# THE LOAD EXERTED BY REMOVABLE PARTIAL DENTURE CLASPS CONFORMING TO AVERAGE TOOTH CURVATURES, IN CLINICALLY ENCOUNTERED UNDERCUTS.


Noland Naidoo

Student number: 586244


A research report submitted to the Faculty of Health Sciences, University of the Witwatersrand, Johannesburg, in partial fulfilment of the requirements for the degree of Master of Dentistry in the branch of Prosthodontics.

Johannesburg, 2016

# DECLARATION

I, Noland Naidoo declare that this research report is my own work. It is being submitted in partial fulfilment of the degree of Master of Dentistry in the branch of Prosthodontics at the University of the Witwatersrand, Johannesburg. It has not been submitted before for any degree or examination at any other University.

Noland Naidoo

12th day of May, 2046 in Johannesburg

# ABSTRACT

**Purpose:** The aim of this study was to establish clinically relevant guidelines for the selection of clasps for removable partial dentures (RPDs) by adapting RPD clasp arms of differing materials and diameter to a three-dimensional model based on the average curvature of premolars and molars.

**Method:** Randomly selected discarded casts were collected that had intact first premolars and first molars. The normal clasp position for the buccal surfaces of these teeth was drawn on the cast and the teeth were then sectioned to this line, and scanned using a flatbed scanner. The average curvature and length for each group was determined using a graphics-drawing programme and a 3D model was printed using these data. Clasp materials were then be adapted to these models using the wrought wires: Leowire® (Leone, Fiorentino, Italy) (stainless steel); Remanium Hard® (Dentaurum, Pforzheim, Germany) (stellite alloy of chromium and cobalt); Noninium® (Dentaurum, Pforzheim, Germany) (stainless steel, nickel free); and cast clasps cast in the stellite alloy Vitallium (Dentsply, Ontario, Canada). Two diameters of the wrought wire clasp groups were used (0.9mm and 1.0mm). Ten samples for each diameter and material were adapted to the 3D models as they would in a clinical case. Each clasp was then randomly deformed beyond its proportional limit in a tensile testing machine (Instron Corporation, High Wycombe, United Kingdom) and the forces exerted at that limit and at deflections of 0.25 mm, 0.5 mm, and 0.75 mm were measured, as these are the clinically encountered undercuts. Statistical analysis was carried out to determine if the forces exerted at these deflections were within the proportional limit of the clasp, but also significantly within the realistic limit (defined as two standard deviations from the proportional limit) to allow for variations in manufacture and application of the clasp arms.

**Results**: The results confirmed the effect of material, length, and diameter on flexibility for the wrought wires. A table was produced with guidelines for those clasps which could be used safely within their realistic limits and therefore should provide longevity of service. The greatest force exerted on premolar clasp length arms was provided by Leowire at 0.25mm undercut (606g); although it was 2% greater than its realistic limit, this was considered sufficiently within the safety factor to recommend its clinical use. The greatest force exerted on molar clasp length arms was provided by Remanium Hard at a deflection of 0.5 mm (417g). Cast clasps for premolars should not be prescribed as they were all well above their realistic limits. Cast clasps should only be selected for molars if the clasp arm is longer than 14.5mm. At a deflection of 0.25 mm the cast arm exerted a force of 773g but whilst this was below its proportional limit, it was higher than its realistic limit.

**Conclusions:** The results of this study provide valid guidelines for the clinical application of clasp arms in removable partial dentures. It is recommended that clinical studies be carried out to confirm the longevity of clasp arms based on these data.

# ACKNOWLEDGEMENTS

I would like to thank my supervisor and mentor, Professor CP Owen for his support, and assistance throughout the development of this experiment and in the writing of this report.

Professor DG Howes for his supervision, guidance and assistance as well as with the printing of the 3D models.

Mrs. Petra Gaylard for her advice and statistical analysis of the results.

Mr. L Cummings for his assistance with the CAD/CAM software and 3D models

Mr. Hiren Canjee for his assistance in the preparation of the clasps samples used in this study.

Dr LM Naidoo for his advice and for sharing his experience from his study.

# TABLE OF CONTENTS

# LIST OF FIGURES

# LIST OF TABLES

# CHAPTER 1 INTRODUCTION AND LITERATURE REVIEW

## 1.1 INTRODUCTION

Tooth loss has a multifactorial aetiology and can have varying effects on an individual's quality of life (Koller et al., 2011). The current trend in the epidemiological literature suggests that there are a decreasing number of patients afflicted with complete edentulism (Chatzivasileiou et al., 2015). However, tooth loss continues to occur for a variety of reasons, and combined with an ageing population, is likely to result in clinicians having to treat a higher incidence of partially edentulous patients.

The treatment modalities available to partially dentate patients include removable partial dentures (RPDs), fixed partial dentures (FPDs), implant supported prostheses (ISPs) or to do nothing. The last option is, however, is dependent on the patient's ability to adequately function in their current state and/or their preferences.

The replacement of missing teeth by means of RPDs is a well-established treatment modality and is a cost-effective option of providing appropriate treatment. However, if the RPD is not well designed, it may cause more harm than good. Therefore, RPDs should be designed from a "bio-functional" perspective (Warr, 1961; Frank & Nicholls, 1981; Owen, 2000).

The prescription of an RPD as a component of a larger plan of treatment must be based on sound principles that aim to maximise retention, support and stability. The responsibility of designing an RPD rests solely with the clinician and a detailed diagrammatic prescription must mandatorily be provided to the dental technician (Sharry, 1977; Owen, 2000). This prescription should be based on evidence and/or clinical guidelines (Firtell, 1968). However, much of the design principles for RPDs are based on anecdotal reports or clinical experience rather than high levels of scientific evidence (Davenport et al., 2001). This is also true for the selection of clasps for RPDs.

Clasps used in RPD designs are derived from wrought wire or are cast components of an all-metal framework. These clasps need to be flexible enough to allow the RPD to be seated and removed by the patient or clinician numerous times without permanently deforming the clasp and without damaging the tooth (Bates, 1980; Frank & Nicholls, 1981; Matheson, 1986; Brockhurst, 1996; Waldmeier, 1996; Owen, 2000). The flexibility of a clasp is influenced by its length, diameter, cross sectional form and by its material e.g. wrought wire clasps are more flexible than cast clasps (Morris et al., 1981a, b; Morris et al., 1983; Stade et. al, 1985; Davenport et al., 2001)

Although there have been attempts in the literature to provide guidelines for RPD designs and clasp selection, the studies generally fall short of simulating the clinical situation and extrapolation from these studies must be done with caution.

## 1.2 LITERATURE REVIEW

Brumfield (1953; cited in Bates, 1965) stated that "history repeatedly records the development of science to explain the rule of thumb methods previously worked out on an artistic or practical basis and this applies to most dental appliances which have been developed by trial and error methods." The field of RPDs is fraught with concepts that are based on anecdotal reports and expert opinion but little has been scientifically validated.

Teeth have curved surfaces and these produce varying degrees of undercut for RPD clasps to engage. These undercuts allow denture clasps to resist dislodging forces experienced by RPDs during function. Hence, clasps are the components of the RPD that provide direct or active retention. Clasps used in RPD designs are derived from wrought wire or are cast components of an all metal RPD framework. The retentive/flexible component of a clasp arm exerts a force on the engaged tooth and this force should neither exceed the properties of the clasp material nor exceed the ability of a tooth to withstand them (Clayton and Jaslow, 1971). Therefore, these clasps need to be flexible enough to deflect out of a retentive undercut several thousand times without permanently deforming, yet be stiff enough to provide retention and be strong enough to prevent accidental damage (Brockhurst, 1996; Waldmeier, 1996).

The flexibility of clasps have been reported in the literature to be affected by its length, diameter, cross sectional form, curvature and the type of clasp alloy used (Clayton and Jaslow, 1971; Brudvik and Wormley, 1973; Morris et al., 1981; Johnson et al., 1983; Morris et al., 1983; Stade et. al., 1985; Waldmeier et al., 1996; Davenport et al., 2001).

Sharry (1977) stated that an RPD should not engage undercuts beyond the limits of their elasticity. However, this statement was based on a consensus reached by members of the Academy of Denture Prosthetics and were not substantiated with scientific data. From the 48 statements on RPDs published by the Academy, ten were related to denture clasps. These were, however, vague and did not offer guidelines to the selection of the appropriate clasp material, diameter or length.

If the amount of force required to overcome an undercut is beyond the elastic/proportional limit of the clasp arm, either the tooth will be affected or the clasp arm will be permanently deformed or will fracture (Bates, 1980; Matheson et al., 1986; Davenport, 2001). According to Vallittu and Kokkonen (1995) there are only a few studies that have examined the effect of clasp fatigue on the retention properties of denture clasps. Saito et al. (2002) followed patients treated with RPDs for at least two years and found that deformation and fractures of the clasp arms were the most frequently reported complication. Mahmoud et al. (2007) also reported that the most common mechanical complications affecting RPDs were permanent deformation and fatigue fracture of the clasp arms.

In general dental practice, often the selection of the clasp material, diameter and undercut it engages is left to the discretion of the dental technician. The design of the RPD is a professional responsibility and should be determined prior to taking the master cast (Sharry, 1977; Brudvik and Morris, 1981; Owen, 2000).

VandenBrink (1993) stated that limited attention is paid to the properties of clasp materials during the prescription stage of RPDs. The majority of the studies that investigated clasp material and design have been laboratory based and have failed to adequately simulate the oral environment in which these clasps must function. Many studies used straight wire samples and some that did use curved wire samples, used curvatures that are not clinically relevant. Some researchers attempted to approach the design of RPDs from a theoretical point of view. Warr (1961) proposed guidelines for the selection of clasps based on a mathematical formula; however, too many assumptions were made for certain elements of the formula that were not substantiated or validated. Therefore, it was suggested that it is more relevant to tests designs by practical means which can then be validated clinically (Bates, 1965).

Taylor and Peyton (1951; cited in Bates, 1965) stated that although clasps arms are more likely to experience bending during their function in the oral environment, laboratory bend tests for metals produce a higher value for the proportional limit than tensile tests. That could create an incorrect profile of the material properties in turn leading to an incorrect application for the use of those materials. Therefore, metals should be tested in tension and according to Bates (1965) tension tests produce a more accurate measure of strain and stress.

VandenBrink et al. (1993) compared various RPD clasp materials and fabrication procedures. They used six precious metals, six base metal alloys and two thermoplastic materials. Their findings were that precious metals displayed a lower stiffness and greater elastic range than the base metals. They also stated that cast base metal alloys were high stiffness materials and were unsuitable for short retentive arms for even 0.25mm undercuts. However, they only evaluated 9mm straight wire samples of diameters of 1mm (or the closest to that) in a unidirectional bending test. Therefore, their data must be interpreted with caution. They did acknowledge the shortfalls of their study stating that future research needs to incorporate wires of varying diameters and radii of curvature and that newer materials also need to be added to the test samples to allow for adequate comparison.

Johnson et al (1983) found that the curvature of a clasp around an abutment increased its resistance to deflection; however, the effect of this curvature on the retentive value of the clasp was not determined. Their study attempted to evaluate the effect of single plane curvature on half-round cast clasps and they showed that as the radius of curvature increased, the flexibility of the clasp decreased. However, the curvatures tested were 90°, 180° and 270° which do not correlate to the curvature of teeth.

Recent research by Goolam (1992) and Naidoo (2009) attempted to standardise the curvature of clasps tested. Goolam (1992) gathered load-deflection data that compared different alloys from which clinical guidelines could be established. Naidoo (2009) used the same methodology and gathered load-deflection data of the most current alloys. However, a limitation of these studies was that they used the average curvature length of premolars and molars instead of the actual curvature. The curvatures of teeth are non-uniform, an observation that could in all likelihood impact significantly on the behaviour of cast and wrought wire clasps.

Frank et al. (1983) investigated the flexibility of cast and wrought wire clasps of different gauges adapted against a single premolar tooth thereby simulating a clinical environment. They tested Ticonium® (CMP Industries LLC, New York, USA) and Vitallium® (Dentsply International, USA) cast clasps, and Ney® PGP (Dentsply International, USA), and Jelenko® (Jelenko Dental Alloys, USA) standard wrought wire clasps. The diameters they tested for each were 18, 19 and 20 gauge (diameters of 1.0mm, 0.9mm and 0.8mm respectively). Their findings were that cast clasps are less flexible than wrought wire clasps of the same gauge. They noted that an increase in flexibility did not equate to an increase in proportional limit. They also concluded that the proportional limit of the clasps they tested was high enough to be used in a 0.5mm undercut without causing permanent deformation of the clasp; however, to avoid excessive force on the abutment teeth they advised only using these clasps in a 0.25mm undercut.

Clayton and Jaslow (1971) and Marei (1995) were also concerned with the effect the clasp would have on the abutment tooth. They evaluated methods aimed at reducing the transmission of excessive force to the abutment tooth and stated that this could be done by using more flexible clasps. Arda and Arikan (2005) also advocated using a clasp arm design that produces less stress on abutment teeth which they claim will produce more predictable long term use of RPDs.

Brudvik and Wormley (1973) found a marked decreased in flexibility when they decreased the length of the wires from 12mm to 8mm. However, they only tested straight wire samples.

Their conclusions were that when the length of the clasp is shorter that a smaller diameter wire should be used to construct the clasp and that the decision of the gauge of the clasp arm must be a clinical one.

Davenport et al. (2001) reported that the length of a clasp is crucial when determining the depth of the undercut it can engage. They stated that cast Chrome-Cobalt (Cr-Co) clasps needed to be at least 15mm to flex 0.25mm without permanently deforming, and that wrought wire clasps needed to be at least 7mm to overcome 0.5mm undercuts without deforming. Their conclusions were, however, formulated from questionnaires that were sent to a selection of clinicians. So their statements were a consensus of expert opinions and were not validated or substantiated. There is a variety of wrought wire clasp alloys available and in varying diameters and their statements do not adequately account for this.

