, and this is an essential factor to be considered when designing and fabricating the monolithic FDP.

The FDPs with sharp embrasure design were the only ones prepared with a sharp milling tool. All the FDPs in the other groups were prepared with a blunt milling tool. This group, together with interproximal disc separation groups, where the bluntly milled separations were sharpened manually with a diamond disc, failed at significantly lower loads compared to the other groups. The small embrasure radii of the FDPs in those groups formed starting points for crack formation, acting as stress concentrators, which most probably played a significant role in the low load-bearing capacity of the FDPs.

The loads in the study were applied at 10° inclination relative to the vertical plane to create a stress distribution mimicking clinical conditions. This test set-up results in major stress concentrations, with compressive stress occurring in the occlusal area and tensile stress occurring in the gingival embrasure of the connector, which is the thinnest part of the framework. The latter contributes to the propagation of the existing flaws created by the sharp milling or post-milling disc separation, leading to fracture at relatively moderate loads. These findings are in agreement with previous studies, which concluded that three-unit FDPs fail at lower load to fracture if they are shaped with a sharp connector design compared to round symmetrical connector designs.^{19,24}

The results also revealed that monolithic FDPs fabricated with blunt embrasures with post-milling separations made by porcelain build-ups to create the desired interproximal embrasure shape withstood the highest load to fracture of all the other FDPs including the ones with a larger connector dimension. This might be because the FDPs in this group were designed with a large radius of curvature to allow space for the added porcelain. Previous studies have shown that large radii of curvature are favorable for the durability of all-ceramic FDPs.¹⁸⁻²⁰

The connector dimension of all groups was identical and was selected based on the manufacturer's instructions and the recommendations from previous studies.¹⁶⁻²⁰ Only the FDPs

with no occlusal embrasures were designed with a larger connector dimension to observe the effect of changing the connector dimension. As expected, apart from the interproximal porcelain separation group, the FDPs of this group demonstrated a higher load to fracture than those with a smaller connector.

Previous studies have reported an average maximum bite force ranging from 90 to 200 N in the anterior region and 300 to 800 N in the posterior region.^{25,26} Hence, all milling and postmilling separation procedures investigated in this study could be used for anterior monolithic zirconia prostheses, but with a low margin of safety when using the sharp milling and post-milling disc separation strategies; however, in the posterior area, both the sharp milling and post-milling disc separation strategies should be avoided, given the loads expected in this area under clinical use.

With regard to failure mode results, all the fractures occurred in the connector area of the FDPs, which is in line with previous reports.^{7,8} The fracture origin of the FDPs with sharp embrasure design and the FDPs with interproximal disc separations was located in the notch area created either during machining with the sharp milling tool or post-milling by using a separation disc. These processes form a pattern of cracks and scratch-like flaws in the interproximal area and subsequently result in detrimental crack growth during function. The fracture origin in the groups with blunt milling, on the other hand, was located close to the corner of the gingival embrasure of the intermediate connector of the loaded pontic. This fracture pattern has been demonstrated in previous *in vitro* studies and may be related to the tensile stress created by the strain in the gingival embrasure of the connector and the compressive stresses in the loaded area.¹⁶⁻²⁰ Further fractographic analysis, using a scanning electron microscope might provide more information on failure behavior with specific fracture features.

The frameworks of the FDPs were not veneered with an actual porcelain material due to the difficulty of standardizing the veneering process; however, all the FDPs in the control group and all the FDPs with interproximal porcelain separation were subjected to a simulated heat treatment process, following the recommended porcelain firing cycles. This was done in line with previous studies to ensure realistic results, since potential changes in the mechanical properties of the Y-TZP might result from the porcelain firing cycles.²⁷

In this study, during mechanical tests (cyclic loads and load to fracture), the external loads were transferred to the FDPs using indenters made of steel. The hardness and stiffness properties of the (steel) contact counterpart may have played a role in the magnitude of stress in the material. High internal forces within zirconia are expected to develop as a result of external loads with stiff indenters. Therefore, it is not possible to translate the resulting load to fracture directly to the clinical situation, where different contact counterparts (teeth, crown materials, fillings, food) with different mechanical properties are present. However, the loading conditions were similar for all the groups included, which justifies comparisons among the different groups. In reality, only clinical studies can confirm the results and real-life differences among monolithic zirconia FDPs fabricated with different connector designs and embrasure shaping methods.

