The Effect of Cyclic Loading on the Zirconia/Titanium Implant Abutment Interface and the Mechanism of Failure between Three Different Types of Abutments

THESIS

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By

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Abstract

A single implant restoration in the anterior portion of the mouth poses many challenges. With the increasing demand for ideal esthetics, masking the greyish discoloration from the titanium abutment is very difficult especially in patients with a high smile line or a thin gingival biotype. The one-piece zirconia abutment introduced, however was shown to have many clinical complications such as fracture of the zirconia connection and wear of the titanium from the implant fixture. The zirconia abutment with a titanium connection was subsequently introduced to maintain the titanium to titanium implant abutment interface. There are various designs of these abutments available in the market.

The purpose of this study was to compare three types of zirconia abutments with different mechanisms of retention of the zirconia to the titanium interface.

15 Implants and abutments (3 Groups: Friction fit connection (Frft), Cemented (Cem) connection and titanium ring (Ti Ring) friction fit connection) were thermally cycled in water for 15,000 cycles and then cyclically loaded for 200,000 cycles or until failure at a frequency of 2 Hz using a sequentially increased loading protocol up to a maximum of 720 N.

The mean load to failure values were as follows:

The mean load to failure value for Group N was 526 N

The mean load to failure value for Group S was 605 N

The mean load to failure value for Group Z was 288 N.

The abutment with the cemented connection showed the highest load to failure value and the abutment with the titanium ring friction fit connection showed the lowest load to failure value.

The one-way analysis of variance was performed using the SAS procedure MIXED (SAS Institute Inc., Cary, NC, USA).

There was a statistically significant difference between the three types of abutments tested ($p<0.05$).

Dedication

This document is dedicated to my father Dr. Glenn Mascarenhas, my mother Mrs. Anjali Mascarenhas, my sisters Shelly, Sheena Mascarenhas and my fiancé Graham Thomas for their unconditional love and support in all my endeavors.

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Publications

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Fields of Study

Major Field: Dentistry

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Chapter 1: Introduction

P.I. Branemark first presented the concept, experimental and clinical application of osseointegration in 1983.[1] The use of dental implants followed by the final restorative prosthesis was proved successful in completely edentulous patients.[2,3,4] In 1986, Jemt was amongst the first to popularize restoring a single tooth or short span edentulous spaces.[5] Dental implants have become the treatment of choice to replace a single missing tooth.[6]

A single anterior implant is highly predictable and has a high success rate.[7,8,9,10] Currently, in their 6th decade of clinical use, the actual design of the root form implant has not dramatically changed, though the number of restorative options has.

Single implant restorations can be classified into anterior and posterior restorations.

For posterior restorations, the direction of the occlusal forces and functionality of the restoration are of primary importance. However, for an anterior restoration, its high visibility makes esthetics a priority.[11] Today, patients' and dentists' demands for optimal esthetics are increasing.[9,11] Restoring a single anterior implant poses many challenges. Biological, functional and biomechanical parameters must be examined preoperatively. Factors such as the amount of alveolar bone, morphologic type of soft tissue, correct positioning of the implant in three dimensions, the provisional phase, the design and material of the implant abutment, and the material and design of the implant crown affect the final outcome of the restoration.[9,11]

A single implant restoration can be either screw-retained or cement-retained. Screw-retained restorations have the significant advantage of being retrievable when needed for hygiene, a repair or when the abutment screw loosens.[12] However, there are some situations where screw retention is not possible due to implant position.[13,14] In these situations, having a screw access hole through the facial surface of the restoration is unacceptable. Cement-retained crowns can be used more universally. In addition to being more esthetic, it is easier to achieve passivity with cement-retained restorations especially for implant-supported fixed partial dentures.[14,15] Because there is no access hole in the restoration, occlusal function is promoted.[14,15] With the correct cement selection and proper handling of the cement, cement-retained restorations can be made retrievable without compromising esthetics and function.[15]

Titanium has been widely used as an implant abutment material. Its high fatigue strength and biocompatibility with the surrounding soft tissues has been well documented.[16] Despite these advantages it is not the preferred choice in the anterior. Patients with a high smile line or patients with a thin gingival biotype are challenging to restore as the grey hue from the metallic abutment shows through resulting in poor esthetics. The significant greyish discoloration at the gingiva caused by metal abutments was a concern which subsequently led to the development of ceramic abutments.[17,18]

Ceramic abutments have significant advantages over the metal abutments for example, improved esthetics, fit, translucency, ease of fabrication, biocompatibility with the surrounding soft tissue similar to that of titanium abutments and high bending strength.[19] The combination of a ceramic abutment and crown provides better translucency and therefore a better esthetic outcome of the restoration as compared to a metal abutment with a porcelain fused to metal crown.[20,21] (Fig.1.2)

Alumina abutments in particular, were thought to be a solution to the problem but were soon shown to have poor resistance to fracture.[22]

As an alternative, the one-piece zirconia abutment was introduced. Zirconium oxide ceramics have shown favorable soft tissue responses similar to titanium.[23] Zirconia also provides some inherent translucency which mimics the optical properties of a natural tooth unlike the titanium abutment.[24]

Fig. 1.1

Fig 1.1 One piece zirconia abutment

Fig. 1.2

Fig. 1.2 Intra-oral view of the one piece zirconia abutment

The first proposal of the use of zirconium oxide was in 1969 for mainly hip head replacements instead of titanium and alumina. Zirconia as a material has gained popularity in dentistry due to its increased fracture resistance, pleasing esthetics and biocompatibility with the surrounding soft tissue.[25]