Waldmeier et al. (1996) found that wrought wire clasps of the same dimension but different alloy composition affected flexibility. However, this was evaluated by performing a three-point bending test on straight wire samples.

The clinical observation of a loss of retention of the RPD after the prosthesis is worn for some time raises the question of whether constant deflection of the clasp during insertion and removal causes fatigue of the clasp or whether the clasp is functioning too close to its proportional limit. Arda and Arikan (2005) stated that metal and metal alloys undergo permanent deformation and fatigue when exposed to repeated stress. They found that cast Cr-Co clasps lost their retentive force significantly over a 36-month period of simulated use. Matheson et al. (1986) also found that repeated deformation and re-adaptation of clasp arms resulted in a clinically significant loss of retention. Vallittu and Kokkonen (1995) stated that the fatigue of a denture was based on the repeated deflection of the clasp during insertion and removal.

Warr (1961) recognised that stresses experienced by clasps should not exceed a critical value. This was to allow the clasp to function without danger of deformation. He believed that the reason why clasps deform or fracture is that they function too close to their proportional limit. Warr defined a concept of the margin of safety which he believed was a realistic expectation of clasp function. He argued that at the proportional limit there is no margin of safety. Warr formulated a mathematical formula to calculate the margin of safety, but too many assumptions had to be made and his theory was never tested in the laboratory or clinically.

Bates (1965) stated that because the clasp is repeatedly bent, it may fail at a low level of stress due to fatigue. He suggested that using a clasp where the force required to deflect its tip to overcome the undercut is less than the proportional limit is sound practice. He also looked at determining how much the force should be below the proportional limit to allow an adequate safety margin. Bates approached this issue from a statistical point of view and stated that a clasp should be selected to function where the force required to overcome the undercut is equal to the proportional limit of that material less two standard deviations. He termed this concept the ‘realistic limit’.

After reviewing the literature on the behaviour of RPD clasps it was evident that all the preceding studies conducted made assumptions as to length and curvature of teeth which are key factors that affect the flexibility of a clasp. It is also evident from the literature that clasp length or curvature has not been clinically evaluated or standardised. It was therefore decided to conduct a study on clasps that reflected clinically encountered tooth curvatures and lengths and produce clinically relevant guidelines on the selection of clasp material for RPD construction.

# CHAPTER 2. AIMS AND OBJECTIVES

## 2.1 AIM

The aim of this study was to establish evidence-based, clinically relevant guidelines for the selection of anatomically correct clasps (cast and wrought wire) to be placed into the appropriate undercuts of abutment teeth in order to provide optimal retention for RPDs.

## 2.2 OBJECTIVES

1. To determine an average length and curvature of clasps in the correct position on premolar and molar teeth.
2. To print a three-dimensional model representing these average curvatures and lengths and to use this model to adapt clasps appropriately.
3. To record the proportional limit of each of these clasps and derive the load exerted in clinically used undercuts.
4. To use these data to establish clinical guidelines for the placement of clasps which will provide optimal force and be within their realistic limit (being two standard deviations from their proportional limit).

# CHAPTER 3. MATERIALS AND METHODS

## 3.1 MATERIALS USED

The various wrought wire alloys and diameters were chosen based on availability in South Africa. The selection of these materials was done purposely to allow the guidelines developed from this study to be relevant to our setting and allow follow up clinical research to be done. The wires which were chosen were grouped according to alloy type, diameter and tooth to which they would be adapted (Table 1). Gold wire was not used as it is no longer readily available and is too expensive to be used routinely.

**Table 1.** Clasp materials tested in the study

| Material | | Diameter (mm) | Number of specimens | |
|---|---|---|---|---|
| | | | Premolar | Molar |
| No. | Wrought wire | | | |
| 1 | Leowire[a] | 0.9 | 10 | 10 |
| 2 | | 1.0 | 10 | 10 |
| 3 | Remanium Hard [b] | 0.9 | 10 | 10 |
| 4 | | 1.0 | 10 | 10 |
| 5 | *Noninium[c] | 0.9 | 10 | 10 |
| | Cast | | | |
| 6 | Vitallium®[d] | | 10 | 10 |

[a] Leowire [chromium stainless steel alloy], Leone, Fiorentino, Italy
[b]Remanium Hard [Chromium nickel stainless steel alloy], Dentaurum, Pforzheim, Germany
[c]Noninium [Nickel free stainless steel], Dentaurum, Pforzheim, Germany
*Only 0.9mm available from the supplier in South Africa
[d] Vitallium [Composition: Cobalt 63.4%, Chromium 29.0%, Molybdenum 5.2%], Dentsply, Ontario, Canada

## 3.2 METHOD

### 3.2.1 Sample size calculation

*Number of teeth to be selected and scanned*

After consultation with a biostatistician, it was determined that thirty premolars and molars would initially be selected and scanned. The relative margin of error (size of the confidence interval in relation to the standard deviation of the data) was then to be calculated. If this was unacceptably high, a further ten teeth for each group would be selected and scanned and the relative margin of error again calculated until the confidence interval in relation to the standard deviation of the data was acceptable. The margin of error for the thirty premolars and molars was found to be acceptable.

*Number of clasps per tested group*

The sample size for the different clasp materials and diameter was ten per group. This was chosen in line with previous research, where the effect size had been shown to yield acceptable relative precision (Brudvik and Wormley, 1973; Matheson et al., 1986; Goolam, 1992; Arda and Arikan, 2005; Naidoo, 2009).

### 3.2.2 Determining the average curvatures

Discarded casts were randomly and anonymously collected from the Wits Oral Health Centre dental laboratory. The casts selected had undamaged first premolars and molars with no evidence of cavitation or fracture. The casts were placed onto a dental surveyor and the maximum curvature of each tooth was established in the normal manner. A line was then drawn to represent the position of a normal clasp arm (figure 1).

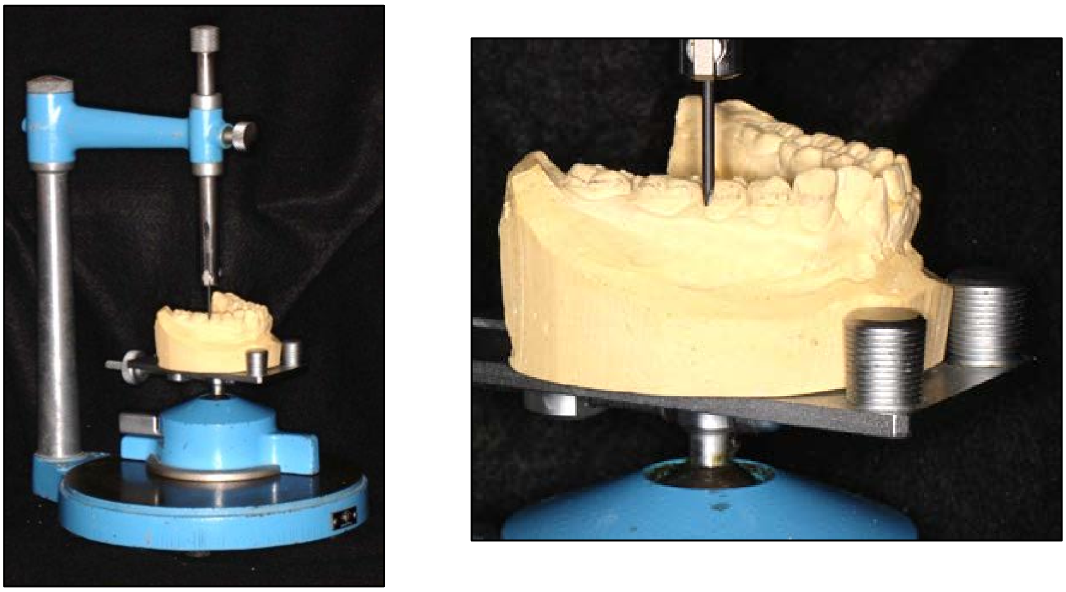


**Figure 1.** (a) Cast placed on a dental surveyor and (b) the maximum curvature being identified with a graphite marker

The teeth were then sectioned into separate dies and each die was trimmed to the line representing the clasp arm. To further assist with the orientation during scanning the lingual/palatal half of the tooth crown was also reduced to create a hemi-sectioned die in the buccal-palatal/lingual direction. The midpoint between the mesial and distal edges were determined and marked on the die. This determined point lined up with the midpoint of the scanning template (figure 2).

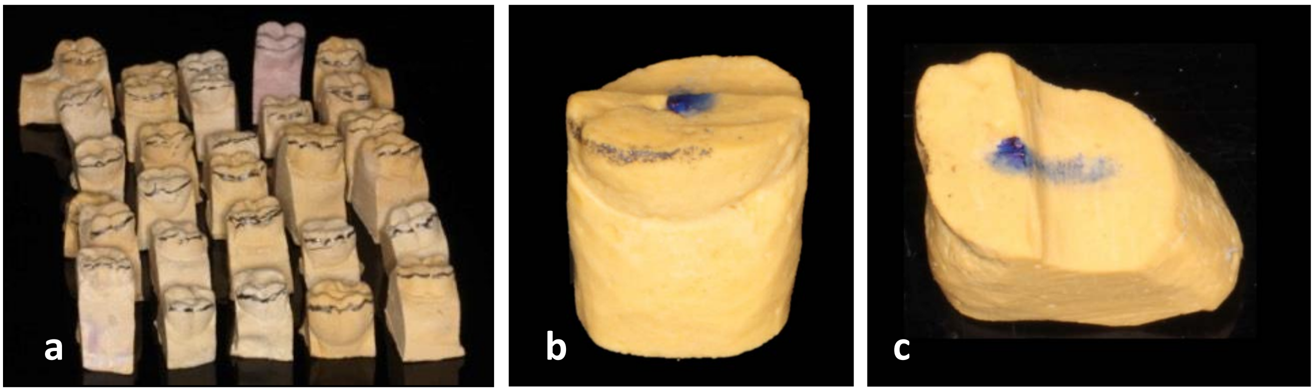



**Figure 2.** (a) Sectioned dies; (b) trimming to the line representing the clasp arm and (c) marking the midpoint on the die

A scanning template (30 mm x 30 mm) was used when scanning each tooth. This was to establish a reference of scale, which allowed for an audit during the study to ensure that magnification errors did not occur. A midpoint was also established on the scanning edge of the template to which the midpoint of each die aligned. This facilitated reproducibility in the next step.

A flatbed scanner (Canon MG3540, Canon, USA) was used and each image was scanned using a 600 dpi resolution setting (figure 3).


**Figure 3.** The dies being scanned on a flatbed scanner with the scanning template.

Once each tooth had been scanned the premolar images and molar images were separated into their respective folders according to their groups. The images (figure 4) per group were imported individually into the software package CorelDraw® (Corel, Ottawa, Canada) onto separate layers.

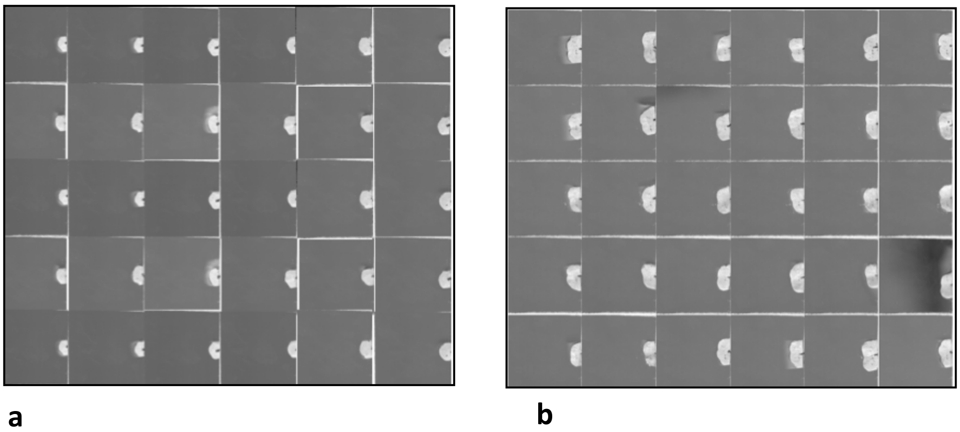



**Figure 4.** (a) Premolar scan images (b) Molar scan images

The edges of the template square were aligned to ensure accuracy. Once all the images were imported, the buccal curvature of each tooth was traced in the software. The traced curvature corresponded to the clinical position of a clasp. An average curvature pattern emerged and this was mapped in software. This curvature represented the determined average curvature for the teeth in that group (figure.5).

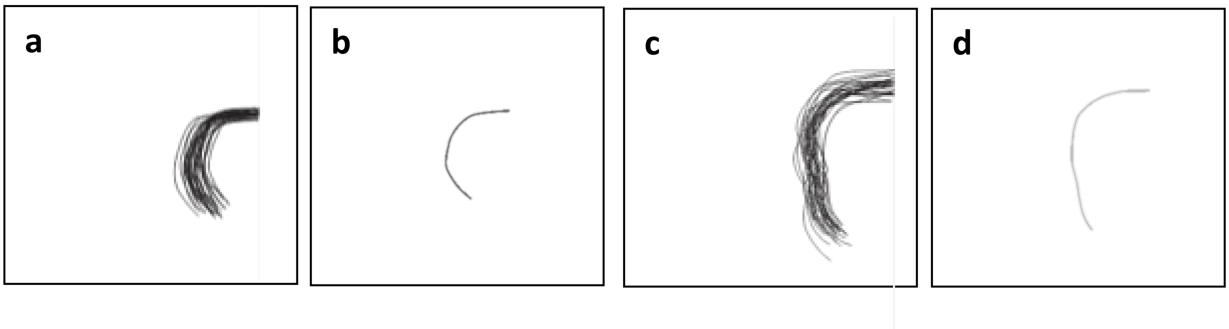



**Figure 5.** The curvature tracings (a) 30 premolars, (b) average premolar curvature tracing, (c) 30 molars, (d) average molar curvature tracing

### 3.2.3 Construction of the average curvature model

The average curvature images for the premolars and molars were imported into the software Solidworks® (Dassault Systèmes Solidworks Corporation, Massachusetts, USA) and 3D models of each curvature were created (figure 6 ).