Conclusions

1. Sharp embrasures and interproximal separations made with diamond discs significantly decreased the load-bearing capacity of monolithic zirconia FDPs compared to FDPs made with blunt embrasures.
2. Blunt embrasures in combination with localized porcelain build-up produced FDPs with high load-bearing capacity in relation to loads that might be expected under clinical use.
3. Great care should be taken when performing interproximal disc separations to avoid decreasing the overall strength of monolithic zirconia FDPs.

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IV

Evaluation of stress distribution in tooth-supported fixed dental prostheses made of translucent zirconia with variations in framework designs: a three-dimensional finite element analysis.

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Stress distribution in translucent zirconia FDPs

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ABSTRACT

Purpose: To evaluate the influence of the framework designs on the stress distribution within tooth-supported partially veneered fixed dental prostheses (FDPs) made of translucent zirconia under simulated loads using a three-dimensional finite element analysis (3D-FEA).

Material and Methods: For a linear 3D-FEA, simplified 3D solid models of prepared abutment teeth (first premolar and first molar) with different 3-unit FDPs were created. The models with different FDPs designs—monolithic zirconia (control); semi-monolithic zirconia with 0.3 mm veneer thickness (SM0.3); semi-monolithic zirconia with 0.5 mm veneer thickness (SM0.5); semi-monolithic zirconia with 0.5 mm veneer thickness supported with cap design (SMC), and semi-monolithic zirconia with 0.5 mm veneer thickness supported with wave design (SMW)—were analyzed using 3D-FEA. The elastic properties of the components (bone, dentine, cement, translucent zirconia, and veneering porcelain) were obtained from the published data for FEA. Simulated static loading forces (300 N) were applied at 10° oblique direction over six points in the occlusal surfaces of the FDPs. Maximum principal stress, shear stress, and safety factor were calculated and analyzed among the different models.

Results: Semi-monolithic with cap design showed the smallest maximum principal stress levels in the veneering porcelain compared to all other models (SM0.3, SM0.5, SMW). The SM0.3 had lower maximum principal stress levels in the veneering porcelain compared to SM0.5. Regarding stresses in the zirconia framework, all models had comparable results in maximum principal tensile stresses, except SMW had a lower value. Maximum principal stress levels were located in the veneer component of SM0.3, SM0.5, and SMW, whereas, such levels were observed in the cervical areas of the zirconia frameworks of SMC and control. The SM0.3 had the highest maximum shear stress levels at the zirconia-veneer interface, while SMW had the lowest shear values. The 3D-FEA models with different FDP designs showed different minimum safety factor levels.

Conclusions: Framework and veneer designs play a significant role in the stress distribution of the partially veneered zirconia FDPs under loading. The FDPs with zirconia frameworks with cap design minimize the maximum principal tensile stress in the veneering porcelain. The FDPs with 0.3 mm-

veneering porcelain show low maximum principal tensile stress in the veneering porcelain, but highest maximum shear stress at the zirconia-veneer interface. The FDPs with wave design of zirconia frameworks minimize the maximum shear stress considerably.

KEYWORDS: All-ceramic restorations; 3D-FEA; monolithic zirconia; porcelain veneer; Y-TZP.

INTRODUCTION

The indications for all-ceramic fixed dental prostheses (FDPs) based on yttria-stabilized tetragonal zirconia polycrystal (Y-TZP) have been extended to the posterior region because of their favorable mechanical properties and the increased demand for highly esthetic restorations.^{1,2} Cohesive failure has been reported in the veneering porcelain of the FDPs made of Y-TZP as a framework material.³⁻⁵ Regarding cohesive failure, several studies suggested possible material-related factors for the unstable behavior of veneered Y-TZP restorations.^{3,6,7} The differences in thermal properties between the zirconia and veneering porcelain in terms of coefficient of thermal expansion (CTE) and thermal diffusivity play significant roles in generating high residual stress leading to chipping problems.^{3,6,7} By controlling the cooling procedures and using compatible CTE values, the high rates of chipping problems in older studies were minimized in recently published ones.^{8,9} Despite that difference, frequent events of cohesive failures still occur.^{4,5}

