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Title: Effect of cantilever length on the fracture resistance of full-arch implant-supported monolithic zirconia prostheses

Running Title: Effect of cantilever length on full arch zirconia prostheses.

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ABSTRACT:

Purpose: The purpose of this study was to test the fracture resistance of screw-retained, implant-supported, monolithic zirconia prostheses with increasing distal cantilever lengths. Three different cantilever lengths based on different anterior-posterior (AP) spread to cantilever ratios were tested.

Materials and Methods: Custom titanium jig was fabricated, and two 4.3 mm regular platform implant analogs (Nobel Biocare/Yorba Linda, CA) positioned and soldered onto the jig. Nobel multiunit scanbodies (Imagine/Virginia, USA) were placed on the implants and scanned with a laboratory optical scanner (Zirkonzahn/Gais, Italy). Half-arch monolithic zirconia prostheses of three different AP spread to cantilever ratios of 1:1, 1:1.5, and 1:2 were designed with Zirkonzahn software (Zirkonzahn/Gais, Italy). Thirty samples were milled and sintered in e.max ZirCAD LT zirconia (Ivoclar Vivadent/Amherst, NY) with ten prostheses per group. The prostheses were bonded to Nobel NMU multiunit ti-bases (Imagine/Virginia, USA) with Panavia 21 resin cement (Kuraray Dental/New York, NY). A universal testing machine (Instron/Norwood, MA) machine was then used to load the sample until any component of the restoration failed. The process was repeated for each of the sample.

Results: The highest mean load to fracture was found in the 1:1 AP spread to cantilever ratio group at 698.51 ±72.387 N, followed by the 1:1.5 AP spread to cantilever ratio group at 423.48 ±92.389 N, and lastly by the 1:2 AP spread to cantilever ratio group at 268.32 ±31.733 N. The 275 N greater fracture load of the 1:1 group compared to the 1:1.5 group, and the 430 N greater fracture load of the 1:1 group compared to the 1:2 group, were statistically significant ($p < 0.001$). The 155 N greater fracture load of the 1:1.5 group compared to the 1:2 group was also statistically significant ($p < 0.001$).

Conclusions: Within the limitations of the study, the following conclusions can be made:

1. Failure of samples due to zirconia fracturing did not relate to the cantilever length
2. The rate of failure of samples due to ti-bases debonding from zirconia and prosthetic screw breaking was more significant had a significant positive correlation to cantilever lengths

INTRODUCTION

Over the past several decades, implants have been proven to be highly successful to restore dental function and esthetics. [1] For those patients with complete edentulism, an implant-supported complete fixed dental prosthesis (ICFDP) has become an attractive treatment option. Conventionally, full arch prostheses are fabricated using a combination of acrylic denture teeth with a metal framework. This method has been shown to be predictable with 99% prosthetic stability for over 15 years in the mandibular arch. [2] However, this type of restoration is not free of prosthetic complications such as acrylic resin veneer fractures or debonding of denture teeth.[3, 4, 5]

Monolithic zirconia has been introduced as an answer to minimize prosthetic failures for ICFDP's. Zirconia, which is the crystalline dioxide form of zirconium, can have the appearance of natural teeth while demonstrating mechanical properties similar to that of metal. [6] In vitro studies of zirconia demonstrate many desirable physical properties such as a flexural strength of 900 to 1200 MPa, a fracture toughness of 9 to $10\frac{MPa}{m}$, a modulus of elasticity similar to that of stainless steel alloys and a compressive strength that can reach 2000 MPa. [6, 7, 8, 9]. It is a polycrystalline ceramic that can exist in three major phases depending on temperature: monoclinic (room temperature to 1170°C), tetragonal (1170-2370°C), and cubic (above 2370 °C). [9] To stabilize and maintain the crystalline structure at lower temperatures, yttria (Y_2O_3) is commonly added to zirconia as an element to alter its mechanical properties. [10] The resulting yttria-stabilized zirconia in its tetragonal phase is prevented from reverting back to its weaker monoclinic phase at room or intraoral temperature by allowing more of an ideal fluorite lattice molecular structure. [11] Another property that makes zirconia a favorable dental restorative material is a phenomenon known as transformation toughening. This occurrence is due to induced phase transformation at a point of stress, such as a crack on the ceramic surface. When the stress level reaches a critical value, zirconia particles at the crack tip undergo transformation from the tetragonal to the monoclinic phase, which is accompanied by a volumetric increase of 4.9%. [12] This increase induces compressive stress in the crack tip. As a result, the stress intensity factors at the crack tip are reduced. Therefore, it takes higher applied loads to increase the stress intensity factor back to the critical level, retarding continued crack propagation and enhancing the fracture toughness of the ceramic. [13]