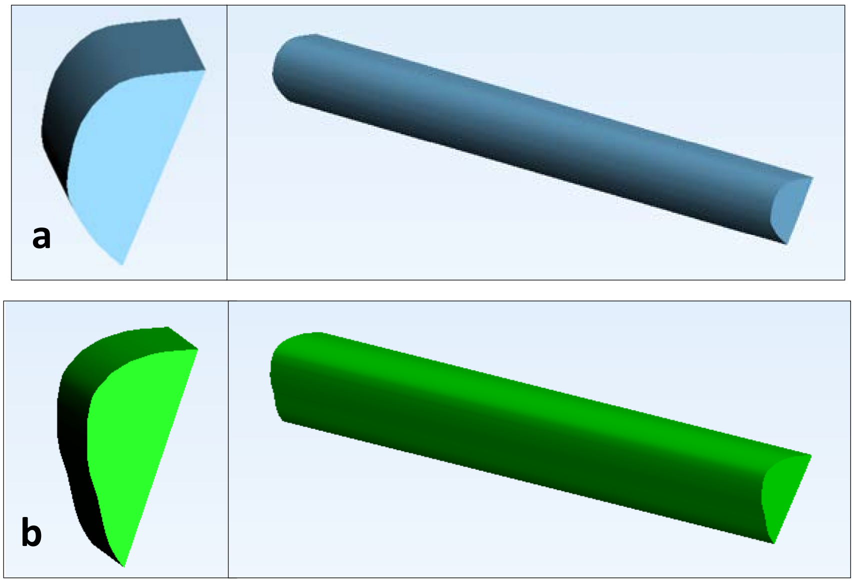



**Figure 6.** 3D models of the (a) average premolar curvature and (b) average molar curvature created in Solidworks©

Additional primitive objects were added to the 3D models as per figure 7a. This was done to facilitate the construction of reproducible clasp arm samples. The combined shape was extruded to measure 100mm in length (figure 7a). This shape would allow adequate space for ten wire samples to be adapted to the curve simultaneously. The 5mm ledge was created on the curve corresponding to the clasp tip. This was done to ensure that each sample terminated at the same point.  Standard Tessellated Language (stl) files of these models were created and exported to a 3-D printer (Objet 350v ®, Objet Inc., Rehovot, Israel) and printed using Objet FullCure720 RGD720 (figure 7b).

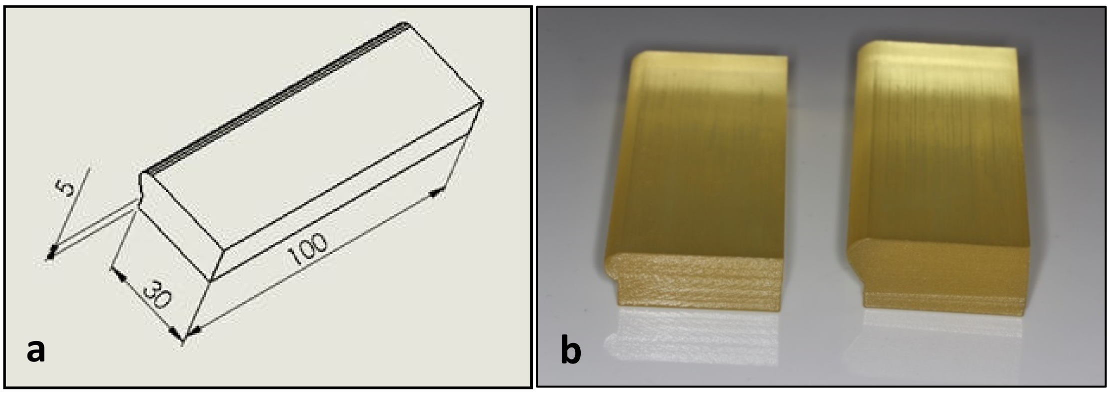



**Figure 7.** (a) Schematic of model exported to 3-D printer (b) printed models

### 3.2.4 Constructing the clasp samples

The various wrought wire and cast clasps samples were adapted to these models by a single dental laboratory technician. This ensured that any inherent error that may be incorporated from his technique would be constant. All the cast clasps were invested in the same flask to eliminate discrepancies that could occur during that process. Each clasp sample was analysed under magnification by a single operator to identify any defects (figure.8).

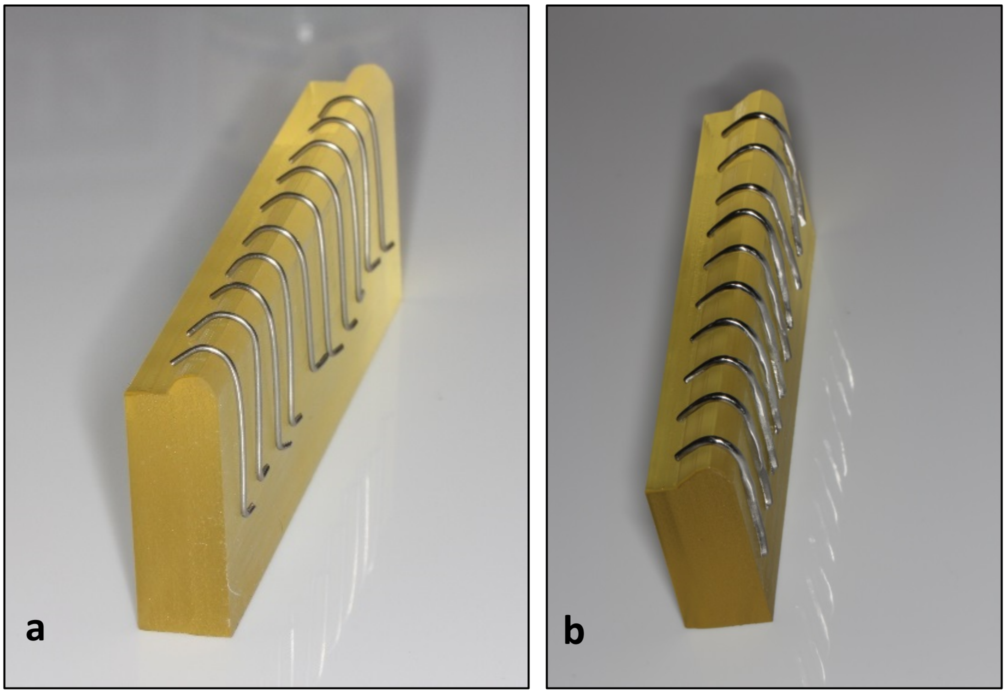



**Figure 8.** Examples of (a) wrought wire clasps and (b) cast clasps adapted to the 3D models

The clasps were then incorporated into acrylic resin blocks measuring 38mm x 25mm x 6mm using a prefabricated mould (figure 9). Ten samples were prepared at the same time using this mould. All the samples were bevelled on the side of the clasp tip to allow for placement of the samples in the tensile testing machine without obstruction. All the samples were checked again by a single operator to ensure that there was no mobility of the wires in the acrylic resin blocks.

**Figure 9.** Prefabricated mould used to embed the clasps into acrylic resin blocks

### 3.2.5 Tensile testing

The clasps embedded in acrylic blocks were tested in a tensile testing machine (Instron Corporation, High Wycombe, United Kingdom). The clasp tips were engaged in a custom-made jig and displaced at a cross-head speed of 0.5 mm / minute using a 2kN load cell (figure10).

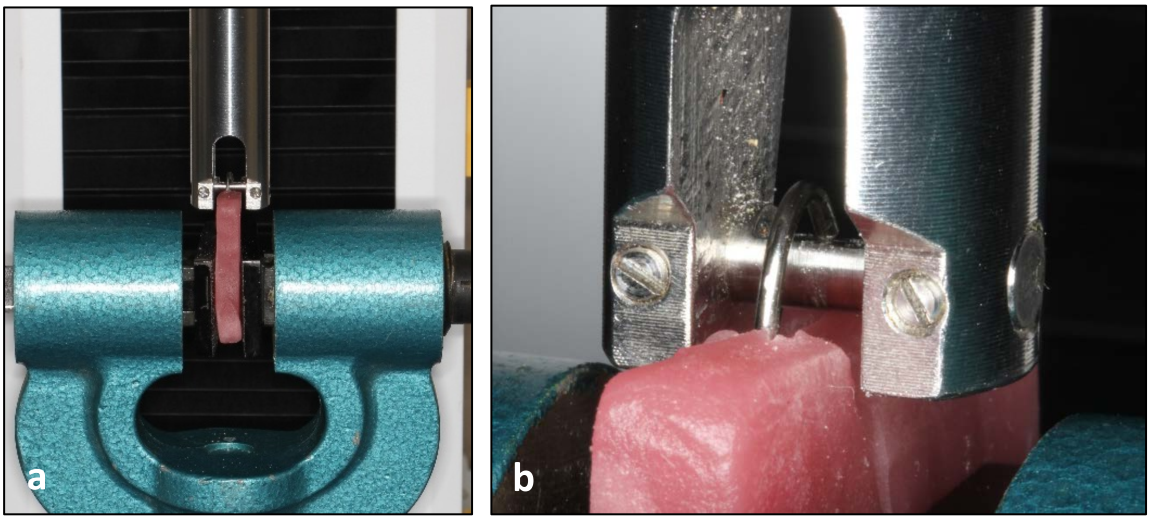



**Figure 10.** (a) Position of the acrylic resin block in the clamp. (b) Clasp tip engaging the jig

The jig was connected directly to the 2KN load cell so that any contact with it would reflect a load. The acrylic resin samples were held in place by a clamp attached to the bottom of the tensile testing machine and each sample was placed and tightened in position by a single operator before each test. The load cell was calibrated before each test to eliminate any measurement caused by the clasp tip touching the jig. Only one test for each sample was made as the clasp arm was displaced until its proportional limit was exceeded as illustrated in figure 11.

**Figure 11.** Permanent deformation of the premolar cast clasp on the right after it was displaced until its proportional limit was exceeded.

All the testing was performed by a single operator to standardise the recorded data. All 120 samples were tested on the same day.

The Bluehill Lite software program (Instron, USA) was used in conjunction with the tensile testing machine. The software was programmed to record the proportional limit for each specimen as well as the load experienced at the following deflections:  0.25mm, 0.5mm and 0.75mm. These deflections represent the common undercuts used clinically. The software was also programmed to account for the slippage that occurs between the clasp tip and the jig platform during the initial movement of the load cell. This ensured that any deviation from this would represent the true proportional limit of the experiment sample (figure 12).

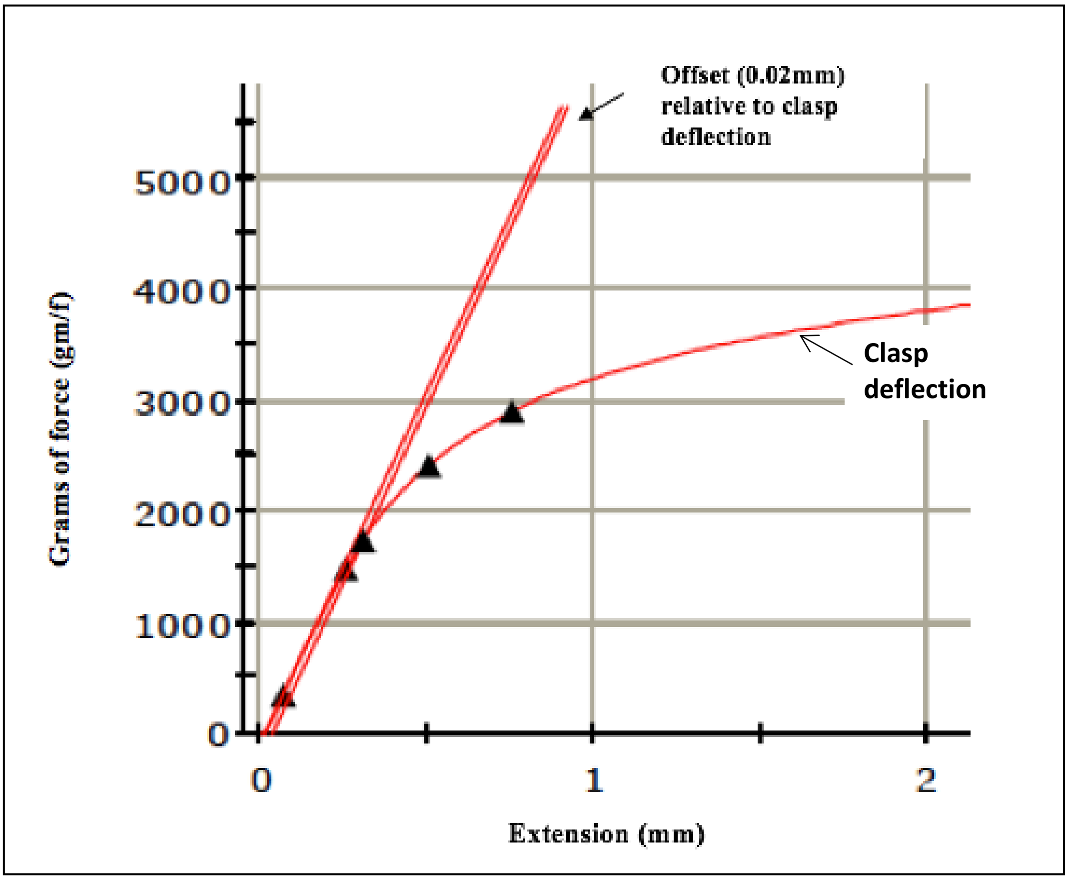



**Figure 12.** Graph generated during the tensile testing for a premolar cast clasp

### 3.2.6 Data Analysis

Data analysis was carried out using SAS (SAS Institute Inc., SAS Software, version 9.3 for Windows, Cary, NC, USA: SAS Institute Inc. (2002-2010)).