Based on previous studies, the structural reliability of veneered Y-TZP restorations may be enhanced by modifying the framework design.¹⁰⁻¹² The different design modifications proposed are focused on minimizing the tensile stresses within the material by providing supportive structures in the zirconia framework. Extending the proximal and lingual collars in the Y-TZP framework and the use of anatomically shaped framework designs have been shown to be a good strategy to avoid chipping problems in the veneering layer.^{13,14,15} With these designs; even veneer thickness would be provided, leading to the avoidance of stress development within the veneering layer during cooling after sintering. However, chipping problems were not fully prevented with those two modified designs, and clinicians still encountered this issue.^{4,5}

New zirconia materials with enhanced translucent properties can be used with monolithic designs. Since this design comes with no veneering layer, no chipping fractures are expected. Previous studies have shown favorable results in the overall strength and clinical performance of monolithic zirconia FDPs.¹⁶⁻¹⁸ Compared to veneered ones, however, monolithic zirconia FDPs still show inferior esthetic properties. Previous laboratory studies with modified partially veneered (semi-monolithic) designs demonstrated

promising results for crowns made of translucent zirconia.¹⁹⁻²¹ These modified designs provide higher esthetic outcomes using veneering materials compared to monolithic restorations while maintaining the strength of monolithic design. The stress distribution in zirconia restorations with semi-monolithic designs is, however, may be more complex compared with monolithic designs. This complexity is related to two main factors; the combination of two materials with different properties and the presence of interfaces between the different layers. The previous studies on semi-monolithic designs did not provide detailed information on the nature of stress distribution within the different components of the partially veneered zirconia restorations during loading.¹⁹⁻²¹ Finite element analysis (FEA), however, is considered to be a powerful tool for evaluating the stress distributions within the restorations made of various materials and designs.²²⁻²⁵ To the authors' knowledge, no study has used 3D-FEA to analyze the stress development in tooth-supported FDPs made of translucent zirconia with different semi-monolithic designs.

Therefore, the study aimed to evaluate the influence of the framework designs on the stress distribution within tooth-supported semi-monolithic FDPs made of translucent zirconia under simulated loads using 3D-FEA. The null hypothesis was that there were no differences in the stress distribution within partially veneered 3-unit FDPs with different designs.

MATERIALS AND METHODS

Simplified 3D solid models of prepared abutment teeth (first premolar and first molar) with different 3-unit FDPs based on the designs were created (Table 1), (Fig 1). The different FDPs designs in this study were adopted from a previous laboratory study where anatomically shaped crowns were mechanically tested with these designs.¹⁹ Simulated loading force (300 N) was applied at 10° oblique direction over six points in the occlusal surfaces of the FDPs. Finite element analysis of the 3D solid models was conducted. The maximum principal stress, shear stress at veneer-zirconia interface, and minimum safety factor were calculated.

The teeth preparations, to receive all-ceramic FDP, were made on mandibular first premolar and first molar plastic models (KaVo YZ; KaVo Dental GmbH, Biberach, Germany) using diamond burs (Two striper, Abrasive Technology Inc., Lewis Center, USA). The second premolar plastic model was removed from the plastic mandibular arch to provide space for the pontic in a three-unit FDP. The teeth preparations were performed with a 15° angle of convergence between the axial walls. The height of preparations was set to 5 mm based on the average dimensions of prepared molars and premolars. The marginal areas of the two abutments were prepared with a rounded 120° chamfer angle. For simplicity of modeling, the occlusal surfaces of the preparations were flat. The marginal, axial, and occlusal surfaces reduction were 0.5 mm, 0.8 mm, and 1.5 mm, respectively. The only exception for the axial surfaces was the buccal surface; it was reduced to 1.2 mm to accommodate the veneering porcelain in the different semi-monolithic designs. All line angles of preparations were rounded to decrease stress development in the FDPs. A full-arch impression (President; Coltene AG, Altstätten, Switzerland) was made and poured with dental stone (Vel-Mix; Kerr Corp, Orange, CA, USA) to produce a master cast. The FDP was completed using a double-scan technique in which the master cast was scanned first by a laboratory scanner (D900L; 3Shape, Copenhagen, Denmark) followed by scanning the full-contour wax-up of the FDP. The scanning data of the FDP and the prepared teeth were transferred to a computer loaded with computer-aided design (CAD) software (3Shape Dental Designer Software) and were saved as an STL file. The solid geometries of the prepared teeth, and the FDPs with different components (veneer, zirconia, cement) were rebuilt from the STL file with a 3D modeling software (Creo Parametric 4.0, PTC, Needham, USA). For the convenience of the simulation process, the prepared teeth in the CAD software were fixated in a simplified block of alveolar bone with no separating PDL gap in between. Additionally, a uniform cement gap was included in the 3D model. Five models with different FDPs designs were tested in this FEA study (Table 1), (Fig 1). The FDPs have simplified anatomical occlusal surface with only two cusps (buccal and lingual). The connector dimensions (height, width) of all the FDPs were set identically to 4 mm × 4 mm and were designed with round interproximal embrasures. The first model had full-monolithic design with no veneering layer (control). The second model had semi-monolithic design with