Zirconia is a polymorph that exists in three forms: Monoclinic (M), Tetragonal (T) and Cubic (C).[25,26] Unalloyed zirconia is in the monoclinic form at room temperature and upon heating up to 1170° C. The structure is tetragonal between temperatures 1170 - 2370° C. From 2370° C to its melting point, it exists in the cubic form. The most desirable phase is the tetragonal phase. Several stabilizing oxides such as CaO, MgO, CeO_2 or Y_2O_3 help retain the tetragonal structure at room temperature and minimizes the stress induced t$\rightarrow$m transformation thereby arresting crack propagation.[26]

$$\text{Monoclinic} \overset{1170^\circ\text{C}}{\longleftrightarrow} \text{Tetragonal} \overset{2370^\circ\text{C}}{\longleftrightarrow} \text{Cubic}$$

Fig 1.3: Schematic representation of the phases of zirconia

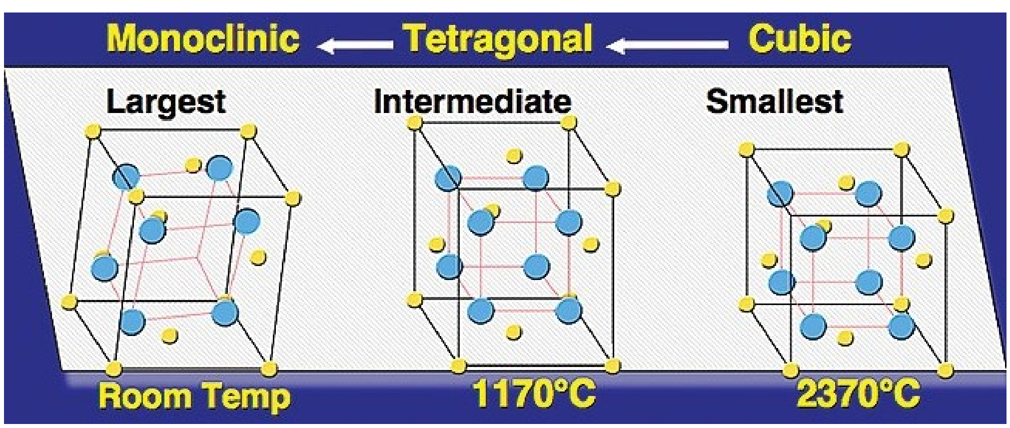

Fig. 1.4: Pictoral Represenation of the phases of zirconia

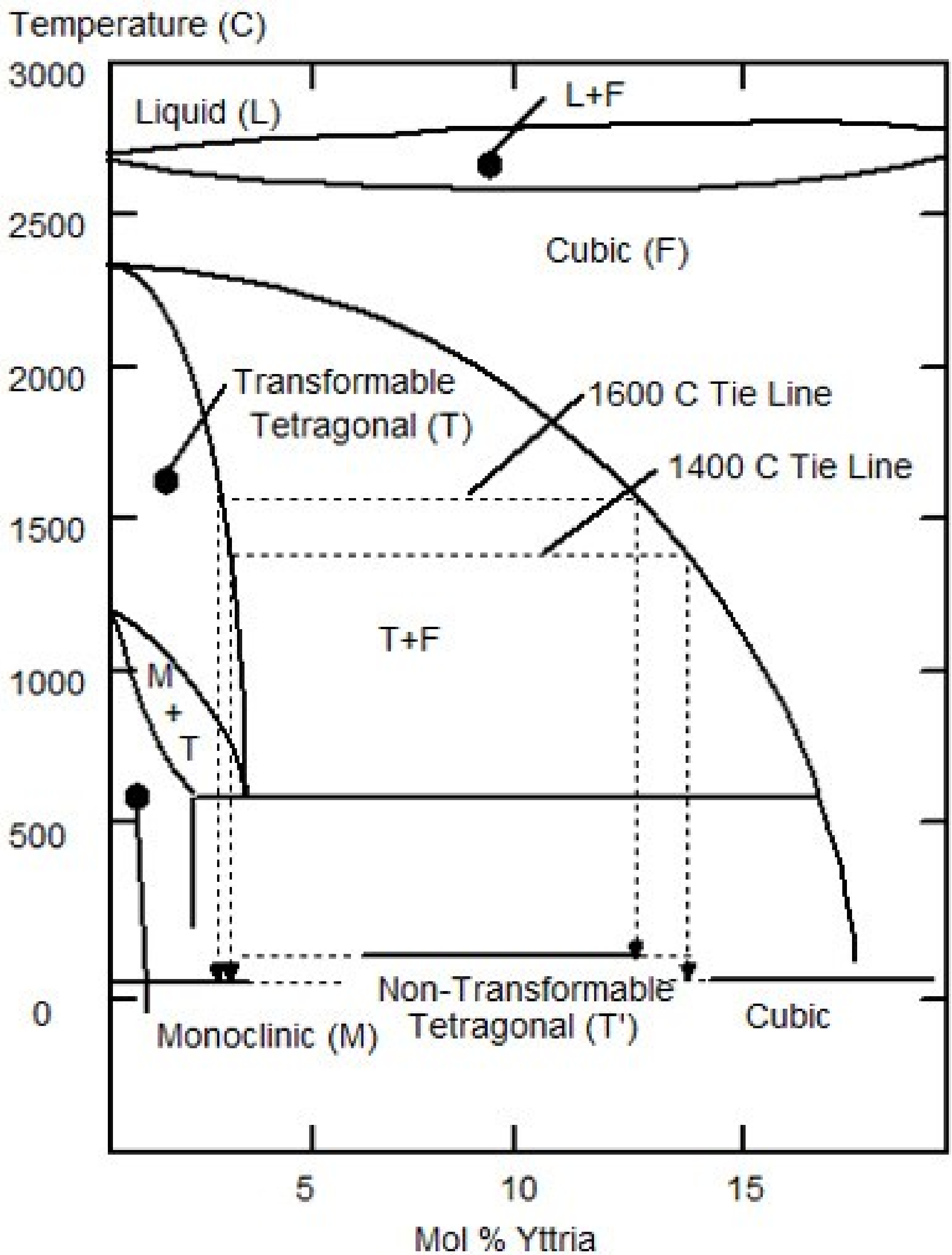

Fig 1.5: Phase diagram for Zirconia

Prior to 1975, the application of pure zirconia was limited.[27] The unique property of transformation toughening of zirconia ceramics was discovered in 1975. Since then, many theories have been developed trying to explain this phenomon.[28,29] The concept behind the different theories is the same. An advancing crack results in the t→m transformation at the tip of the crack. The increase in volume associated with this transformation, results in inhibition of the crack thus toughening the ceramic.[26,29]

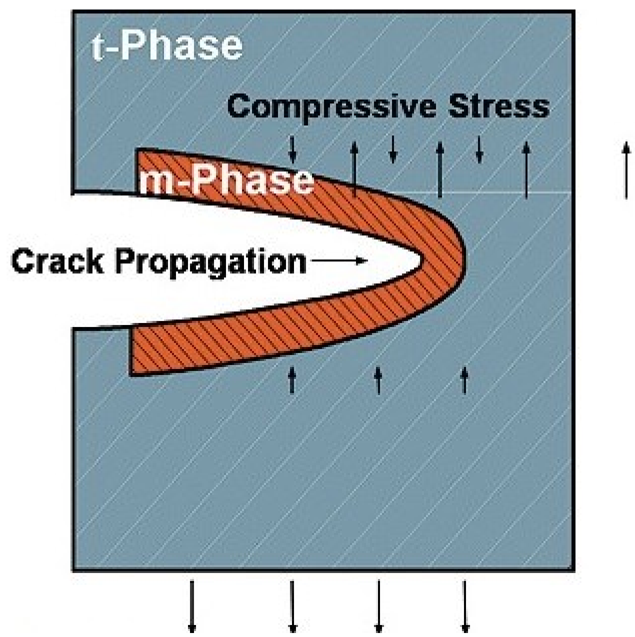

Fig 1.6: Transformation toughening of zirconia also known as active crack resistance.

The three most commonly used types of zirconia are:

1) 3Y-TZP – 3mol% Yttria Tetragonal Zirconia Polycrystals
2) GTA – Glass Infiltrated Zirconia toughened Alumina
3) PSZ– Partially Stabilized Zirconia

3Y-TZP is the most commonly used type of zirconia in dentistry. The grain size (~0.2-1µm) is a critical factor in determining the mechanical properties of this type of zirconia.[26] Above a critical grain size, 3Y-TZP is less stable and more susceptible to t→m transformation. Higher sintering temperatures and longer sintering times result in a larger grain size. In situations where there is an unequal distribution of the stabilizing oxide yttira, it results in the development of the cubic phase and the tetragonal phase is depleted of yttria stabilizer ions making it less stable.[26]