For each experiment (combination of tooth type and clasp type), the univariate statistics (mean, standard deviation) for the force at 0.25, 0.50 and 0.75 mm deflection, as well as at the proportional limit, were tabulated in a spreadsheet. The realistic limit (Bates, 1965) was calculated for each experiment from the proportional limit as: mean less 2 standard deviations.

# CHAPTER 4. RESULTS

The lengths of the determined average curvature for the premolar and molars dies selected were 9mm and 14.5mm respectively.

None of the wrought wire samples fractured during testing however one molar cast clasp (figure.13a) and five of the premolar cast clasps (figure.13b) fractured.

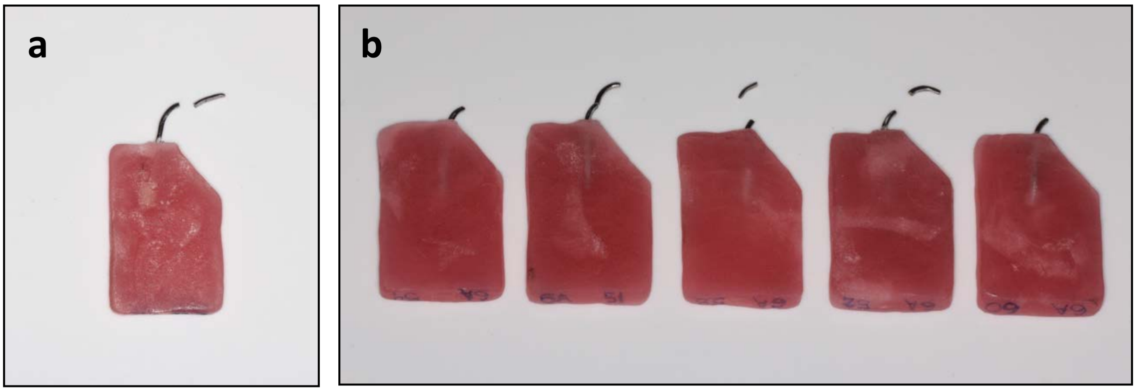



**Figure 13.** Bent and fractured premolar cast clasps (a) and a fractured molar clasp (b)

Table 2 shows the univariate statistics for each combination of tooth type and clasp type.

**Table 2.** Tabulated results from the load tests for each experiment and the calculated realistic limit

| Tooth | Clasp material | Mean proportional limit (g/f) | Mean force at deflection (g/f) | | | Standard deviation (sd) | Realistic Limit |
|---|---|---|---|---|---|---|---|
| | | | 0.25mm | 0.50mm | 0.75mm | | |
| Premolar | Leowire 0.9mm | 1002.7 | 302.9 | 583.3 | 849.1 | 215.3 | 572.0 |
| | Leowire 1.0mm | 1130.0 | 676.3 | 1212.6 | 1755.0 | 233.2 | 663.6 |
| | Remanium Hard 0.9mm | 896.3 | 420.0 | 799.6 | 1162.1 | 211.5 | 473.3 |
| | Remanium Hard 1.0mm | 857.5 | 535.2 | 1027.2 | 1474.5 | 139.8 | 577.9 |
| | Noninium 0.9mm | 573.7 | 360.1 | 618.7 | 817.5 | 53.6 | 466.5 |
| | Vitallium | 1457.4 | 1178.9 | 1927.4 | 2364.1 | 332.8 | 791.8 |
| Molar | Leowire 0.9mm | 1219.0 | 105.6 | 203.2 | 288.0 | 387.9 | 443.2 |
| | Leowire 1.0mm | 855.6 | 358.8 | 656.5 | 916.0 | 81.4 | 692.8 |
| | Remanium Hard 0.9mm | 698.7 | 124.0 | 235.2 | 334.4 | 74.2 | 550.3 |
| | Remanium Hard 1.0mm | 915.3 | 218.7 | 417.3 | 604.0 | 142.0 | 631.3 |
| | Noninium 0.9mm | 704.9 | 160.1 | 274.3 | 362.5 | 123.3 | 458.3 |
| | Vitallium | 1310.0 | 773.3 | 1463.8 | 1987.5 | 271.0 | 768.1 |

Histograms and the relative standard deviation (RSD) were inspected to identify and correct/exclude any outliers and none were found (Appendix 1). The 5% significance level was used throughout. The p-values were calculated for the mean force exerted at the different deflections to determine if they were statistically significantly different from the realistic limit for that wire (Table 3).

**Table 3.** Tests for significance between the deflection at different undercuts and the realistic limit of each wire. Only those shown are statistically significant.

| Clasp Type | Tooth Type | p-values | | |
|---|---|---|---|---|
| | | Deflection 0.25 mm | Deflection 0.50 mm | Deflection 0.75 mm |
| 0.9mm Leowire | premolar | <0.0001 | | |
| | molar | <0.0001 | <0.0001 | <0.0001 |
| 1.0 mm Leowire | premolar | | | |
| | molar | <0.0001 | | |
| 0.9mm Remanium Hard | premolar | 0.016 | | |
| | molar | <0.0001 | <0.0001 | <0.0001 |
| 1.0 mm Remanium Hard | premolar | | | |
| | molar | <0.0001 | <0.0001 | |
| 0.9 mm Noninium | premolar | 0.0006 | | |
| | molar | <0.0001 | <0.0001 | 0.0003 |
| Vitallium | premolar | | | |
| | molar | | | |

The mean force at each of the three deflections was compared to the realistic limit to determine which wire types and diameters exceeded their realistic limit (table 4). Charts where generated to illustrate the mean load compared to the proportional limit and realistic limit for each material at the clinically relevant undercuts for the premolar samples (figures 14-19) and molar samples (figures 20-25).

**Table 4.** Mean loads and realistic limits (in parentheses) for premolar and molar clasps. Mean loads highlighted in red represent values which have exceeded the realistic limit for that deflection

| Clasp Material | Diameter | Deflection | | | | | |
|---|---|---|---|---|---|---|---|
| | | Premolar | | | Molar | | |
| | | 0.25mm | 0.5mm | 0.75mm | 0.25mm | 0.5mm | 0.75mm |
| Leowire | 0.9mm | 302.9 (572) | 583.3 (572) | 849.1 (572) | 105.6 (443.2) | 203.2 (443.2) | 288 (443.2) |
| | 1.0mm | 676.3 (663.6) | 1212.6 (663.6) | 1755 (663.6) | 358.8 (692.8) | 656.5 (692.8) | 916 (692.8) |
| Remanium Hard | 0.9mm | 420 (473.3) | 799.6 (473.3) | 1162.1 (473.3) | 124 (550.3) | 235.2 (550.3) | 334.4 (550.3) |
| | 1.0mm | 535.2 (577.9) | 1027.2 (577.9) | 1474.5 (577.9) | 218.7 (631.3) | 417.3 (631.3) | 604 (631.3) |
| Noninium | 0.9mm | 360.1 (466.5) | 618.7 (466.5) | 817.5 (466.5) | 160.1 (458.3) | 274.3 (458.3) | 362.5 (458.3) |
| Vitallium | - | 1178.9 (791.8) | 1927.4 (791.8) | 2364.1 (791.8) | 773.3 (768.1) | 1463.8 (768.1) | 1987.5 (768.1) |

***Wire samples adapted to the average curvature of premolars***

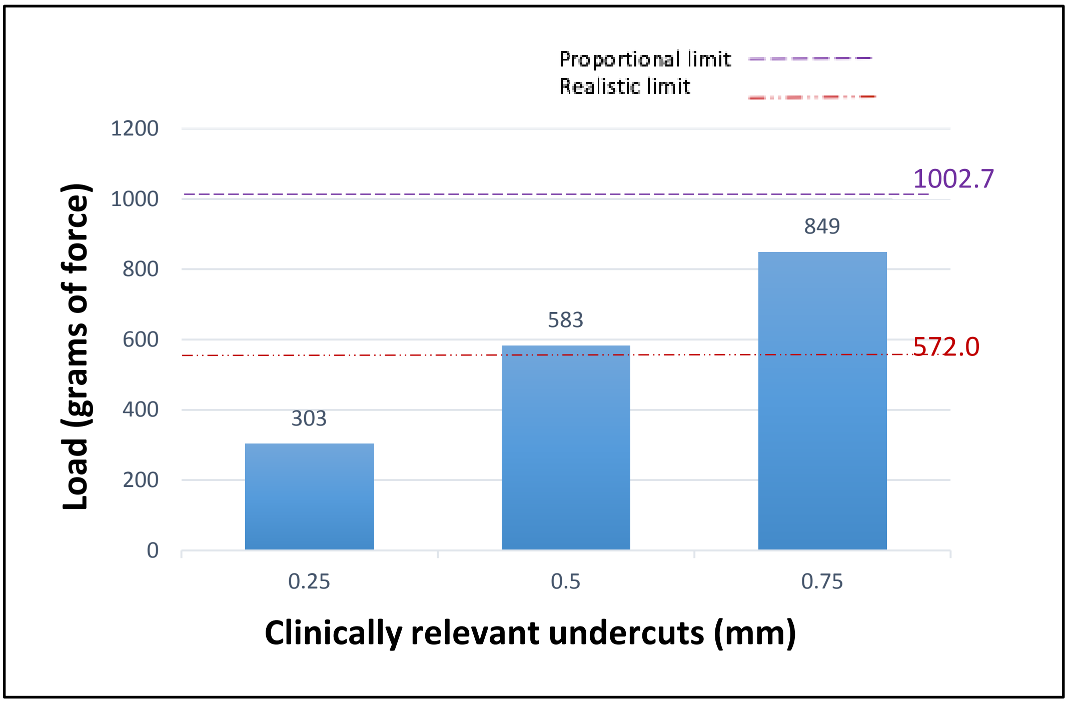



**Figure 14**. Mean load exerted by 0.9mm Leowire wrought wire samples adapted to the average curvature of premolars

**Figure 15.** Mean load exerted by 1.0mm Leowire wrought wire samples adapted to the average curvature of premolars

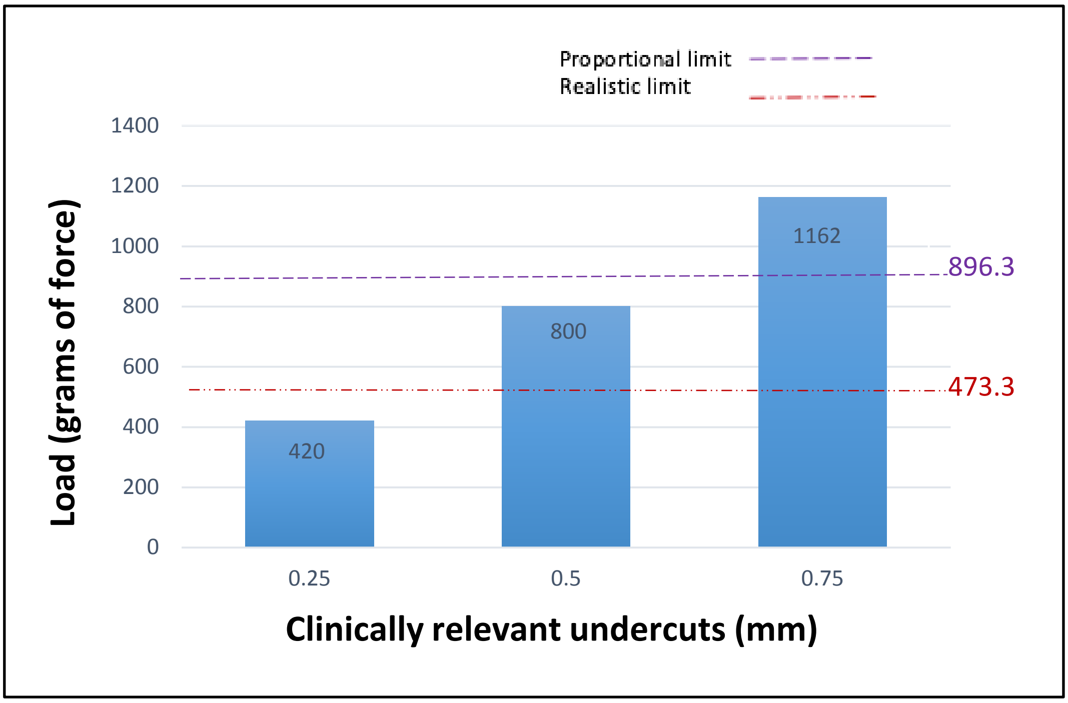



**Figure 16.** Mean load exerted by 0.9mm Remanium Hard wrought wire samples adapted to the average curvature of premolars

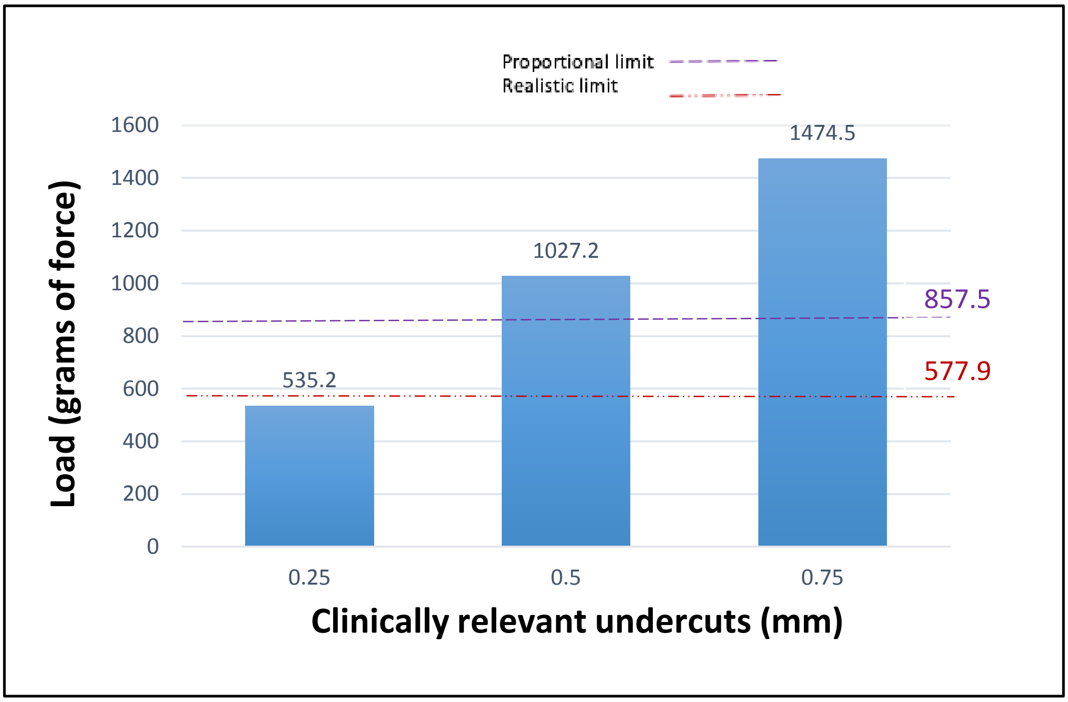



**Figure 17.** Mean load exerted by 1.0mm Remanium Hard wrought wire samples adapted to the average curvature of premolars