a 0.3 mm thickness of the buccal veneer (SM0.3). The buccal veneer was extended into half of the occlusal surface, and the veneer-zirconia junction was located 2.0 mm away from the simulated loading sites. Proximally, the veneer layer was extended to interproximal contact areas. The third model had semi-monolithic design like the second model but with a 0.5 mm buccal veneer thickness (SM0.5). The zirconia thickness, in this model, had to be reduced to accommodate the increased veneer thickness. For the fourth model (SMC) and the fifth model (SMW), the occlusal cap projection and wave-like projections were added in the zirconia framework respectively to support a 0.5 mm buccal veneer thickness. The veneer-zirconia junction of all models was designed with rounded corners to avoid stress concentration. The total occlusal thickness, in all models, was set to 1.5 mm; including only zirconia material in the control model, and zirconia with veneer in semi-monolithic models.

The 3D solid models with different components (veneer, zirconia, cement, dentin, bone) were exported to the FEA program (ANSYS Mechanical Pro; Professional Version 19.2, Canonsburg, USA). After uploading the models, the meshing process was performed on the models with four-node linear tetrahedral elements, based on the convergence test. The criterion for convergence analysis was set to be less than 6 % changes in the highest von-Mises stress of bone and dentin between the element sizes.²⁶ For accurate analysis, the mesh size was set to 0.7 mm for bone and dentin components. For the zirconia framework and the veneer components, it was set to 0.25 mm and 0.1 mm respectively. The total number of elements for each model was 1,086,445 with a small range of difference ($\pm 101,232$) between the models because of the different designs. The mechanical properties in (Table 2) were applied for each element in the 3D-FEA model.²⁷⁻³⁰ The 3D-FEA models used in this study were based on the following assumptions: (1) All components of the models were linear elastic, isotropic, and homogenous. (2) There were no flaws in any component. (3) All components of the models were considered perfectly bonded. (4) There was a uniform 100 μm cement space based on values from previous studies.^{31,32} (5) The effect of the PDL on generating the stresses within the models was negligible, and the gap junction between the bone and dentin was immovable with no translation or rotation in any direction. (6) The cementum layer that covers the root surface and the pulp chamber were negligible and were included in the dentin of the

teeth since they are thin components with no or lesser effects on generation the stress.³³ The simulation and the numerical analysis of the 3D-FEA models in the present study were performed using the Ansys program to evaluate the different stress fields for the different models. The simulated loading forces of 300 N were applied occlusally over six different loading points (C1-C6). The loading forces were applied at 10° oblique angle relative to the long axis of the clinical crown. The oblique loading forces were directed toward the buccal cusps of the pontic and one of the retainers (Fig 2). The loading points were simulated over the 1.0 mm² circular areas on the occlusal surface. After the simulation, the maximum principal stress, the maximum shear stress at the zirconia-veneer interface, and minimum safety-factor levels were calculated. The analysis of the maximum principal stress was used to differentiate areas of tensile (positive) and compressive (negative) stress in the veneer and zirconia components of all models.

RESULTS

The maximum principal (tensile and compressive) stresses introduced to both veneering porcelain and zirconia framework of the different model designs are presented in Table 3 and Figures 4,5. The results demonstrated that the veneer thickness and the zirconia framework design had considerable effects on the maximum stress distributions within the simulated 3D-FEA models in the 10° oblique load condition.

Concerning veneering porcelain in SM0.3 model with a 0.3 mm-veneer thickness, the maximum principal stresses were lower compared to SM0.5 model with a 0.5 mm-veneer thickness. The SMC model resulted in the smallest maximum principal stresses in the veneering porcelain compared to all other models (SM0.3, SM0.5, SMW). Regarding the stresses in zirconia framework, all models had comparable results in maximum principal tensile stresses, except SMW model had a lower value. The SMC model resulted in a larger maximum principal compressive stresses compared to all other models.