There are two main methods of fabrication of a zirconia restoration.

1) Soft machining of pre-sintered zirconia blanks followed by final sintering
2) Hard machining of a fully sintered blank

Soft machining of 3Y-TZP block and sintering at a later stage reduces the stress induced t→m transformation and leads to a final surface virtually free of the undesirable monoclinic phase.[26]

Despite the advantages, several complications have been reported with the one-piece zirconia abutment. These abutments have been shown to have a lower fracture resistance as compared to titanium abutments.[30] Attempting to tighten the abutment screw prior to correct seating of the abutment generates high internal stresses which can lead to fracture.[31] Wear caused by the zirconia on the internal connection of the implant has also been reported.[32]

To address these concerns, a zirconia abutment with a titanium interface was introduced. This maintains the titanium to titanium implant abutment interface. These abutments have also been shown to have a higher fracture resistance as compared to the one-piece zirconia abutment.[33] The titanium interface may be in the form of a sleeve or a ring (Fig. 1.9) and can be either cemented (Fig. 1.8) or friction fitted (Fig. 1.7) to the zirconia abutment. While there is one study comparing the two under static load, there are no studies to show which mechanism of retention of the titanium to zirconia abutment performs better under cyclic loading.[34]

The purpose of this study was to evaluate the effect of cyclic loading on the zirconia/titanium implant abutment interface and the mechanism of failure between three different types of abutments.

The hypothesis was that there is a significant difference in the load to failure values of abutments with different mechanisms of retention of the titanium interface to the zirconia superstructure.

Fig. 1.7

Fig. 1.7 Zirconia abutment with titanium sleeve friction fit (Frft)

Fig. 1.8

Fig. 1.8 Zirconia abutment with a titanium sleeve cemented to it (Cem)

Fig. 1.9

Fig. 1.9 Zirconia abutment with a titanium ring friction fit to it (Ring frft).