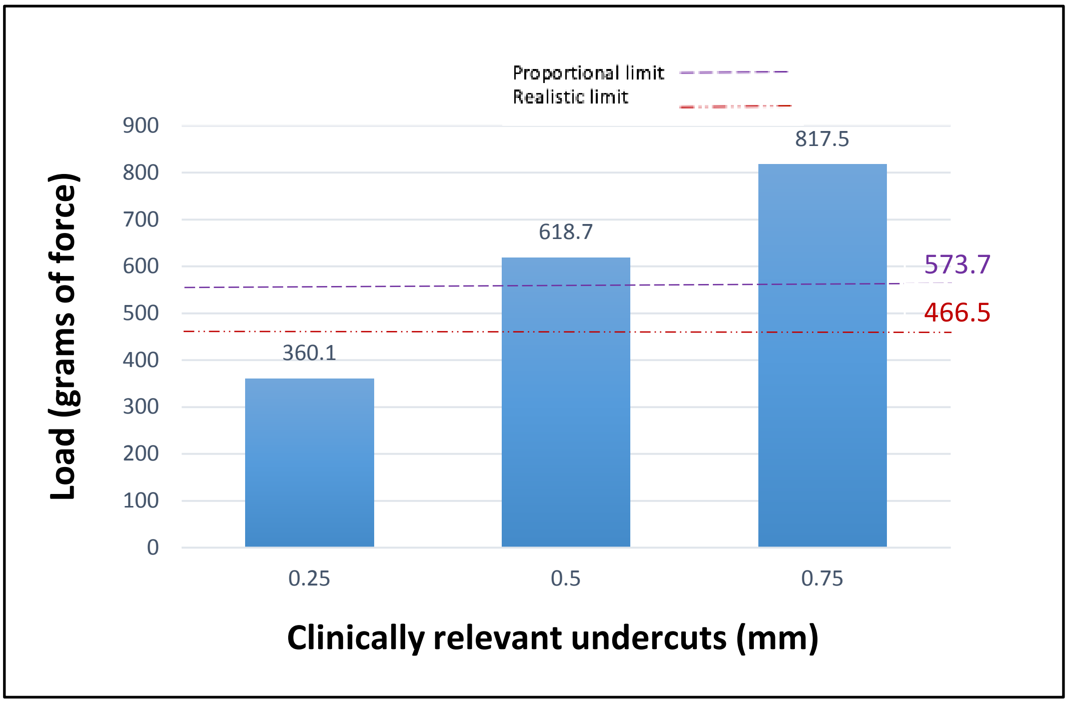



**Figure 18**. Mean load exerted by 0.9mm Noninium wrought wire samples adapted to the average curvature of premolars

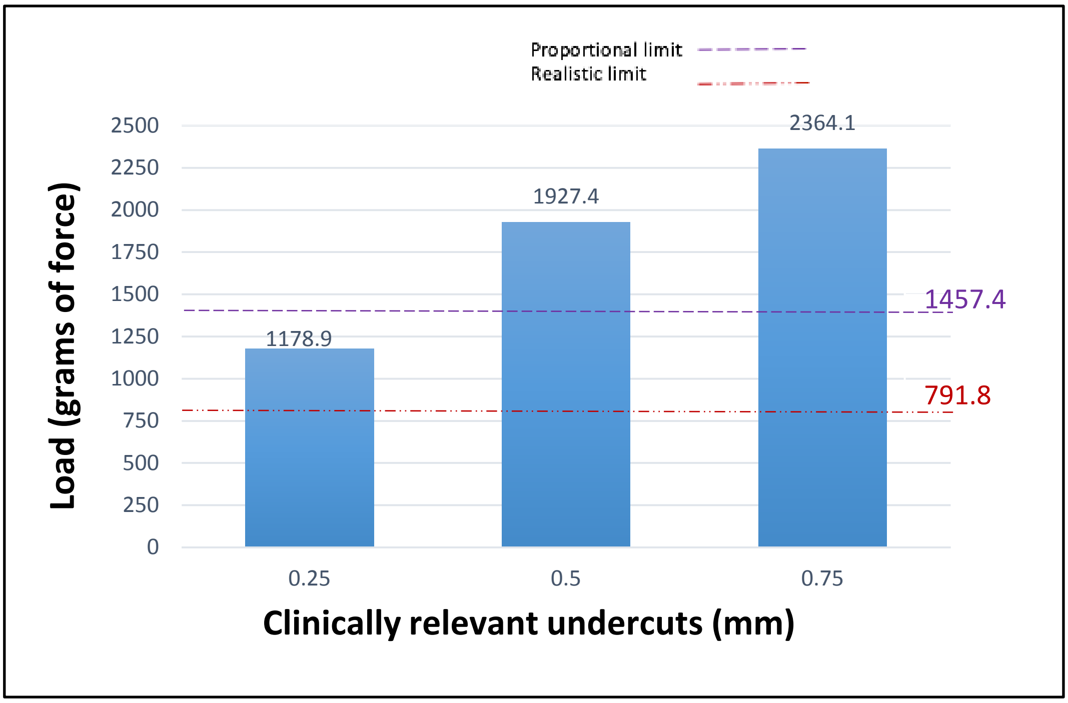



**Figure 19**. Mean load exerted by Vitallium cast samples adapted to the average curvature of premolars

**Wire samples adapted to the average curvature of molars**

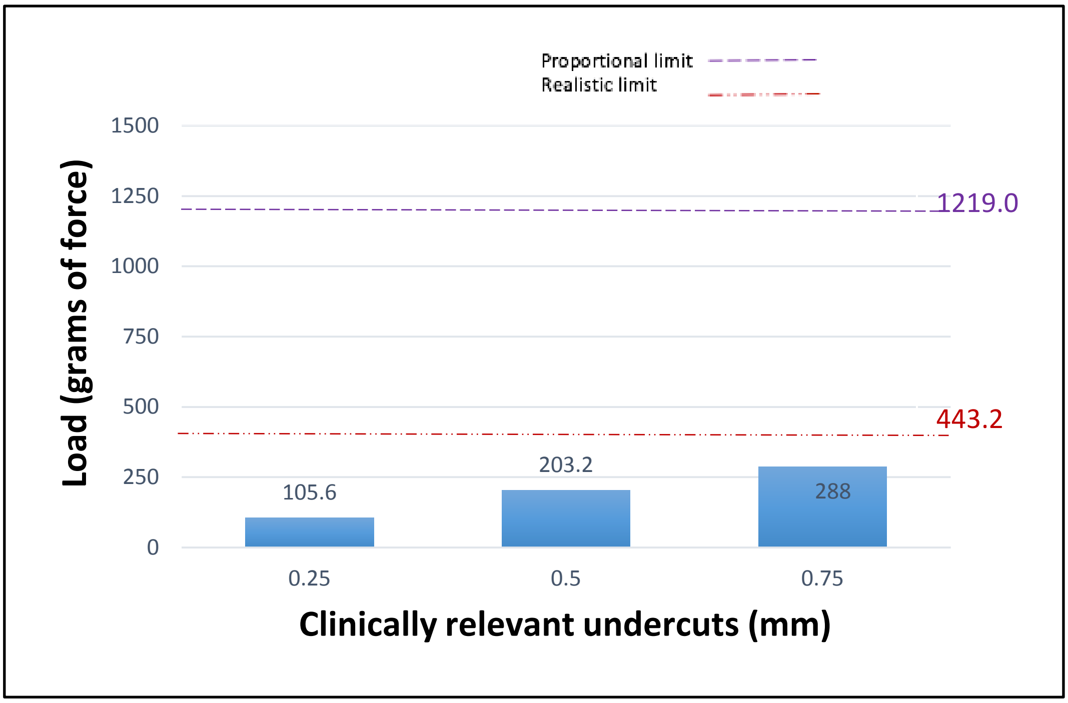



**Figure 20.** Mean load exerted by 0.9mm Leowire wrought wire samples adapted to the average curvature of molars

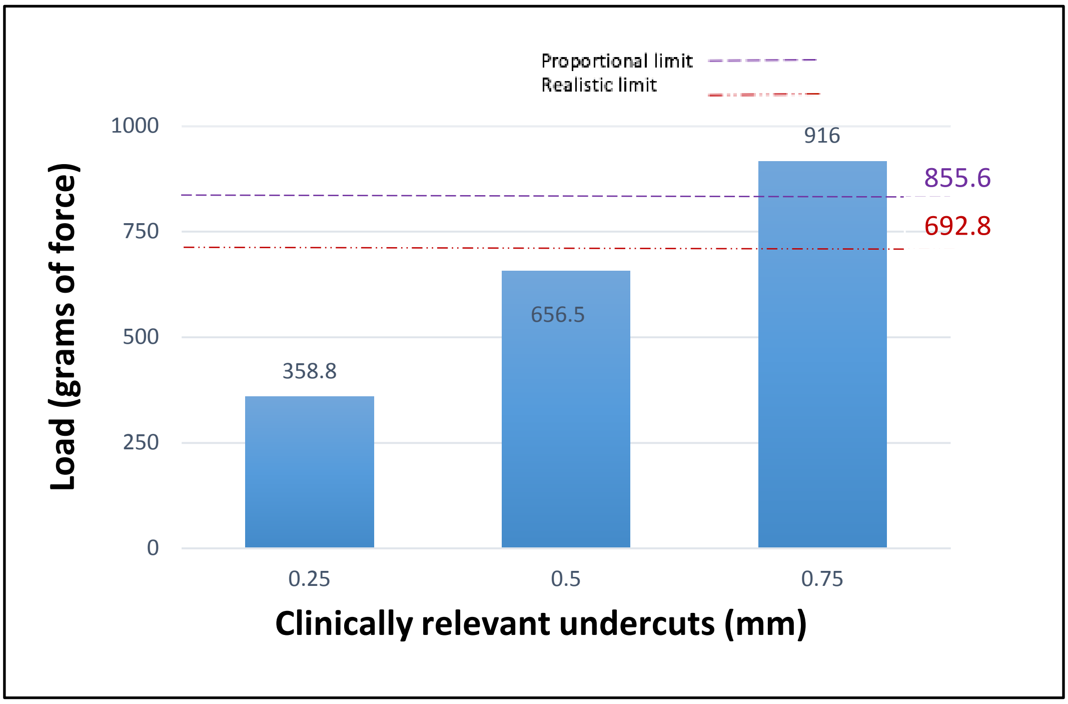



**Figure 21.** Mean load exerted by 1.0mm Leowire wrought wire samples adapted to the average curvature of molars

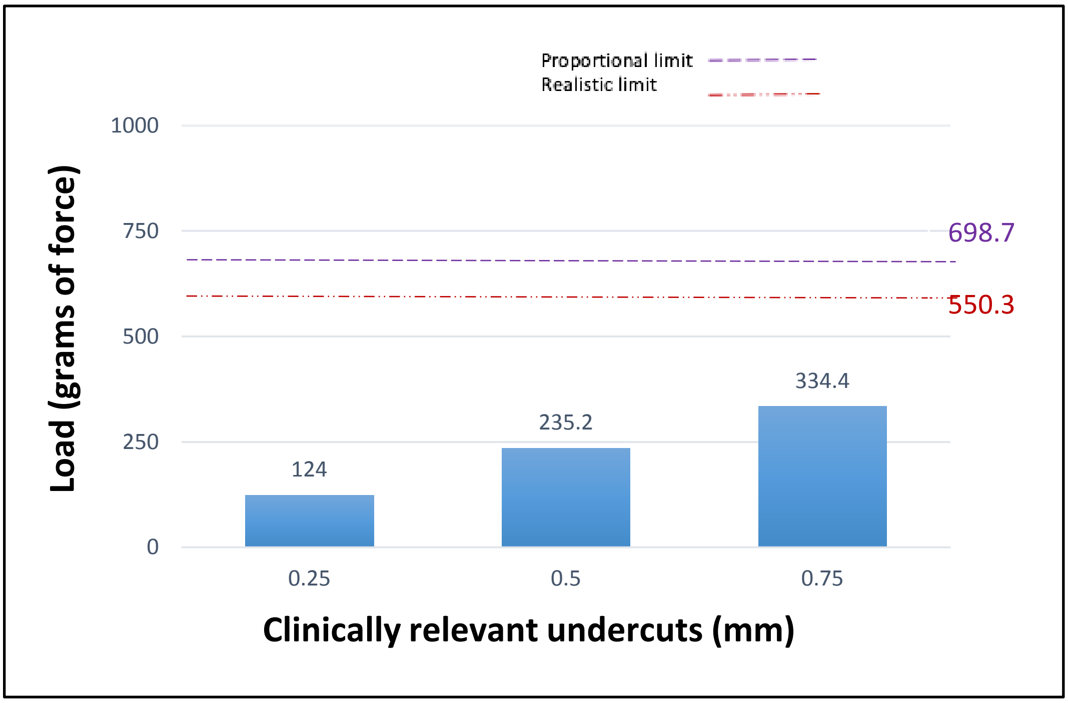



**Figure 22.** Mean load exerted by 0.9mm Remanium Hard wrought wire samples adapted to the average curvature of molars

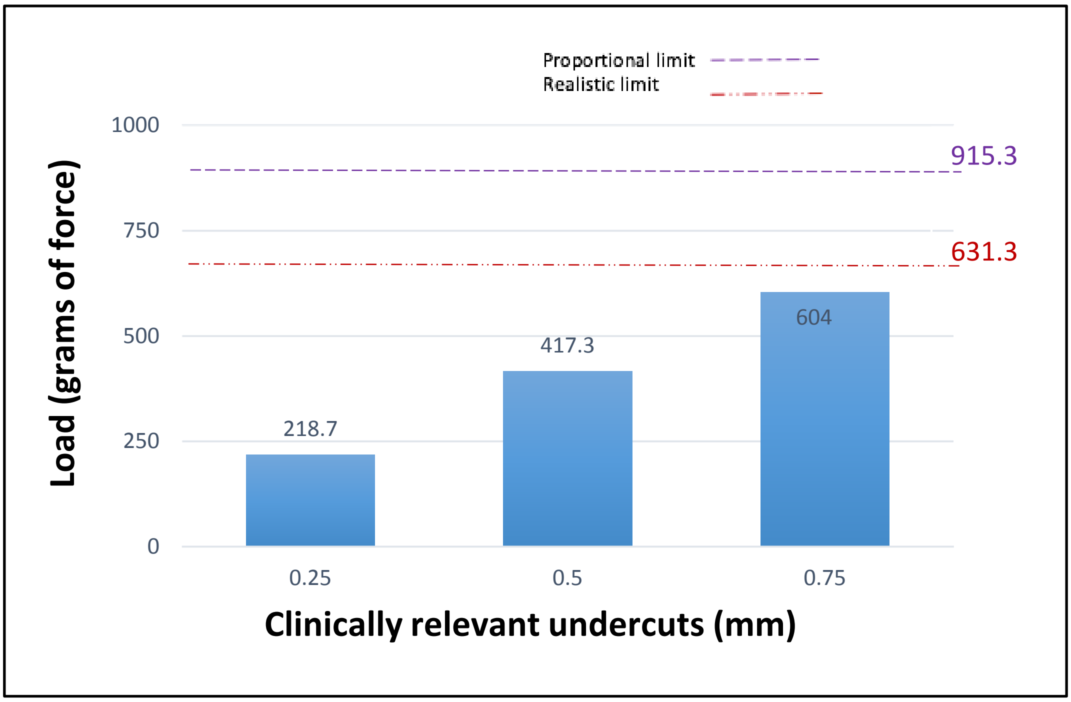



**Figure 23.** Mean load exerted by 1.0mm Remanium Hard wrought wire samples adapted to the average curvature of molars

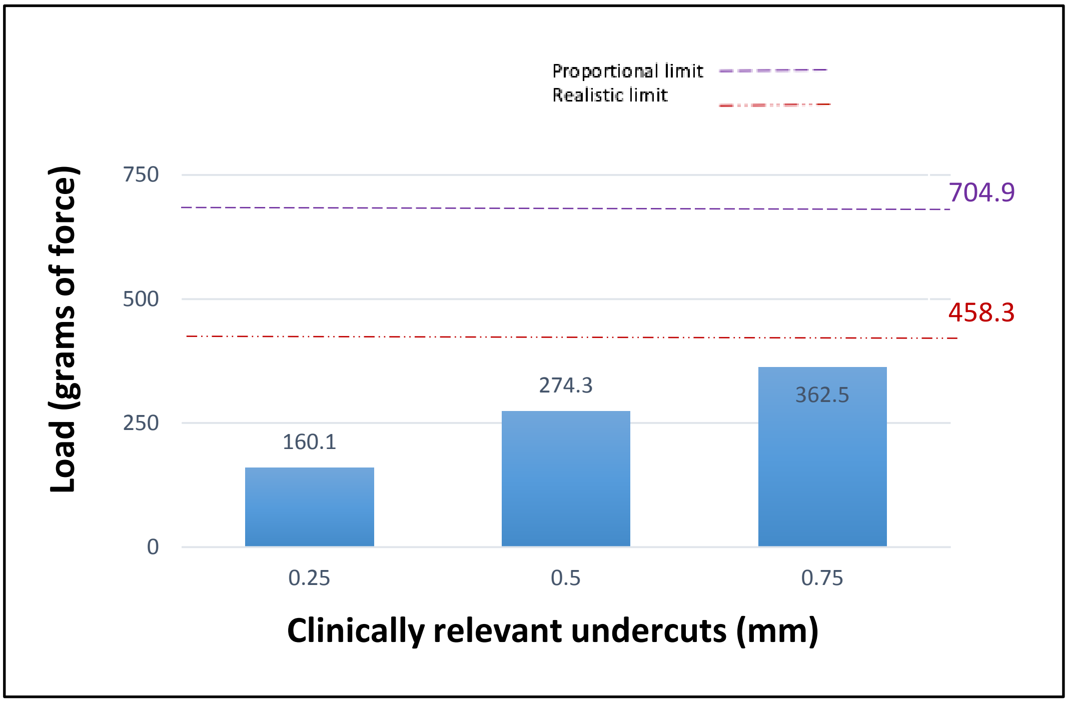



**Figure 24**. Mean load exerted by 0.9mm Noninium wrought wire samples adapted to the average curvature of molars

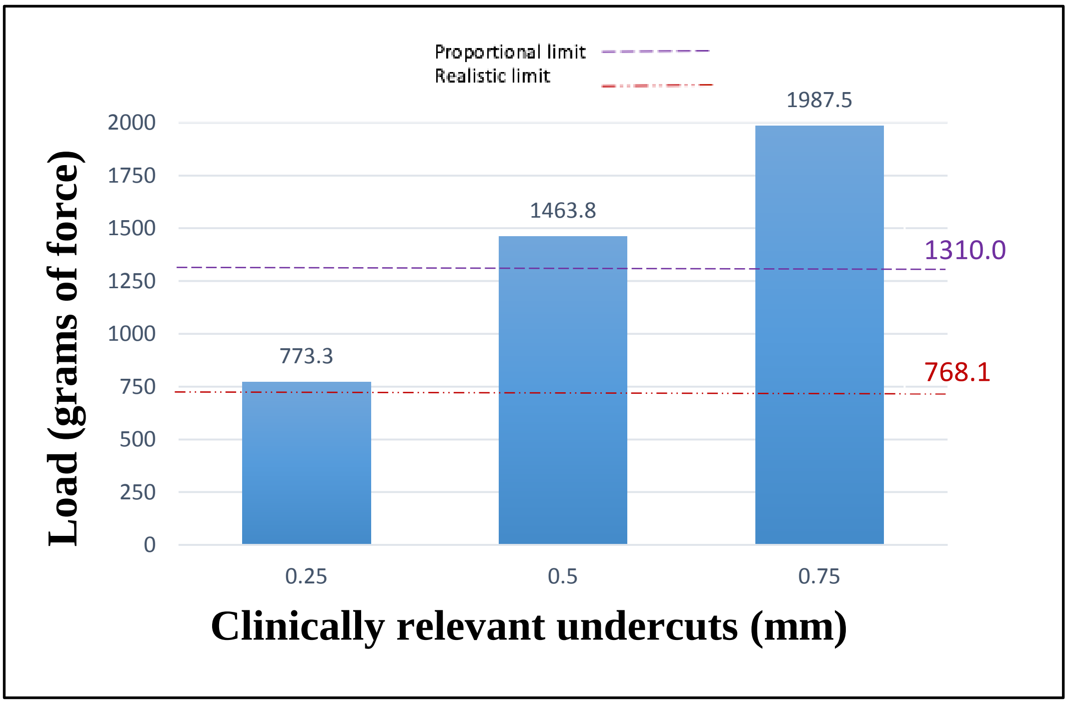



**Figure 25.** Mean load exerted by Vitallium cast samples adapted to the average curvature of molars

Analyses of the results from tables 3 and 4 as well as the figures 14-25 were carried out. The findings were as follows:

- The mean loads exerted on premolar teeth were within their realistic limit at the 0.25mm deflection for the 0.9mm Leowire, 0.9mm Remanium Hard, 0.9mm Noninium and 1.0mm Remanium Hardwire samples. However, the mean load exerted by Leowire 1.0 mm diameter wire for premolar clasps at 0.25 mm undercut was only 2% above the value for the realistic limit.
- The mean force exerted by cast clasps for the premolars were not within the realistic limit for any of the deflections, although they were within their proportional limit at the 0.25mm deflection (table 5).
- Five of the ten samples had fractured during the testing (figure 13).

**Table 5.** Mean load exerted by Vitallium cast clasps adapted to the average curvature of molars and premolars. Values in parenthesis are the determined realistic limits

| Clasp Material | Deflection | | | | | | | |
|---|---|---|---|---|---|---|---|---|
| | Premolar | | | | Molar | | | |
| | 0.25mm | 0.5mm | 0.75mm | Proportional Limit | 0.25mm | 0.5mm | 0.75mm | Proportional Limit |
| Vitallium | 1178.9 (791.8) | 1927.4 (791.8) | 2364.1 (791.8) | 1457.4 | 773.3 (768.1) | 1463.8 (768.1) | 1987.5 (768.1) | 1310 |

- The mean forces exerted on molar teeth were within the realistic limit at 0.25mm, 0.50mm and 0.75mm deflections for the 0.9mm Leowire, 0.9mm Remanium Hard and 0.9mm Noninium wire samples and for for the 1.0mm Leowire samples at 0.25mm and 0.50mm deflections.
- The mean forces exerted on molar teeth by the 1.0mm Remanium Hard wire samples were below their determined realistic limit at 0.25mm, 0.50mm and 0.75mm deflections. However, the p-value for the 0.75mm deflection was 0.27. The raw data showed that 3 of the 10 samples were above the realistic limit, which would account for the lack of statistical significance.

- The mean forces exerted on molar teeth by the cast clasps were not within the realistic limit for any of the deflections (table 5). However, the mean force exerted at the 0.25mm deflection was below the proportional limit.
- The mean forces exerted by the wrought wire samples that were within their realistic limit are tabulated in Table 6 in rank from highest to lowest.

**Table 6.** Mean forces exerted (in parentheses) by wrought wire samples that were within their realistic limit ranked from highest to lowest

| Tooth | Deflection | Diameter | Force exerted from highest to lowest (f/g) | | |
|---|---|---|---|---|---|