Table 4 shows the locations of the maximum principal stresses in the different models, considering the six loading points (C1-C6) in this study. The results showed that the maximum principal tensile and compressive stress levels were located in the veneer component of SM0.3, SM0.5, and SMW models. The only exception to this pattern was SMC model where the maximum principal tensile stress levels were

observed at the cervical areas of the zirconia framework. Similarly, when analyzing the model with the monolithic design, the peak of tensile stress levels occurred at the cervical areas of the monolithic zirconia, in addition to the loading sites.

Regarding the maximum shear stress values at the zirconia-veneer interface, SM0.3 model showed the highest maximum shear stress, while SMW model had the lowest value (Fig 5). For the specific 10° load condition, the 3D-FEA models with different FDP designs showed different minimum safety-factor levels (Fig 6).

DISCUSSION

This FEA study evaluated the effect of varying the zirconia framework design and the veneer thicknesses on the stress distribution of partially-veneered (semi-monolithic) 3-unit FDPs under loading. The null hypothesis tested was rejected since there were differences in the stress distribution within the FDPs with different semi-monolithic designs.

The analysis of the stress development and the failure criteria in FEA studies can be displayed by both the equivalent von Mises stress and the maximum principal stress. The analysis of von Mises stress, however, does not have valid failure criteria for brittle materials such as porcelain and zirconia and is normally used to display stress levels for ductile materials. Hence, the maximum principal stress was used. For brittle materials, the maximum principal stress is a good indicator for the strength of materials and limits of failure; when the maximum principal stress values exceed the strength in either compression or tension, the material is assumed to fail. When tensile stress develops, rather than compressive stress, the brittle material becomes more sensitive to loading forces, leading to fracture.^{3,34}

For the specific loading conditions applied in this study, the model assemblies with the different designs demonstrated the minimum safety factor regarding the ultimate tensile strength values of the zirconia and the veneering porcelain materials. These results mean that during loading the models, the veneer and zirconia materials would not fail because of the tensile or compressive stresses generated within the FDPs. In this study, the veneer layer and the zirconia framework of the models with different semi-monolithic designs were subjected to different levels of tensile and compressive stresses during the oblique load forces. The results of the high tensile stress values in the veneering porcelain (SM0.3 and SM0.5 models) were correlated with several clinical trials that have reported that layered zirconia FDPs are associated with more chippings of the veneering porcelain than porcelain fused to metal FDPs.³⁻⁵ The results also have shown that the maximum tensile stress levels in the veneering porcelain were decreased in SM0.3 and SM0.5 models, while the compressive stresses were increased, as the veneer thickness decreased from 0.5 mm to 0.3 mm. Since the total thickness in all designs was the same, the reduction in

the veneer thickness in SM0.3 model enables thicker supporting material underneath it. The results were in agreement with previous studies showing the importance of coping thickness for the load-bearing capacity of minimally veneered zirconia crowns.^{10,19,35} In the clinical scenario, the starting points for the chipping fracture of the bilayer zirconia restorations are mainly affected by the existence of critical flaws in the veneer layer.³⁴ Therefore, thicker porcelain layers would be expected to have more flaws leading to fracture at lower loads. Moreover, the zirconia and porcelain materials have low thermal diffusivity leading to generate residual thermal stresses during the cooling process for the porcelain.^{3,35} However, in the present study, all components of the FDPs were assumed to be flawless, and the residual thermal stress factor has been neglected in the simulation. This result highlights the significant role of the design in this study on developing the different stress distribution within the restoration; even though the differences between SM0.3 and SM0.5 models were not as significant as in the case of real crowns which have other critical factors as discussed earlier.

Another finding in this study showed that SM0.3 model with 0.3 mm veneer thickness showed highest maximum shear stress levels at the zirconia-veneer interface compared to the other models. Clinically, this result might increase the risk for adhesive fractures rather than cohesive fractures in the veneering porcelain of the FDPs with this design. This factor might not be an important clinical problem since vast clinical trials have found that the majority of failures were cohesive subsurface-fractures within the veneering porcelain.^{4,5}

In SMC model, the presence of occlusal cap-shaped projection in the zirconia framework reduced the maximum principal stress levels in the 0.5 mm-veneering porcelain. These results are in agreement with previous study which has shown favorable load-bearing capacity for crowns with this cap design under mechanical testing.¹⁹ The highest maximum tensile stress levels in this model, however, were located at the cervical areas of the zirconia framework away from the loading sites similar to the monolithic design. For all models, comparable principal tensile stress values were developed within the zirconia frameworks. On the other hand, regarding the principal compressive stress values, the models with monolithic and cap designs had the highest values within the zirconia frameworks. This result is not

significant since these stress values are low compared to the ultimate compressive strength of the zirconia material.