Chapter 2: Materials and Methods

Fifteen Implants with their respective abutments were tested.

There were three groups and five specimens per group.

Group friction fit: This group included 5 implants (Nobel Replace, Nobel Biocare, Yorba Linda, CA, USA), RP 4.3 x 13mm with their abutments (Nobel Procera esthetic abutment, Nobel Biocare, Yorba Linda, CA, USA). The abutments consisted of a titanium insert friction fitted to a pre-fabricated zirconia superstructure. The implants were embedded in metal tubes up to the connection using a core build up material (Rock Core, Danville Materials, San Ramon, CA, USA). The abutment was torqued to 35 Ncm per the manufacturer's recommendations. Screw access holes were plugged with gauze strips and composite material (Telio® CS Inlay, Ivoclar Vivadent Inc., Amherst, NY, USA). A metal coping was checked for appropriate fit with fit checker (GC Corporation, Tokyo, Japan) and then seated using medical adhesive (Factor II, Inc., Lakeside, Arizona, USA). One specimen broke while setting up the test and had to be excluded from the study.

Fig. 2.1

Fig 2.1: Zirconia abutment with friction fit connection

Fig. 2.2

Fig. 2.2 Abutment torqued to the implant

Fig. 2.3

Fig. 2.3 Coping seated with medical adhesive

Cemented group: This group consisted of 5 implants (Straumann LLC, Andover, MA, USA) BL, RC, 4.1x12 mm and 5 abutments. Each abutment consisted of a titanium sleeve (Straumann Variobase, Straumann LLC, Andover, MA, USA) cemented to a custom zirconia abutment (milled to the dimensions of the Nobel Procera abutment), (Fig. 2.5) using a composite resin cement (Panavia™ f 2.0, Kuraray Medical Inc., Okayama, Japan). The manufacturers recommended cementation protocol was used. Similar to the implants in the friction fit group, they were embedded in the same core build up material (Rock Core, Danville Materials). The abutment was torqued to 35Ncm per the manufacturer's recommendations (Fig. 2.6). Screw access holes were plugged with a composite resin (Telio® CS Inlay, Ivoclar Vivadent). A metal coping was checked for fit with fit checker (GC Corporation), and then seated using a medical adhesive (Factor II, Inc.), (Fig. 2.7).

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Fig. 2.4

Fig. 2.4 Titanium sleeve and milled zirconia abutment

Fig. 2.5

Fig. 2.5 Titanium sleeve cemented to abutment

Fig. 2.6

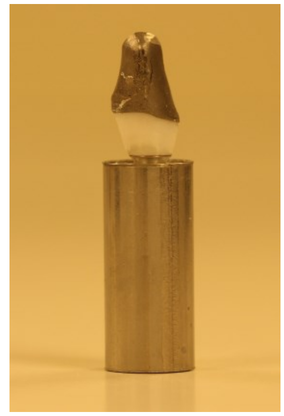

Fig. 2.7

Fig. 2.6 Torqued abutment

Fig. 2.7 Coping seated

Friction fit ring group: This group consisted of 5 implants (Zimmer TSV, Carlsbad, CA, USA) 4.1x13mm. The prefabricated zirconia abutments have a titanium ring press fit to its base with a zirconia hex (Zimmer Esthetic Abutment, Carlsbad, CA, USA), (Fig. 2.8). The implants were embedded in the same core build up material (Rock Core, Danville Materials). The sizes of the metal tubes holding them were standardized for all the specimens. The abutments were torqued to 35 Ncm per the manufacturer's recommendation (Fig. 2.9). Screw access holes were plugged with a composite resin (Telio® CS Inlay). The coping was checked for fit with fit checker (GC Corporation) and then seated with medical adhesive (Factor II, Inc.), (Fig. 2.10).

Fig. 2.8

Fig. 2.8: Zirconia abutment with Ti ring friction fit

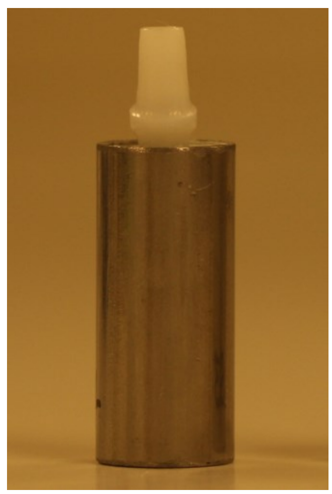

Fig. 2.9

Fig. 2.9 Torqued abutment in place

Fig. 2.10

Fig. 2.10 Coping seated.

To simulate intra-oral conditions, all 15 abutments were thermally cycled in water between 5-55^{o}C for 15,000 cycles with a 15 second dipping time and a 10 second hang time prior to torquing. The specimens shown above were mounted onto a metal jig such that the abutment was at a 30^{o} off-axis angulation. Masticatory forces were simulated using a testing machine (MTS - Machine Testing Systems). The mount simulated class I incisor relationship according to the ISO standard 14801:2007 (Fig. 2.11).

Abutments were fatigued until fracture to a maximum of 200,000 cycles at a frequency of 2 Hz (2 cycles/second).

A stepped load fatigue protocol was used. The load (N) was increased as follows 110, 148, 222, 296, 339, 415, 291, 567, 643, 720 N. Each increment occurred every 20,000 cycles.