| **Premolar** | 0.25mm | 0.9mm | Remanium Hard (420) | Noninium (360.1) | Leowire (302.9) |
| **Molar** | 0.25mm | 0.9mm | Noninium (160.1) | Remanium Hard (124) | Leowire (105.6) |
| | 0.50mm | | Noninium (274.3) | Remanium Hard (235.2) | Leowire (203) |
| | 0.75mm | | Noninium (362.5) | Remanium Hard (334.4) | Leowire (288) |
| **Molar** | 0.25mm | 1.0mm | Leowire (358.8) | - | Remanium Hard (218.7) |
| | 0.50mm | | Leowire (656.5) | - | Remanium Hard (417.3) |

# CHAPTER 5. DISCUSSION

Davenport (2000) stated that clasps tend to lose their efficiency over time. Arda and Arikan (2005) showed that over a 36-month period RPD clasps underwent permanent deformation and fatigue when exposed to repeated stress. This raises the question of whether constant deflection of the clasp during insertion and removal causes fatigue of the clasp (Vallittu and Kokkonen, 1995) or whether the clasp functioned too close to its proportional limit (Bates, 1965). Deformation of the clasp arm may lead to unfavourable stresses to the abutments and the RPD itself (Keltjens et al., 1997).

The variations noted in the load exerted at the different deflections and in the proportional limits for the different materials and diameters indicate that there are inconsistencies in the manufacturing or casting of these materials. Therefore, a margin of safety needs to be applied when developing guidelines for the selection of these materials to be used as RPD clasps. For this reason, Bates (1965) concept of the ‘realistic limit’ was applied in this study as a measure of what this margin of safety should be.

Previous studies have made assumptions as to the length and curvature of teeth. Davenport et al. (2001) stated that length of a clasp is crucial when determining the depth of undercut. They stated that wire clasps need to be at least 7mm in length to overcome 0.5mm undercuts without deforming. They also stated that a cast clasp needs to be at least 15mm. According to them, anything shorter than 15mm would be too rigid to disengage 0.25mm undercuts without permanently deforming or damaging the tooth. However, these statements were unsubstantiated by scientific evidence and were based on the clinical opinion of colleagues.

The models created to represent the average curvature of molars and premolars in this study attempted to simulate the clinical length and curvature to which clasps for premolars and molars would be formed. The average lengths of the clasps adapted to these models were determined by measuring both the model and the clasps adapted to the model. The lengths were 9mm for premolars and 14.5mm for molars. This is in contrast to the lengths used by Goolam (1992) and Naidoo (2009) in which 12mm and 20mm for premolars and molars respectively. They had, however, based these figures on the average diameters of premolars and molars which they measured from mesial contact to distal contact and not on the actual length of the clasp arm.

The length of the clasp in this study proved to be a significant factor in the flexibility of the clasps. All the 0.9mm molar wire samples (14.5mm) were able to function within their realistic limit for all three deflections (0.25mm, 0.50mm and 0.75mm) as opposed to the premolar wire samples (9mm) that were only able to function within their realistic limit at 0.25mm deflections. Therefore, the flexibility decreased as the length decreased. Not surprisingly, the retentive force of the clasps for the given undercut were also affected by the length of the clasp as the stiffer the clasp the greater the force required for its flexure.

Diameter affects the flexibility of clasps and this was also confirmed. The increased diameter resulted in an increase in the load exerted for the given undercut which in turn resulted in some clasps exerting forces beyond their realistic limit at the larger undercuts (0.50mm and 0.75mm).