The model with the wave design (SMW) had a lesser effect on reducing the principal tensile stress within the 0.5 mm-veneering porcelain compared to the model with a cap design (SMC). This finding contradicted previous results showing a better load-bearing capacity for the crowns with wave design.¹⁹ This result might be explained because the strength of wave design depends on reducing the residual thermal stresses which are generated during porcelain firings and the cooling of the real crowns. The thermal properties factor was not considered in the present study. The shear strength of the material is an important property to avoid adhesive-type fractures between the veneer and the framework. The model with wave framework design showed the lowest shear stress at the zirconia-veneer interface compared to all other models, including the cap design.

Previous studies on clinical fractures have mentioned the intaglio surface of all-ceramic restorations and the cervical area as the most frequent fracture sites, rather than the loaded surface of restoration.³⁶⁻³⁸ The findings of this study demonstrate that at the loading point, the subsurface under the occlusal surface of the veneering porcelain, close to the zirconia-cement interface, as well as the cervical area, shows the largest tensile stress in the zirconia framework. However, the principal tensile and compressive values were low compared to the ultimate tensile and compressive strengths of the zirconia.

Regarding the load cases, when conducting a FEA study, the simulated occlusal forces with oblique angles should be considered rather than the forces with vertical angles. Previous studies demonstrated that oblique loading forces generated more stress when compared to vertical loading forces.^{39,40} The results of this study confirmed the significant effect of eccentric lateral force angles to generate high tensile stress in the FDPs. Previous studies also showed that the connector areas of posterior FDPs, where there is insufficient supporting framework material, were frequently subjected to clinical failure.^{4,5,11} Based on these reports, the loading conditions in this study were simulated at specific points at both the connector and the buccal functional cusps of the mandibular 3unit-FDPs where probable fracture might occur.

Many important features should be considered when conducting a FEA study. Features such as the mesh size and number and type of elements can affect the results significantly, reducing the reliability of the results. The meshing process was performed based on the convergence test, to avoid this effect on the results. Optimum meshing was useful to improve the overall precision of the models because the mesh was produced at the finest size possible. However, several limitations need to be mentioned regarding this FEA study. The results of this study must not directly be extrapolated to the clinical situation, regardless of the well-known high reliability of FEA as a powerful tool in much scientific research, including the dental field. Many clinical characteristics were not considered in this present study, such as cyclic fatigue loads and sliding (dynamic) movements of masticatory forces, as well as the thermal cycles. Furthermore, setting up the geometry for the reconstructions and boundary conditions were based on different simplified assumptions. Such assumptions may impact the results of the study. Notably, the study used some simplified anatomical characteristics that were incorporated in the FEM models, such as the flat occlusal surfaces of the preparations and the anatomical occlusal surface of the FDPs with only two cusps. Also, both abutments (molar and premolar) present similar dimensions. While the need for those anatomical simplifications in FEM models is important for simplicity of modeling and simulation avoiding complex calculations, such alterations and modifications in contact surfaces can undoubtedly generate a different pattern of stress distribution, particularly in the case of FDPs, which may not represent a real clinic scenario. In addition these factors, the maximum principal and shear stresses were only two of many other factors to determine the overall strength of the FDPs.

CONCLUSIONS

The framework and veneer designs play a significant role in the stress distribution of the partially veneered zirconia FDPs under loading. The FDPs with zirconia frameworks with cap design minimize the maximum principal tensile stress in the veneering porcelain. The FDPs with 0.3 mm-veneering porcelain show low maximum principal tensile stress in the veneering porcelain, but highest maximum shear stress

at the zirconia-veneer interface. The FDPs with zirconia frameworks with wave design minimize the maximum shear stress considerably.

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TABLES

Table 1. Models with different FDPs designs created in FEA study.