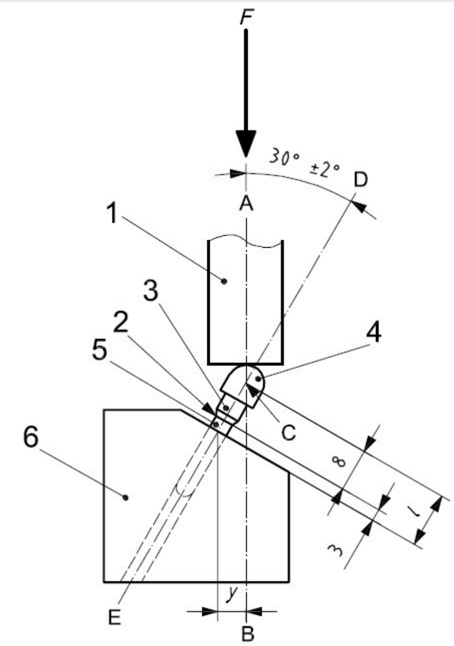

Fig.2.11: Schematic of test set up according to the ISO standard 14801:2007

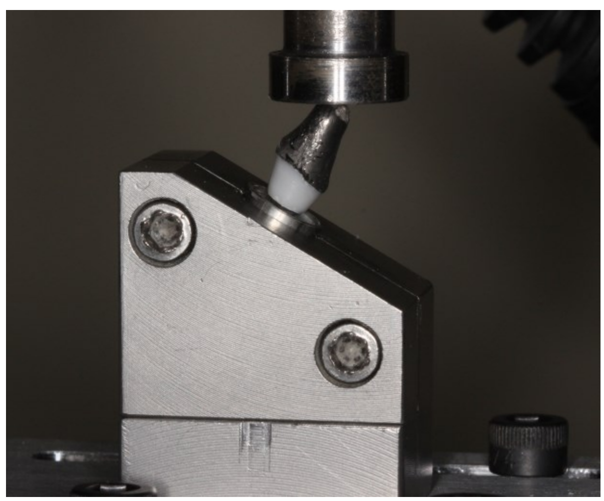

Fig. 2.12 Implant and abutment mounted on testing machine.

The one-way analysis of variance was performed using the SAS procedure MIXED (SAS Institute Inc., Cary, NC, USA).

Chapter 3: Results

The zirconia abutments tested in this experiment exhibited peak load value ranging from 261- 611N (Appendix table A.1).

The mean load to failure value for the friction fit group was 526 N with a standard deviation of 1.4 N ($p<0.0001$)

The mean load to failure value for the cemented group was 605 N with a standard deviation of 3.2N ($p<0.0001$)

The mean load to failure value for the friction fit Ti ring group was 288 N with a standard deviation of 26.7 N ($p<0.0001$)

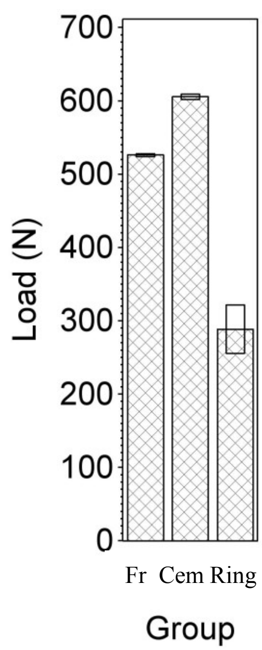

Figure 3.1: Mean load to failure values for the friction fit, cemented and friction fit ring groups.

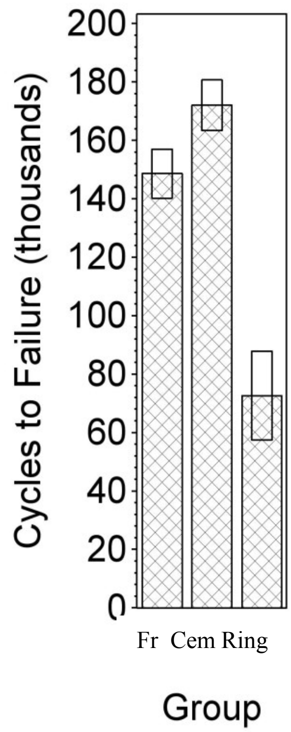

Figure 3.2: Mean number of cycles to failure for the friction fit, cemented and friction fit ring groups.

Friction Fit group: All 4 specimens, failure resulted in the zirconia abutment separating from the titanium sleeve. (Fig. 3.3)

Fig 3.3: Separation of the zirconia superstructure from the titanium insert.

Cemented group: Two modes of failure were seen. In 2 of the 5 specimens in this group, the titanium sleeve fractured at the connection to the implant (Fig. 3.4, 3.5). The remaining three fractured at the junction of the sleeve to the abutment (Fig. 3.6). No failure of the cement bond was noted.

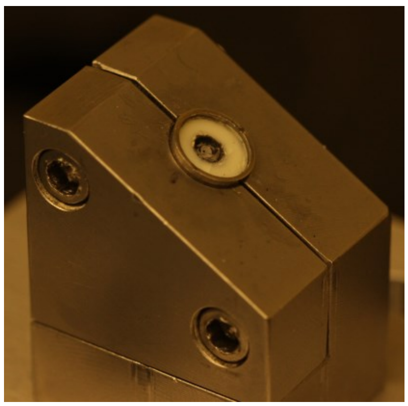

Fig. 3.4

Fig. 3.4 Fractured abutment

Fig.3.5

Fig. 3.5: Titanium fractured at the implant to abutment connection

Fig. 3.6

Fig. 3.6 Abutment fractured at the junction of the titanium sleeve and the abutment.