One of the most important considerations when designing zirconia prostheses is the anterior-posterior (AP) spread of the implants. AP spread is regarded as the distance measured from a line drawn through the center of the two most anterior implants to a line drawn through the distal of the two most terminal implants in a completely edentulous arch. The amount of cantilever allowed distal to terminal implants is then determined by multiplying a factor ranging from 1.5 to 2.5. [14, 15] Rangert provides simple guidelines and proposes a cantilever length of up to 20 mm for an AP spread of 10 mm (twice the AP spread) to control occlusal loads on implants and prosthetic reconstruction on mandibular ISFPs. [14] English proposed that posterior cantilever in mandibular ISFPs should be no more than 1.5 times the AP spread, allowing a 10-12 mm posterior cantilever for the mandible and a reduced length of 6-8mm in the maxilla.[15] Cooper proposed another acceptable AP spread, incorporating it into his “Rules of 10”. [16] These rules address factors affecting implant and prosthesis longevity, including magnitude of forces, resistance of the prosthesis against these forces, and the biology of bone and its ability to respond to loading environments. [16] The first two rules describes the minimum requirements for implant length and restorative space, with both needing to be 10 mm. [16] In his third rule, Cooper asserts that the anterior-posterior distribution of the implants must be greater than 10 mm for an implant-supported fixed prosthesis (ISFP).[16] The concept of distal cantilevers needs to be considered during the treatment planning stages of long span prostheses due to their importance in the overall success of the treatment. Cantilevers distal to terminal implants can cause mechanical complications such as fracture of the restoration, fracture of abutment screws, and screw loosening.[25]

A review of the literature revealed that mechanical complications, such as cantilever and abutment fractures, is a potential problem for computer-aided design and computer-aided manufactured (CAD/CAM) screw-retained, implant-supported monolithic zirconia prostheses.[17, 18] However, studies describing the effect of increased cantilever lengths on implant supported fixed prostheses in edentulous arches are lacking. A retrospective study by Vizcaya analyzed twenty CAD/CAM double full-arch monolithic zirconia fixed prostheses (MZ-FP) and measured the average extension cantilever segments.[19] The author found no implant failures, no changes in the occlusal surface, no chipping of the cuspid or incisal edges, and no distal extension fractures during follow-up periods ranging from 2 to 7 years. Although average values of cantilever lengths and zirconia thickness at SAOs were obtained based on successful prostheses after 2-7 years follow up, the author stated that additional in vitro and clinical studies will be required for more scientific analysis of the criteria for design of this type of prosthesis to minimize prosthetic complications.

The purpose of this study is to test the fracture resistance of screw-retained, implant-supported, monolithic zirconia prostheses with increasing distal cantilever lengths. This will provide a guideline on how far one can predictably cantilever distally from the most distal implant screw access opening before catastrophic failure occurs. The null hypothesis is that there will no difference in fracture resistance at increasing cantilever lengths.

MATERIALS AND METHODS

An Instron universal testing machine (Instron/Norwood, MA) was used to test the effect of cantilever length on pre-determined zirconia thickness at the distal screw access opening. The specimens were loaded until failure. The pre-determined zirconia thickness at the distal SAO and the height of the connector were based on a previous study by Vizcaya.[19] To mimic the average values of the mandibular prostheses as described in this study, height of the specimens were set to 11.89 mm and the widths were set to the values in Table 1.