Understanding the nature of forces that are involved in dislodging a denture is essential in determining the appropriate clasp and denture design for a given situation. Caldwell (1962) measured the force required to separate various food types lodged between two flattened enamel surfaces. Bates (1963) then adapted that method and calculated the force required to dislodge sticky toffee held between two porcelain plates. He used this measured force to calculate the force required to retain a unilateral distal extension RPD. His calculation was based on the assumption that the maximum displacing force on an RPD occurred when the teeth had penetrated the food followed by the jaw opening after 0.5 seconds. He suggested that the amount of resisting force required to retain a unilateral distal extension RPD could possibly be as high as 2600 grams. He acknowledged that this value was determined by simplifying the complex masticatory process and that for other foods the force may not be that large. He concluded that the force required to retain a unilateral distal extension RPD which replaced 3 teeth, ranged from 850g to 1500g for ‘normal foods’ (corn flakes, mashed potatoes, etc.). In contrast, Frank and Nicholls (1981) reported that the amount of force required to retain a mandibular bilateral distal extension RPD was between 300 to 750 grams of force. They came to this conclusion by designing an *in vitro* model with a bilateral distal extension RPD and subjected it to a test which removed it along a single centred path of removal.

These study designs do not replicate the clinical situation where the direction, magnitude and point of application of the forces on RPDs vary vastly. It must also be recognised that RPDs are subject to forces other than those along their path of insertion or removal. There are other factors that aid in RPD retention such as indirect retention, guide plane retention, the muscular control of the patient against the polished surface of the denture and other inherent physical forces which arise from covering the mucosa by the denture (Davenport, 2000).

Guiding surfaces and guide planes produce passive retention which assist the clasps in retaining the denture (Bates, 1980). This effect is greater in a bounded saddle than in a distal extension base. Therefore, the force required to retain a distal extension RPD is expected to be higher than a force required to retain a bounded saddle RPD.

Vanzeveren et al. (2003) reported a high prevalence of abutment tooth loss among patients using distal extension base RPDs. They also reported that abutments in bilateral distal extension base RPDs had a shorter survival than those in unilateral distal extension bases. Whether this was related to the increased occlusal load that these teeth experience in these scenarios or the effect of the RPD clasp on the abutment tooth, is unclear as this was not evaluated.

Worthington et al. (2003) reported results from an *in vivo* study which showed that the average occlusal forces during normal function ranged from 200 Newton (20,387g) to 450 Newton (45,871g). These values indicate that teeth are able to withstand high forces as a result of the physiological behaviour of their periodontal ligament (PDL). However, when the PDL is compromised the ability of the abutment tooth to withstand these forces is compromised. Tada et al. (2013) eluded to this indirectly with their prognostic risk factors, and their finding that abutment teeth are compromised when the crown root ratio is altered as a result of clinical attachment loss, as well as when there are probing depths greater than 5mm. It could be deduced, therefore, that the status of the abutment tooth is more significant than the actual force exerted by the clasp.

There appears to be a lack of guidance in the literature regarding the amount of force an abutment tooth can withstand from RPD clasps and until clinically validated studies can be done, it would be difficult to determine if higher forces have unfavourable effects on the abutment teeth. Therefore, until those studies can be done, a force which affords the RPD the greatest opportunity to be retained during function should be advocated. Hence, it would seem wise to choose clasp arms which will proved the greatest retention at all times, provided that those clasp arms can function elastically, which in this study has been defined as within a realistic limit of two standard deviations from the proportional limit.

The results from this study suggest that when clasping premolars only wrought wire samples of 0.9mm diameter should be used and undercuts of no more than 0.25mm be engaged. Molars clasped with 0.9mm wrought wire clasps engaging an undercut of 0.25mm exerted less force than their premolar counterpart, confirming that an increase in length of the clasp arm increased its flexibility. Similarly, molar clasp arms needed to engage deeper undercuts to exert a force comparable to the force exerted by premolar length clasp arms. (table 6).

Increasing the diameter of the molar wrought wire clasp from 0.9mm to 1.0mm significantly increased the force exerted at 0.25mm, 0.50mm and 0,75mm undercuts. This confirmed the influence of diameter on flexibility

The premolar clasp of 1.0 mm Leowire was above its realistic limit but by only 2% and so if higher forces are required clinically, it would be acceptable to use this combination, with the proviso that its longevity may be slightly compromised.

The loads exerted on premolar teeth by the cast clasps were found to be well beyond the realistic limits for all three depths of undercut (table 5). Although the mean load exerted for the cast premolar clasps at 0.25mm undercut was below the proportional limit, it was much higher than the realistic limit at this deflection. Five of the ten samples had fractured during the testing (figure 13). However, this does not coincide with clinical experience and there may be several reasons for this, including that technicians may not be placing the clasp tips into the undercut of the tooth. Firtell et al., (1985) reported that many of the clasps sent to

commercial labs in their study placed clasps in less undercut than prescribed. This study has found that molar cast clasps exerted high mean loads at all 3 depths of undercuts which exceeded the realistic limit (table.5). Only one molar cast clasp fractured during the testing. Although the mean load exerted for the molar cast clasps at 0.25mm of undercut was well below the proportional limit of the material, it was nevertheless higher than the realistic limit at that undercut. Clinically, this is the smallest undercut that can be detected, and if it used, its longevity may be a problem. There are many situations where it is most appropriate to place a cast clasp on a molar tooth, so if this is done, the patient should be warned of the possibility that it may fracture during the lifetime of the denture (Keltjens et al. 1997).
Casting the cast clasps together in one casting is a limitation of this study as this did not allow for natural variations that occur during casting. Firtell et al. (1985) reviewed the literature and found that numerous publications had identified that there are various factors that can affect the accuracy of the cast RPD which in turn affects the force the cast clasp will exert on the abutment tooth. Rudd and Rudd (2001a, b, c) identified 243 possible errors that could occur during the casting of an RPD. Most of these errors are technical and may not be identified even after a complication has occurred.

A follow up study were each clasp is cast individually would allow for the natural variations that occur during the casting process and may be more clinically applicable.

# CHAPTER 6. CONCLUSION AND RECOMMENDATIONS

## 6.1 CONCLUSION

Permanent deformation and fatigue fracture are among the most common mechanical complications that can affect RPD clasps (Keltjens et al. 1997; Saito et al., 2002; Mahmoud et al., 2007). The resultant loss of retention and reduced stability can compromise the comfort of the patient. An RPD clasp design based on sound knowledge of the behavioural characteristics of the various clasps materials and diameters should decrease the incidence of these mechanical complications.

Within the limitations of this study it was shown that:

- The flexibility of wrought wire clasp samples was influenced by its alloy type, diameter, length and depth of undercut.
- All tested wrought wires clasps of 0.9mm in diameter can be used to engage 0.25mm undercuts of premolars.
- All tested wrought wire clasps of 0.9mm in diameter can be used on molars for all three clinical undercuts; however, the retentive force is low and therefore a diameter of 1.0mm is suggested and only to be used in 0.25mm undercuts for Leowire and in 0.25mm or 0.50mm undercuts for Remanium Hard.
- 0.9mm Noninium wrought wire clasps can be used for patients with Nickel allergies for premolars (0.25mm undercut) and molars (all three clinical undercuts).
- Cast clasps should not be used on premolars as the length of the average premolar is insufficient to allow the clasp to be inserted and removed without fracture or fatigue of the clasp.

- Cast clasp of 14.5mm in length may be used cautiously on molars in undercuts of 0.25mm only. However, a longer clasp arm would probably increase its flexibility and reduce the risk of permanent deformation.

The following clinical guidelines for the selection of RPD clasps were made on the basis of this study for the materials used (table 7).

**Table 7.** Clinical guidelines for the clasp selection for molars and premolars based the expected realistic limit of function for each.

| | Premolars | Molars | | | |
|---|---|---|---|---|---|
| | 0.25mm | 0.25mm | | 0.5mm | 0.75mm |
| | Stainless Steel/Cr-Co | Cast | Stainless Steel | Stainless Steel/Cr-Co | Stainless Steel / Cr-Co |
| **Wire name and diameter (force in grams)** | Leowire 0.9mm (303)<br>Remanium Hard 0.9mm (420)<br>Leowire 1.0mm (606) | Vitallium (773) | Leowire 1.0mm (359) | Remanium Hard 1.0mm (417) | |
| **For nickel sensitive patients** | Noninium 0.9mm (360) | | Noninium 0.9mm (160) | Noninium 0.9mm (274) | Noninium 0.9mm (363) |

## 6.2 RECOMMENDATIONS

Further clinically relevant studies are required to investigate the ideal forces required to retain both distal extension RPDs and bounded saddle tooth supported RPDs. Investigations into other factors such as guide plane retention, indirect retention, etc. need to be conducted to assess their effect on the overall retention of an RPD. A follow up study should be conducted where the cast clasps are cast individually. This will assess whether the natural variation that occurs during the casting process significantly varies the results achieved in this study.

The guidelines produced from this study need to be validated by assessment in a clinical setting over an adequate period of follow up.

# REFERENCES


Arda T, Arikan A. An in vitro comparison of retentive force and deformation of acetal resin and cobalt-chromium clasps. J Prosthet Dent. 2005; 94: 267- 274

Bates JF. Retention of cobalt-chromium partial dentures. Dent Practise Dent Rec. 1963; 14: 168-171

Bates JF. The mechanical properties of the cobalt-chromium alloy and their relation to partial denture design. Br Dent J. 1965; 119: 389-395

Bates JF. Retention of partial dentures. Br Dent J. 1980; 149: 171-174.

Brockhurst PJ. A new design for partial denture circumferential clasp arms. Aus Dent J. 1996; 41:317-323.

Brudvik JS, Morris HF. Stress-relaxation testing. Part III: Influence of wire alloys, gauges and lengths on clasp behaviour. J Prosthet Dent. 1981; 46: 374- 379

Brudvik JS, Wormley JH. Construction techniques for wrought wire retentive clasp arms as related to clasp flexibility. J Prosthet Dent. 1973; 30: 769-774

Caldwell RC. Adhesion of foods to teeth. J Dent Res. 1982; 41: 821-832

Chatzivasileiou K, Kotsiomiti E, Emmanouil I. Implant-assisted removable partial dentures as an alternative treatment for partial edentulism: a review of the literature. Gen Dent. 2015; 63:21-25.

Clayton JA, Jaslow C. A measurement of clasp forces on teeth. J Prosthet Dent. 1971; 25: 21-43

Davenport JC, Basker RM, Heath JR, Ralph JP, Glantz P-O, Hammond P. Clasp design. Br Dent J. 2001; 190:71-81

Davenport JC, Basker RM, Heath JR, Ralph JP, Glantz P-O. Retention. Br Dent J. 2000; 189 (12), 646-657.

Firtell DN. Effect of clasp design upon retention of removable partial dentures. J Prosthet Dent. 1968; 20: 43-52

Firtell DN, Muncheryan AM, Green AJ. Laboratory accuracy in casting removable partial denture frameworks. J Prosthet Dent. 1985; 54: 856-862

Frank RP, Brudvik JS, Nicholls JI. A comparison of the flexibility of wrought wire and cast circumferential clasps. J Prosthet Dent. 1983; 49: 471-476.

Frank RP, Nicholls JI. A study of the flexibility of wrought wire clasps. J Prosthet Dent. 1981; 45:259-267.

Goolam R. The selection of wrought wire clasps for removable partial dentures. 1992. Unpublished research report.

Johnson DL, Stratton RJ, Duncanson MG. The effect of single plane curvature on half-round cast clasps. J Dent Res. 1983; 62:833-836.

Keltjens HMAM, Mulder J, Käyser AF, Creugers NHJ. Fit of direct retainers in removable partial dentures after 8 years of use. J Oral Rehab. 1997; 24: 138- 142

Koller B, Att W, Strub J. Survival rates of teeth, implants and double crown-retained removable dental prostheses: a systematic review. Int J Prosthodont. 2011; 24: 109-117.

Mahmoud AAA, Wakabayashi N, Takahashi H. Prediction of permanent deformation in cast clasps for denture prosthesis using a validated nonlinear finite element model. Dent Mater. 2007; 23: 317-324

Marei MK. Measurement (in vitro) of the amount of force required to dislodge specific clasps from different depths of undercut. J Prosthet Dent. 1995; 75: 258-263

Matheson GR, Brudvik JS, Nicholls JI. Behaviour of wrought wire clasps after repeated permanent deformation. J Prosthet Dent. 1986; 55: 226-231

Morris HF, Asgar K, Brudvik JS, Winkler S, Roberts EP. Stress-relaxation testing. Part IV: Clasp pattern dimensions and their influence on clasp behaviour. J Prosthet Dent. 1983; 50: 319-326.

Morris HF, Asgar K, Roberts EP, Brudvik JS. Stress-relaxation testing. Part II: Comparison of bending profiles, microstructures, microhardnesses and surface characteristics of several wrought wires. J Prosthet Dent. 1981a; 46: 256- 262

Morris HF, Asgar K, Tillitson E. Stress-relaxation testing. Part I: A new approach to the testing of removable partial denture alloys, wrought wires and clasp behaviour. J Prosthet Dent. 1981b; 46: 133-141.

Naidoo LM. Guidelines for the use of wrought wire clasps for removable partial dentures. 2009. Unpublished research report.

Owen CP. Fundamentals of removable partial dentures (2nd Ed.). 2000. Ndabeni: UCT Press.

Rudd RW, Rudd KD. A review of 243 errors possible during the fabrication of a removable partial denture: Part I. J Prosthet Dent. 2001a; 86: 251-261

Rudd RW, Rudd KD. A review of 243 errors possible during the fabrication of a removable partial denture: Part II. J Prosthet Dent. 2001b; 86: 262-276

Rudd RW, Rudd KD. A review of 243 errors possible during the fabrication of a removable partial denture: Part III. J Prosthet Dent. 2001c; 86: 277-288

Saito M, Notani K, Miura Y, Kawasaki T. Complications and failures in removable partial dentures: a clinical evaluation. J Oral Rehab. 2002; 29: 627- 633

Sharry J. Principles, concepts, and practices in prosthodontics - 1976. J Prosthet Dent: 1977; 37: 204-221.