Friction fit ring group: Two of the five abutments fractured at the hex. For the other three specimens, the zirconia fractured at its weakest portion. Typical mode of failure was at the margin of the abutment near its thinnest portion.

Fig. 3.7

Fig. 3.7: Abutment fractured at the hex

Fig. 3.8

Fig. 3.8 Fractured at the margin of the abutment on the facial.

The Tukey-Kramer test was used to compare pairs of groups for differences of the least squares means for the load and the number of cycles (Tables 3.2 and 3.3).

Variable	Group	N	Mean	Std. Dev.	LCLM	UCLM
Load to Failure (N)	Frft	4	526.00	1.41	523.75	528.25
	Cem	5	605.40	3.21	601.42	609.38
	Ring frft	5	288.68	26.73	255.49	321.87
Number of Cycles to Failure (Thousands)	Frft	4	148.65	5.30	140.21	157.09
	Cem	5	172.04	6.99	163.36	180.71
	Ring Frft	5	72.65	12.31	57.36	87.94

Table 3.1: Load to failure and No. of Cycles to failure

Differences of Least Squares Means									
Effect	Group	Group	Estimate	Standard Error	DF	t Value	Pr > |t|	Adjustment	Adj P
Group	Frft	Cem	-79.4000	9.6648	11	-8.22	<.0001	Tukey-Kramer	<.0001
Group	Frft	Ring	237.32	9.6648	11	24.56	<.0001	Tukey-Kramer	<.0001
Group	Cem	Ring	316.72	9.1121	11	34.76	<.0001	Tukey-Kramer	<.0001

Table 3.2: Differences of Least Square Means – Load to Failure

Differences of Least Squares Means									
Effect	Group	Group	Estimate	Standard Error	DF	t Value	Pr > |t|	Adjustment	Adj P
Group	Frft	Cem	-23388	5336.57	11	-4.38	0.0011	Tukey-Kramer	0.0029
Group	Frft	Ring	76000	5336.57	11	14.24	<.0001	Tukey-Kramer	<.0001
Group	Cem	Ring	99388	5031.37	11	19.75	<.0001	Tukey-Kramer	<.0001

Table 3.3: Differences of Least Square Means – No. of Cycles

Chapter 4: Discussion

This in vitro study demonstrated that abutments with various mechanisms of retention of the titanium interface to the zirconia superstructure have different mean load to failure values and different modes of failure. The hypothesis was accepted. Even though there were significant differences in the mean load to failure values, all of them exceeded the physiologic incisal load in the anterior which is known to be approximately 90-120 N.[35,36,37] The three types of abutments used were a zirconia abutment friction fit to a titanium sleeve (Nobel Procera Esthetic Abutment) , zirconia abutment with a zirconia hex and a titanium ring press fit to its base (Zimmer Contour Esthetic Abutment – straight), and the titanium sleeve (Straumann Variobase) cemented to the zirconia superstructure with a composite resin cement (Panavia™ f 2.0) per manufacturers recommendations. The super structure for the Straumann abutment was a copy of the friction fit abutment design and was fabricated using CAD/CAM technology. The abutment with a Ti ring friction fit to its base was a prefabricated abutment and had dimensions significantly smaller than the other two groups. The dimensions of the abutments for the friction fit and cemented groups were standardized. For the ring group, l=11mm (moment arm length) Fig. 2.11. The friction fit and cemented groups had l=13mm. The ring group with a shorter moment arm fractured sooner than the other two groups. The effect of the moment arm length on the bending moment has been documented.[38,39] The longer the moment arm, the greater the bending moment. This finding can be attributed to the abutment design and dimensions. Three of the five abutments in the ring group fractured at its weakest portion which was at the margin of the abutment on the facial. The other two specimens fractured at the hex. Fracture of the zirconia hex is a common clinical complication. The presence of the titanium ring in this case was thought to maintain a titanium to titanium implant abutment interface without generating stresses at the hex. The thickness of zirconia influencing its fracture resistance has been reported. None of the abutments tested were prepared. Significant grinding on zirconia introduces microcracks which has been shown to reduce its fracture resistance.[40] For the cemented group, even though the abutments failed, there was no failure of the cement bond seen. One can speculate that the implant connection between the groups was not standardized. The friction fit abutment has a tri-lobe connection. The cemented abutment used in this study has the cross fit connection which prevents rotation through its orthogonal connection and it also has a deep conical connection. The friction fit ring abutment has a friction fit hexagonal implant to abutment connection. While it would be ideal to have the same type of connection for all the abutments, they are specific to their implant system and hence could not be standardized.

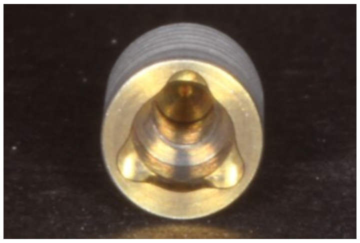

Fig. 4.1

Fig. 4.1: Tri-lobe connection

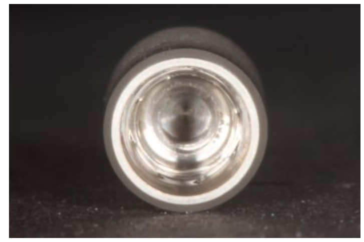

Fig. 4.2

Fig. 4.2 Deep conical connection

Fig. 4.3 Friction fit hexagonal connection.