The zirconia tested was e.max ZirCAD LT (Ivoclar Vivadent/Amherst, NY). Two 4.3 mm implant diameter analogs (Nobel Biocare/Yorba Linda, CA) positioned 20 mm apart to mimic implants in the lateral incisor and second premolar areas. A ResinRock type IV stone (Whipmix/Louisville, KY) model was made that included two 4.3 mm implant analogs (Nobel Biocare/Yorba Linda, CA) situated 20 mm apart. An STL file of the stone model was obtained using a laboratory optical scanner (Zirkonzahn/Gais, Italy) and used to communicate the positions of the implant analogs for fabrication of a titanium (Ti6AL4V) standardized master model for testing (Figure 1). An implant analog was scanned separately to obtain its exact dimension and size, allowing space in the titanium model for the analogs to be placed. The analogs were soldered to the titanium master model. Straight regular platform (RP) multiunit abutments (Nobel Biocare/Yorba Linda, CA) were placed on the implant analogs and torqued to 15Ncm using a torque wrench (Nobel Biocare/Yorba Linda, CA). Implant scanbodies (Imagine/Virginia, USA) were then attached to the multiunit abutments and the model was scanned using an optical scanner (Zirkonzahn/Gais, Italy). Samples for three different groups of AP spread to cantilever length ratios were designed--1:1, 1:1.5, and 1:2, utilizing digital design software (Zikonzahn/Gais, Italy) as shown in Figure 2.

Ten samples per cantilever length totaling 30 monolithic zirconia prostheses were dry milled with the PrograMill PM7 (Ivoclar Vivadent/Amherst, NY) and sintered with the Zenotech Fire P1 (Wieland Dental/ Pforzheim, Deutschland) as per the manufacturer's specification by a single technician. Struts that were designed, milled, and sintered to support the long-span zirconia were then removed with diamond discs and polished with Dialite ZR discs (Brasseler/Savannah, GA). Nobel NMU multiunit ti-bases (Imagine/Virginia, USA) were bonded to the intaglio surface of the prostheses with Panavia 21 resin cement (Kuraray Dental/New York, NY) after air abrading the zirconia and multiunit abutments with aluminum oxide, applying Z Prime for zirconia, and applying Monobond Plus for the abutments. The specimens were torqued to the testing models with prosthetic screws (Imagine/Virginia, USA) to 15 Ncm using a torque wrench (Nobel Biocare/Yorba Linda, CA). Mounted and fixed in a universal testing machine (Instron/Norwood, MA), the specimens were subjected to load to failure tests, with the load (in Newtons, N) axially applied over the center of the cantilever with a 5 mm round indenter, 1 mm away from the distal terminus of the sample (Figure 3).

The load to failure for each specimen was confirmed via Bluehill Universal software (Instron/Norwood, MA) which has a set displacement figure to note when failure occurs. When a specimen failed, the zirconia samples were removed, and the testing model cleaned in an ultrasonic to prepare for the next specimen.

RESULTS

SPSS statistical software was used for all statistical analyses. The result is summarized in Table 3, which includes the mean load to fracture, upper bound and lower bound confidence interval, standard deviation, minimum and maximum load to fracture. The highest mean load to fracture was found in the 1:1 AP spread to cantilever ratio group at 698.51 ±72.387 N, followed by the 1:1.5 AP spread to cantilever ratio group at 423.48 ±92.389 N, and lastly by the 1:2 AP spread to cantilever ratio group at 268.32 ±31.733 N. The mode of failure is described in Table 2.

In this study there was one discrete independent variable (three categories of cantilever lengths), and one dependent continuous variable (fracture load). Therefore, a one-way ANOVA statistical analysis was selected to determine if the fracture load for zirconia prostheses on titanium-bases was significantly different for groups with different AP spread to cantilever ratios (Table 4). There were no outliers, as assessed by a boxplot (Figure 4). Data from all three groups showed a normal distribution as assessed by a Shapiro-Wilks test ($p>0.05$) (Table 6). The Levene test revealed that the data did show equality of variances across all samples. (Table 7, $p>0.05$) Because of the equal variance in the data, a Scheffe post hoc analysis was conducted. (Table 8). This revealed that the 275 N greater fracture load of the 1:1 group compared to the 1:1.5 group, and the 430 N greater fracture load of the 1:1 group compared to the 1:2 group, were statistically significant ($p < 0.001$). The 155 N greater fracture load of the 1:1.5 group compared to the 1:2 group was also statistically significant ($p < 0.001$).