Model	Design of 3-unit FDPs
Control	Monolithic zirconia
SM0.3	Semi-monolithic zirconia with 0.3 mm veneer thickness
SM0.5	Semi-monolithic zirconia with 0.5 mm veneer thickness
SMC	Semi-monolithic zirconia with 0.5 mm veneer thickness supported with cap design
SMW	Semi-monolithic zirconia with 0.5 mm veneer thickness supported with wave design

Table 2. Material and tissue elastic properties of 3D-FEA models. All values were obtained from the published data for FEA.²⁶⁻²⁹

Tissue/Material	Poisson's ratio	Young's modulus (GPa)	Ultimate tensile strength (MPa)
Bone	0.31	16.3	44
Dentin	0.31	18.6	48
Cement	0.27	6.48	97
Translucent zirconia	0.30	210	1000
Porcelain	0.19	74	100

Table 3 Maximum principal stress in veneering porcelain and zirconia frameworks of different model designs at 10° oblique load condition. All values in megapascal (MPa).

Model design	Veneering porcelain		Zirconia framework	
	Maximum principal tensile stress (positive values)	Maximum principal compressive stress (negative values)	Maximum principal tensile stress (positive values)	Maximum principal compressive stress (negative values)
Control	N/A*	N/A	53.4	100.5
SM0.3	61.6	100.7	53.6	27.5
SM0.5	70.1	113.8	52.2	13
SMC	21.2	1.6	52.9	157
SMW	58.5	121.4	42.3	10.4

*N/A: Not applicable

Table 4. Locations of maximum principal stress in the different models regarding the six loading points at 10° oblique loads.

Model design	Location of maximum principal stress*	
	Maximum tensile stress	Maximum compressive stress
Monolithic	M1	C4
Semi-monolithic zirconia (0.3 mm)	C5	C4
Semi-monolithic zirconia (0.5 mm)	C2	C4
Semi-monolithic zirconia (cap)	M2	C3
Semi-monolithic zirconia (wave)	C3	C4

*(C) indicates locations of maximum principal stress levels at the loading points; C2,C3 indicate the loading points at pontic, C4,C5 indicate the loading points at retainer. (M) indicates locations of maximum principal stress levels away from the loading points; M1 Marginal area of the non-loaded retainer (lingual surface), M2 Marginal area of the loaded retainer (buccal surface)

FIGURE LEGENDS

Figure 1 CAD images with illustrations showing dimensions of all models with variations in zirconia framework design and veneer thickness. NB: In the 2D cross-section, the blue shade represents the veneer thickness layer.

Figure 2 Schematic illustration of loading points (C1-C6) and load direction. NB: C1, C2, and C3 indicate loading points at pontic where veneer is fully supported by zirconia, partially supported by zirconia, and partially supported by zirconia near connector respectively. C4, C5, and C6 indicate loading points at retainer where veneer is fully supported by zirconia, partially supported by zirconia, and partially supported by zirconia near proximal ridge respectively.

Figure 3 Maximum principal stress distributions for 3D-FEA models with different FDPs designs.

Figure 4 Different views of 3D-FEA models; (A) FDP with cap design, (B) FDP with wave design, (C) FDP with 0.3 mm veneer thickness, and (D) FDP with 0.5 mm veneer thickness.

Figure 5 Maximum shear stress at zirconia-veneer interface for each model design.

Figure 6 Minimum safety-factor levels for each model design at 10° oblique load condition.

Figure 1 CAD images with illustrations showing dimensions of all models with variations in zirconia framework design and veneer thickness. NB: In the 2D cross-section, the blue shade represents the veneer thickness layer.

Figure 2 Schematic illustration of loading points (C1-C6) and load direction. NB: C1, C2, and C3 indicate loading points at pontic where veneer is fully supported by zirconia, partially supported by zirconia, and partially supported by zirconia near connector respectively. C4, C5, and C6 indicate loading points at retainer where veneer is fully supported by zirconia, partially supported by zirconia, and partially supported by zirconia near proximal ridge respectively.

Figure 3 Maximum principal stress distributions for 3D-FEA models with different FDPs designs.

Figure 4 Different views of 3D-FEA models; (A) FDP with cap design, (B) FDP with wave design, (C) FDP with 0.3 mm veneer thickness, and (D) FDP with 0.5 mm veneer thickness.

Figure 5 Maximum shear stress at zirconia-veneer interface for each model design.

Figure 6 Minimum safety-factor levels for each model design at 10° oblique load condition.

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