To maintain the intimate fit of the implant to abutment the same implant system and their respective abutments were used. Also, no aftermarket abutments were used.[41] The core build up material Rock Core was used to embed the implant because it has a modulus of elasticity similar to bone.[42,43]

Oral fluids are known to facilitate stress corrosion of ceramics. Exposure to a moist environment results in slow crack growth and can lead to failure of the ceramic. To account for this being a factor for failure, the abutments were thermally cycled in water between 5-55°C for 15,000 cycles.[44]

The load applied ranged from 110N – 720N. The worst case scenarios were tested in this study. Though the bite forces in the anterior are significantly lower than the load applied in this study, there are studies to show that they can go as high as 720N in male bruxers.[45,46]

This was an in vitro study and it is difficult to draw conclusions regarding clinical situations from this. An attempt was made to simulate the intraoral environment as much as possible. However, further clinical studies are required to confirm these findings.

Chapter 5: Conclusion

Within the limitations of this study, the following conclusions can be drawn.

1) There was a statistically significant difference ($p<0.05$) in the mean load to failure values between the pairs of groups.
2) The abutments in the cemented group showed the highest resistance to fracture.
3) The abutments in the friction fit ring group showed the lowest resistance to fracture.
4) The mode of failure for all three abutments was different.

References

1. Brånemark PI. Osseointegration and its experimental background. J Prosthet Dent. 1983;50(3):399–410.

2. Adell R, Lekholm U, Rockler B, Brånemark PI. A 15-year study of osseointegrated implants in the treatment of the edentulous jaw. Int J Oral Surg. 1981;10(6):387–416.

3. Adell R, Eriksson B, Lekholm U, Brånemark PI, Jemt T. Long-term follow-up study of osseointegrated implants in the treatment of totally edentulous jaws. Int J Oral Maxillofac Implants. 1990;5(4):347–59.

4. Branemark PI, Adell R, Albrektsson T, Lekholm U, Lundkvist S, Rockler B. Osseointegrated titanium fixtures in the treatment of edentulousness1. Biomater Silver Jubil Compend. 2006;4:17–20.

5. Jemt T. Modified single and short-span restorations supported by osseointegrated fixtures in the partially edentulous jaw. J Prosthet Dent. 1986;55(2):243–7.

6. Fugazzotto Pa, Ryan Dr. Biology of Implant Esthetics: Tooth Replacement in the Anterior Maxilla. J Esthet Restor Dent. 1997 Sep;9(5):248–54.

7. Jemt T, Lekholm U, Adell R. Osseointegrated Implants in the Treatment of Partially Edentulous Patients : A Preliminary Study on 876. 1997;(April 1968).

8. Henry PJ, Tolman DE, Bolender C. The applicability of osseointegrated implants in the treatment of partially edentulous patients: three-year results of a prospective multicenter study. Quintessence Int. 1993;24(2):123–9.

9. Holst S, Blatz MB, Hegenbarth E, Wichmann M, Eitner S. Prosthodontic considerations for predictable single-implant esthetics in the anterior maxilla. J Oral Maxillofac Surg. 2005;63:89–96.

10. Hultin M, Gustafsson a, Klinge B. Long-term evaluation of osseointegrated dental implants in the treatment of partly edentulous patients. J Clin Periodontol. 2000;27(2):128–33.

11. Sadan A, Blatz MB, Salinas TJ, Block MS. Single-implant restorations: A contemporary approach for achieving a predictable outcome. J Oral Maxillofac Surg. 2004;62:73–81.

12. Vigolo P, Vigolo P, Givani MA, Majzoub DDSZ. single-tooth crowns : a 4-year prospective clinical Cemented Versus Screw-Retained A 4-year Prospective Clinical Study.

13. Chee W, Jivraj S. Screw versus cemented implant supported restorations. Br Dent J. 2006;201(8):501–7.

14. Lee A, Okayasu K, Wang H-L. Screw- versus cement-retained implant restorations: current concepts. Implant Dent. 2010;19(1):8–15.

15. Hebel KS, Gajjar RC. Cement-retained versus screw-retained implant restorations: Achieving optimal occlusion and esthetics in implant dentistry. J Prosthet Dent. 1997;77(1):28–35.

16. Sghaireen MG. Fracture resistance and mode of failure of ceramic versus titanium implant abutments and single implant-supported restorations. Clin Implant Dent Relat Res. 2013;1–8.

17. Prestipino V, Ingeber A. Esthetic High-Strength Implant Abutments. Part I. J Esthet Restor Dent. 1993 Jan 1;5(1):29–36.

18. Prestipino V, Ingeber A. Esthetic High-Strength Implant Abutments. Part II. J Esthet Restor Dent. 1993 Mar;5(2):63–8.

19. Prestipino V, Ingeber A. All-Ceramic Implant Abutments: Esthetic Indications. J Esthet Restor Dent. 1996 Jan;8(1):255–62.

20. Nakamura T, Saito O, Mizuno M, Kinuta S, Ishigaki S. Influence of abutment substrates on the colour of metal-free polymer crowns. J Oral Rehabil. 2003;30(2):184–8.

21. Brodbeck U. The ZiReal Post: A new ceramic implant abutment. J Esthet Restor Dent. 2003;15(1):10–23.

22. Butz F, Heydecke G, Okutan M, Strub JR. Survival rate, fracture strength and failure mode of ceramic implant abutments after chewing simulation. J Oral Rehabil. 2005;32(11):838–43.