DISCUSSION

Due to anatomical limitations, more specifically lack of posterior bone for implants, distal cantilevers have widely been used in full-arch implant-supported fixed prostheses. Previous treatment modalities involved the use of either casted or milled alloy framework as the substructure of choice. With the advent and advances in ceramic materials, metal-free, monolithic zirconia has found its way into the prosthodontist's armamentarium to replace alloys in these prostheses. However, studies are lacking on the survival and success rates of monolithic zirconia long span prostheses, especially as it relates to posterior cantilever lengths. As a result, there is limited scientific evidence for how far a distal cantilever can be made before prosthetic failure occurs. Therefore in this study, three different AP spread to cantilever ratios (1:1. 1:1.5, 1:2) were compared. Material thickness, connector sizes, and the location of the screw hole access were standardized. For the purpose of this study, half-arch prostheses extending from mandibular central incisor to second molar were with SAOs through the lateral incisor and second premolar areas. These prostheses were used to conduct testing unilaterally instead of fabricating full-arch prostheses to test both sides separately. This decision is to reduce possible confounding factors on the non-testing side while applying load on the testing side. The upper limit of the force applied was set at 900N to incorporate maximum force one can generate on their posterior teeth. The maximum voluntary force that can be generated by the elevator muscles during clenching ranges from 300N to 600N in the molar region for healthy adults with natural dentition.[20, 21, 22] Some studies even report that maximum biting force can reach approximately 800N in the molar region.[23, 24]

The expectation in this study was to find prosthetic failure at the zirconia level, more specifically at the most distal screw access opening one would find zirconia fracturing at the due to the material being thinner and the distal implant working as the fulcrum of the lever, with tensile forces concentrated as the cantilever is loaded. Therefore, it was the researchers' expectation that the failure of the samples would be at the screw access opening. However, in this study, none of the 30 zirconia samples fractured from loading to failure. Instead, the mode of failure was from a combination of either the mesial multiunit ti-bases debonding from the zirconia samples or the prosthetic screw breaking. As a result, no correlation could be found between zirconia failure and AP spread to cantilever ratio.

As aforementioned, failures did occur at the ti-base level. From the ti-base level, the results reveal statistically significant differences in load to failure. When comparing the mean value when the samples failed in each group, there was a statistically significant difference that positively correlated with the cantilever lengths. From the results, it posited that more force is applied to the prosthesis and its supporting parts as the distance between the fulcrum and the loading point is increased.

Some authors describe potential screw fracture or loosening due to increased cantilever lengths. [24] Study by McAlarney and Stavropoulos also stated that loads applied to implants increased as cantilevers on full arch prostheses increased, which could then also lead to increased prosthetic complications. [26] However, while many report acrylic teeth and denture base fracture as commonly observed technical complication, screw fracturing is not reported as frequently. [27] Especially comparing with this study, which had a very high percentage of prosthetic failure due to screw fracture, the results do not agree with what is reported in literature. One possible explanation for the difference is the nature of the study. This in vitro study aimed to observe how half-arch prostheses would fail under unilateral force on the cantilevers. The clinical studies on the other hand would observe how full-arch prostheses would perform in clinical setting.

There are also limitations of this study that possibly impacted the results. One is that the samples did not undergo cyclic loading or thermocycling. The environment of this study, due to such reasons, was not able to replicate the oral environment with various factors that could impact the performance of zirconia samples. Since the samples were loaded until failure and not loaded cyclically, which would attempt to mimic the intraoral masticating pattern, it would be difficult to gather direct clinical relevance from this in vitro study. Furthermore, due to lack of thermocycling as part of the sample preparation, low temperature degradation on zirconia could not be evaluated as part of the result. As such, clinical performance may differ from the result of this study.

CONCLUSIONS

The following conclusions can be made within the limitations of the study:

1. Failure of samples due to zirconia fracturing did not relate to the cantilever length.
2. The rate of failure of samples due to ti-bases debonding from zirconia and prosthetic screw breaking was more significant and had a significant positive correlation to cantilever lengths.

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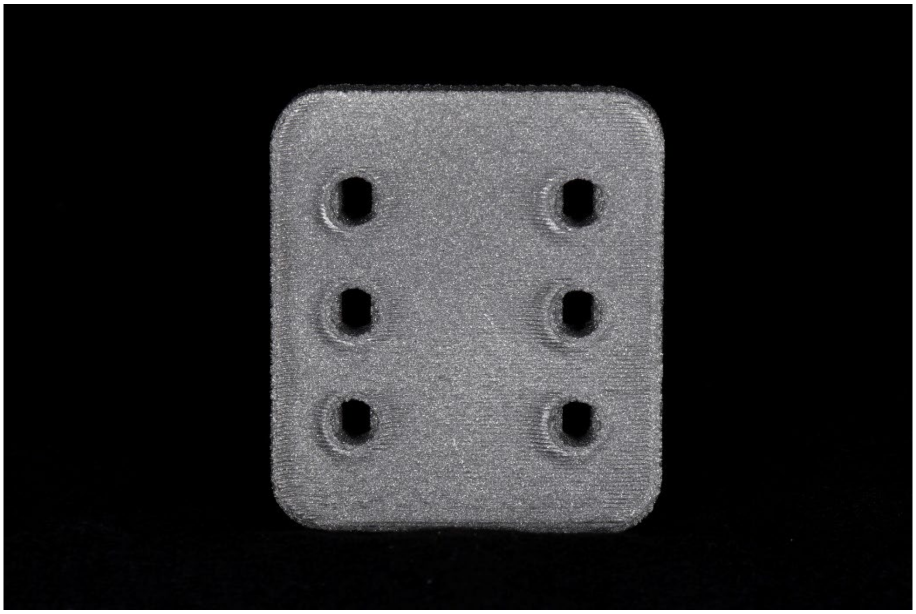

Figure 1. Titanium model fabricated using an Arcam A1 (Arcam AB/Montreal, Canada) from Walter Reed National Military Medical Center

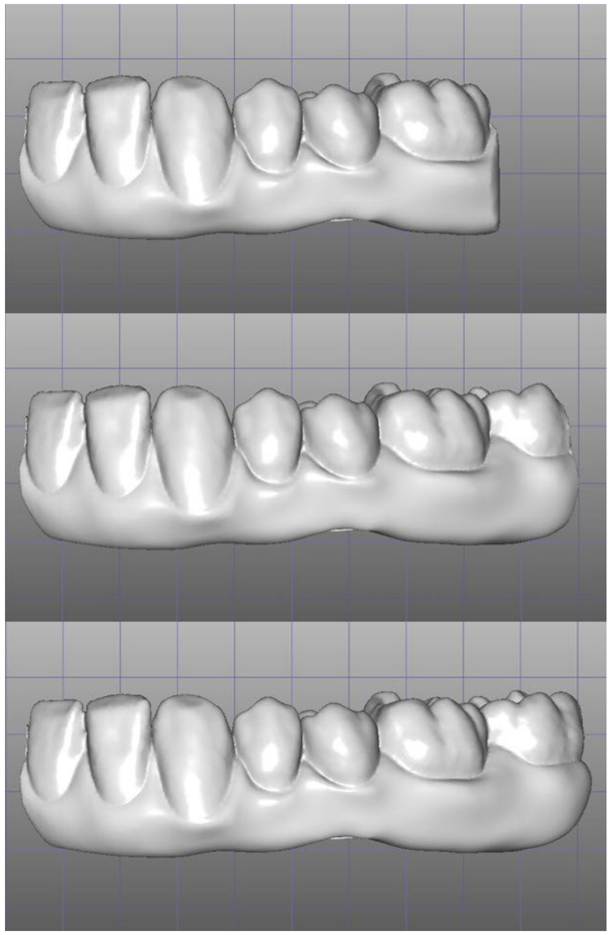

Figure 2. Sample designed with digital design software (Zikonzahn/Gais, Italy)