23. Glauser R, Sailer I, Wohlwend A, Studer S, Schibli M, Schärer P. Experimental zirconia abutments for implant-supported single-tooth restorations in esthetically demanding regions: 4-year results of a prospective clinical study. Int J Prosthodont. 2004;17(3):285–90.

24. Sadan A, Blatz MB, Lang B. Clinical considerations for densely sintered alumina and zirconia restorations: part 2. Int J Periodontics Restorative Dent. 2005;25(4):343–9.

25. Piconi C, Maccauro G. Zirconia as a ceramic biomaterial. Biomaterials [Internet]. 1999 Jan;20(1):1–25.

26. Denry I, Kelly JR. State of the art of zirconia for dental applications. Dent Mater. 2008;24(3):299–307.

27. Marshall D, Evans A, Drory M. Transformation toughening in ceramics [Internet]. Fracture mechanics of 1983.

28. Ruf H, Evans AG. Toughening by Monoclinic Zirconia. J Am Ceram Soc. 1983;66(5):328–32.

29. Marshall D, Evans A, Drory M. Transformation toughening in ceramics. Fract Mech. 1983;70(10):689–98.

30. Foong JKW, Judge RB, Palamara JE, Swain M V. Fracture resistance of titanium and zirconia abutments: An in vitro study. J Prosthet Dent. The Editorial Council of the Journal of Prosthetic Dentistry; 2013;109(5):304–12.

31. Aboushelib MN, Salameh Z. Zirconia implant abutment fracture: clinical case reports and precautions for use. Int J Prosthodont. 2009;22(6):616–9.

32. Cavusoglu Y, Akca K, Gurbuz R, Cehreli MC. A Pilot Study of Joint Stability at the Zirconium or Titanium Abutment/Titanium Implant Interface. Int J Oral Maxillofac Implant. 2014;29(2):338–43.

33. Stimmelmayr M, Sagerer S, Erdelt K, Beuer F. In vitro fatigue and fracture strength testing of one-piece zirconia implant abutments and zirconia implant abutments connected to titanium cores. Int J Oral Maxillofac Implant. 2013;28(2):488–93.

34. Kim JS, Raigrodski AJ, Flinn BD, Rubenstein JE, Chung KH, Mancl L a. In vitro assessment of three types of zirconia implant abutments under static load. J

Prosthet Dent. The Editorial Council of the Journal of Prosthetic Dentistry; 2013;109(4):255–63.

35. Gibbs CH, Mahan PE, Mauderli a, Lundeen HC, Walsh EK. Limits of human bite strength. J Prosthet Dent. 1986;56(2):226–9.

36. Waltimo a, Könönen M. A novel bite force recorder and maximal isometric bite force values for healthy young adults. Scand J Dent Res. 1993;101(3):171–5.

37. Richter EJ. In vivo vertical forces on implants. Int J Oral Maxillofac Implants. 1997;10(1):99–108.

38. Truninger TC, Stawarczyk B, Leutert CR, Sailer TR, Hämmerle CHF, Sailer I. Bending moments of zirconia and titanium abutments with internal and external implant-abutment connections after aging and chewing simulation. Clin Oral Implants Res. 2012;23(1):12–8.

39. Mühlemann S, Truninger TC, Stawarczyk B, Hämmerle CHF, Sailer I. Bending moments of zirconia and titanium implant abutments supporting all-ceramic crowns after aging. Clin Oral Implants Res. 2014;25(1):74–81.

40. Alqahtani F, Flinton R. Postfatigue fracture resistance of modified prefabricated zirconia implant abutments. J Prosthet Dent. The Editorial Council for the Journal of Prosthetic Dentistry; 2014;112(2):299–305.

41. Yilmaz B, Salaita LG, Seidt JD, Clelland NL, McGlumphy EA. Load to failure of different titanium abutments for an internal hexagon implant. J Prosthet Dent. 2015 Jun 3.

42. Rho JY, Ashman RB, Turner H. Young's modulus of trabecular and cortical bone material: Ultrasonic and microtensile measurements. J Biomech. 1993;26(2):111–9.

43. Passos SP, Freitas AP, Jumaily S, Santos MJMC, Rizkalla AS, Santos GC. Comparison of mechanical properties of five commercial dental core build-up materials. Compend Contin Educ Dent. 2013;34(1):62–3, 65–8.

44. Daou EE. The zirconia ceramic: strengths and weaknesses. Open Dent J. 2014;8:33–42.

45. Gibbs CH, Anusavice KJ, Young HM, Jones JS, Esquivel-Upshaw JF. Maximum clenching force of patients with moderate loss of posterior tooth support: A pilot study. J Prosthet Dent. 2002;88(5):498–502.

46. Lamey PJ, Burnett C a., Fartash L, Clifford TJ, McGovern JM. Migraine and masticatory muscle volume, bite force, and craniofacial morphology. Headache. 2001;41(1):49–56.

Appendix A: Fracture values of Abutments

Group	Specimen	Load (N)	No. of cycles
Z	1	330	80070
	2	280	60010
	3	261	60450
	4	298.4	88110
	5	274	74610
N	1	525	142230
	2	528	146370
	3	526	153150
	4	-	-
	5	525	152850
S	1	603	164730
	2	611	164210
	3	605	177770
	4	604	178050
	5	604	175430