	Buccal (mm)	Lingual (mm)
Mid-crown	3.56	2.07
Margin	3.08	2.00
Mid-gingival	3.15	2.99

Table 1. The average zirconia thickness around distal SAO at three different levels of mandibular prostheses (mid-crown, margin, mid-gingival)

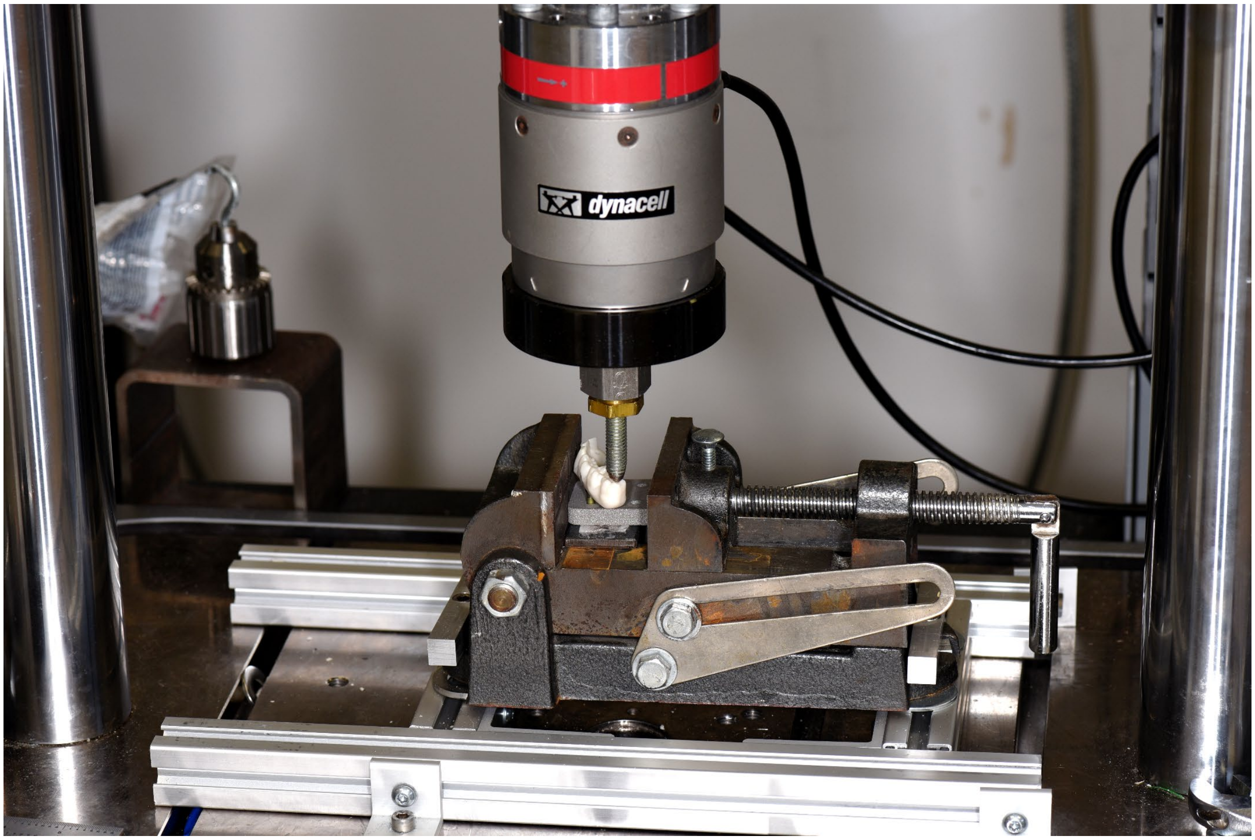

Figure 3. Samples loaded on a universal testing machine (Instron/Norwood, MA)

	Mesial	Distal
1:1	50% screw, 50% cement	100% screw
1:1.5	30% screw, 70% cement	100% screw
1:2	70% screw, 30% cement	100% screw

Table 2. Mode of failure

1:1	Mean	698.51
	Lower bound for 95% confidence interval for mean	646.73
	Upper bound for 95% confidence interval for mean	750.30
	Standard deviation	72.387
	Minimum	619
	Maximum	845
1:1.5	Mean	423.48
	Lower bound for 95% confidence interval for mean	357.39
	Upper bound for 95% confidence interval for mean	489.57
	Standard deviation	92.389
	Minimum	260
	Maximum	538
1:2	Mean	268.32
	Lower bound for 95% confidence interval for mean	245.62
	Upper bound for 95% confidence interval for mean	291.02
	Standard deviation	31.733
	Minimum	230
	Maximum	320

Table 3: Descriptive Data

	Sum of Squares	df	Mean Square	F	Sig.
Between Groups	949290.942	2	474645.471	96.325	.000
Within Groups	133043.236	27	4927.527		
Total	1082334.18	29			

Table 4: One way ANOVA

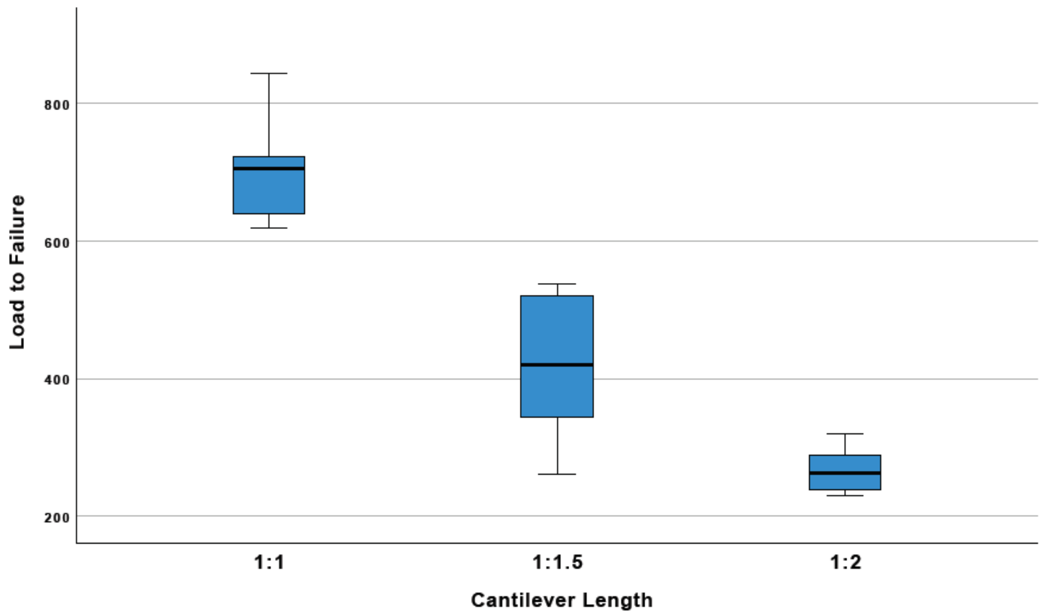

Figure 4. Boxplot

	Statistic	df	Sig.
1:1	.905	10	.246
1:1.5	.938	10	.532
1:2	.924	10	.391

Table 6: Test of Normality (Shapiro-Wilks)

		Levene Statistic	df1	df2	Sig.
Load to Failure	Based on Mean	3.179	2	27	.058
	Based on Median	3.134	2	27	.060
	Based on Median and with adjusted df	3.134	2	20.249	.065
	Based on trimmed mean	3.258	2	27	.054

Table 7: Levene Test for Homogeneity of Variances

Cantilever Length (a)	Cantilever Length (b)	Mean Difference (a-b)	Std. Error	Sig.	Lower Bound 95% Confidence	Upper Bound 95% Confidence
1:1	1:1.5	275.037*	31.393	.000	193.73	356.34
	1:2	430.195*	31.393	.000	348.89	511.50
1:1.5	1:1	-275.037*	31.393	.000	-356.34	-193.73
	1:2	155.158*	31.393	.000	73.85	236.47
1:2	1:1	-430.195*	31.393	.000	-511.50	-348.89
	1:1.5	-155.158*	31.393	.000	-236.47	-73.86

*. The mean difference is significant at the 0.05 level.

Table 8: Post-Hoc Test (Scheffe)