Stade EH, Stewart GP, Morris HF, Pesavento JR. Influence of fabrication technique on wrought wire clasp flexibility. J Prosthet Dent. 1985; 54: 538-542

Tada S, Ikebe K, Matsuda K, Maeda Y. Multifactorial risk assessment for survival of abutments of removable partial dentures based on practice based longitudinal study. J Dent. 2013; 41: 1175-1180.

Vallittu PK, Kokkonen M. Deflection fatigue of cobalt-chromium, titanium and gold alloy denture clasp. J Prosthet Dent. 1995; 74: 412- 419

VandenBrink JP, Wolfaardt JF, Faulkner GM. A comparison of various removable partial denture clasp materials and fabrication procedure for placing clasps on canine and premolar teeth. J Prosthet Dent. 1993; 70 :180-8.

Vanzeveren C, D’hoore W, Bercy P, Leloup G. Treatment with removable partial dentures: a longitudinal study. Part II. J Oral Rehab. 2003; 30: 459–469.

Waldmeier MD, Grasso JE, Norberg GJ, Nowak MD. Bend testing of wrought wire removable partial dentures alloys. J Prosthet Dent. 1996; 76: 559-65

Warr JA. Numerical system of clasp design. J Prosthet Dent. 1961; 11:1105-1111.

Worthington P, Lang B, Rubenstein J. Osseointegration in Dentistry. An overview. 2nd Edition. 2003. Quintessence, Chicago. 56-59

**Cited references**

Brumfield RC. Dental Gold Structures. 1953. J.F. Jelenko & Co., New York. Cited in Bates J. The mechanical properties of the cobalt-chromium alloy and their relation to partial denture design. Br Dent J. 1965; 119: 389-395.

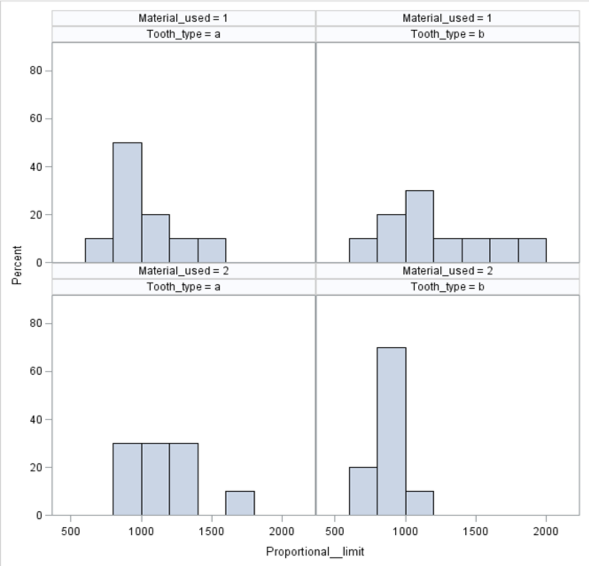
Taylor DF, Peyton FA. A comparison of the tensile and bending properties of dental gold wires. J Dent Res. 1951; 30: 290 – 301. Cited in Bates J. The mechanical properties of the cobalt-chromium alloy and their relation to partial denture design. Br Dent J. 1965; 119: 389-395.

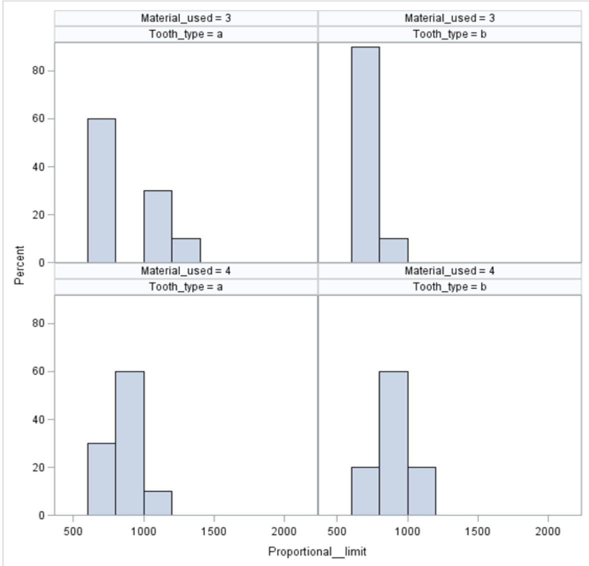
# APPENDIX 1

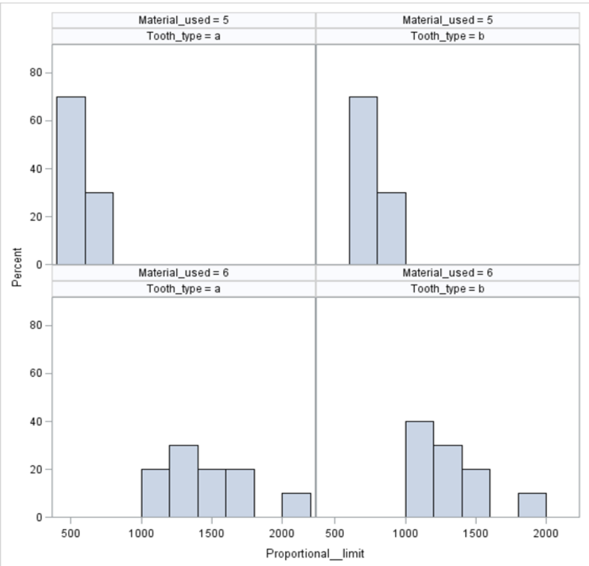
*Tabulated results from the experiments to determine RSD and realistic limit*

| Clasp Type | Tooth Type | Proportional limit (g) | | | | Deflection 0,25mm (g) | | | | Deflection 0,5 mm (g) | | | | Deflection 0,75 mm (g) | | | | Realistic limit (g) |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | n | mean | sd | rsd (%) | n | mean | sd | rsd (%) | n | mean | sd | rsd (%) | n | mean | sd | rsd (%) | |
| 0.9mm Leowire | premolar | 10 | 1003 | 215 | 21 | 10 | 303 | 45 | 15 | 10 | 583 | 96 | 16 | 10 | 849 | 151 | 18 | 572 |
| | molar | 10 | 1219 | 388 | 32 | 10 | 106 | 28 | 27 | 10 | 203 | 49 | 24 | 10 | 288 | 67 | 23 | 443 |
| 1.0 mm Leowire | premolar | 10 | 1130 | 233 | 21 | 10 | 676 | 118 | 17 | 10 | 1213 | 305 | 25 | 10 | 1755 | 424 | 24 | 664 |
| | molar | 10 | 856 | 81 | 10 | 10 | 359 | 62 | 17 | 10 | 657 | 116 | 18 | 10 | 916 | 157 | 17 | 693 |
| 0.9mm Remanium | premolar | 10 | 896 | 212 | 24 | 10 | 420 | 56 | 13 | 10 | 800 | 121 | 15 | 10 | 1162 | 172 | 15 | 473 |
| | molar | 10 | 699 | 74 | 11 | 10 | 124 | 21 | 17 | 10 | 235 | 38 | 16 | 10 | 334 | 54 | 16 | 550 |
| 1.0 mm Remanium | premolar | 10 | 858 | 140 | 16 | 10 | 535 | 66 | 12 | 10 | 1027 | 114 | 11 | 10 | 1475 | 173 | 12 | 578 |
| | molar | 10 | 915 | 142 | 16 | 10 | 219 | 27 | 12 | 10 | 417 | 51 | 12 | 10 | 604 | 72 | 12 | 631 |
| 0.9 mm Noninium | premolar | 10 | 574 | 54 | 9 | 10 | 360 | 65 | 18 | 10 | 619 | 109 | 18 | 10 | 818 | 136 | 17 | 467 |
| | molar | 10 | 705 | 123 | 17 | 10 | 160 | 38 | 24 | 10 | 274 | 47 | 17 | 10 | 363 | 54 | 15 | 458 |
| Vitallium | premolar | 10 | 1457 | 333 | 23 | 10 | 1179 | 287 | 24 | 10 | 1927 | 429 | 22 | 10 | 2364 | 422 | 18 | 792 |
| | molar | 10 | 1310 | 271 | 21 | 10 | 773 | 96 | 12 | 10 | 1464 | 167 | 11 | 10 | 1988 | 245 | 12 | 768 |

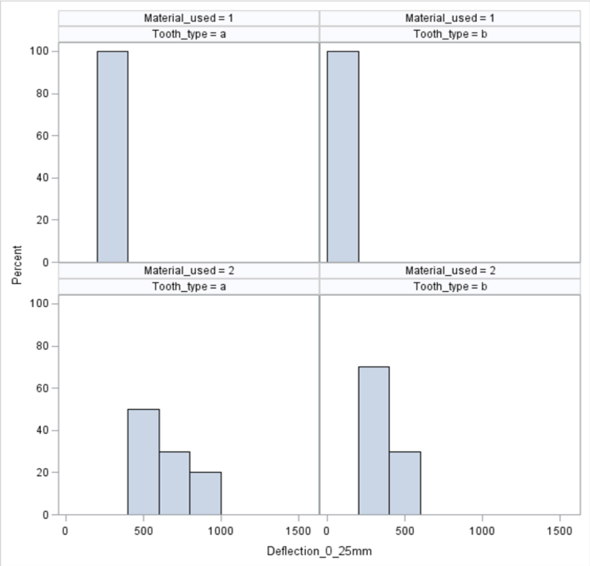
*Histograms inspected to identify and correct/exclude any outliers for the proportional limit*

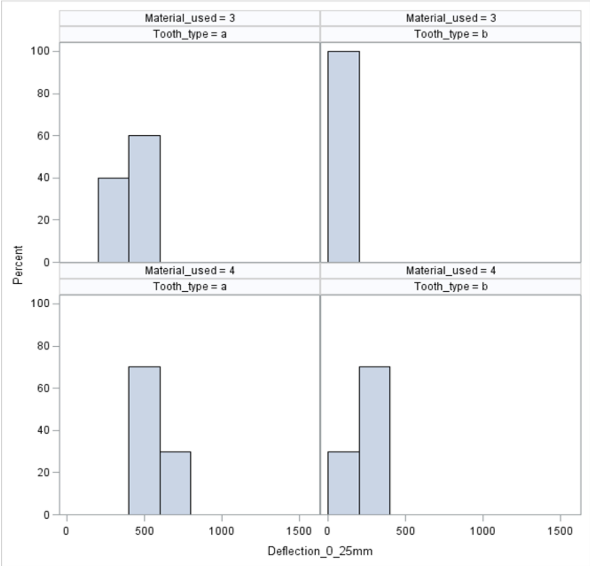


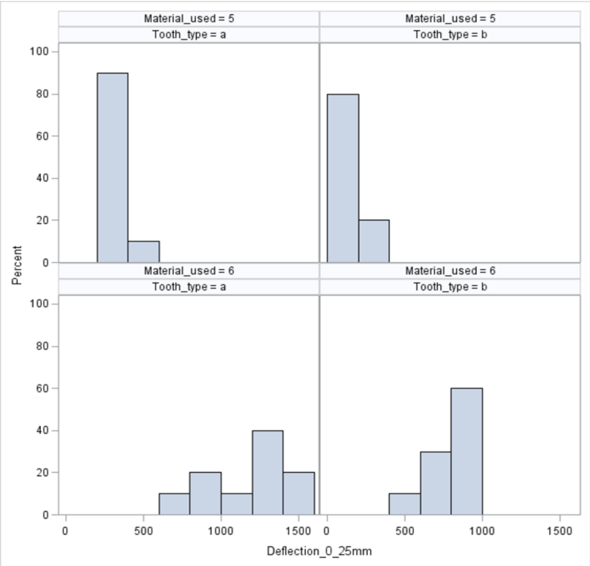



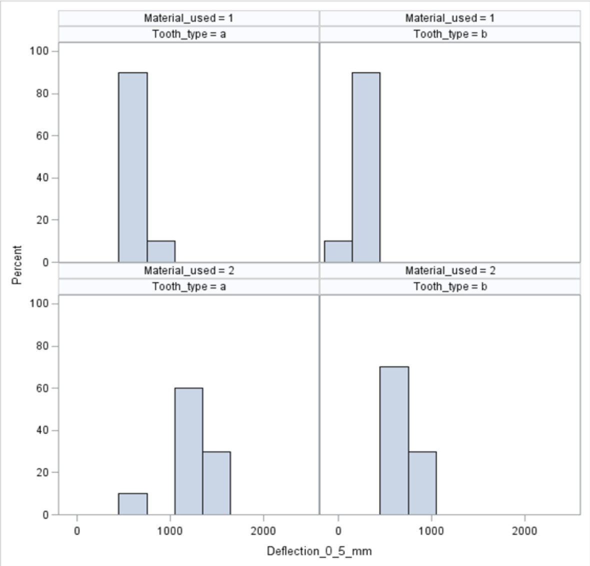
*Histograms inspected to identify and correct/exclude any outliers for the deflection at 0.25mm*

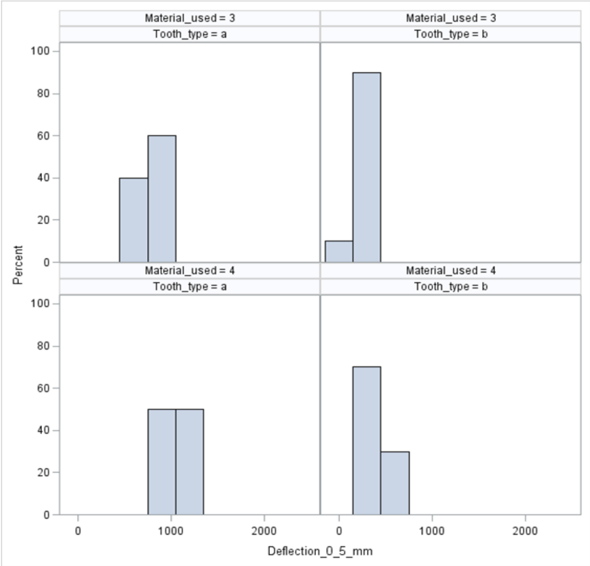


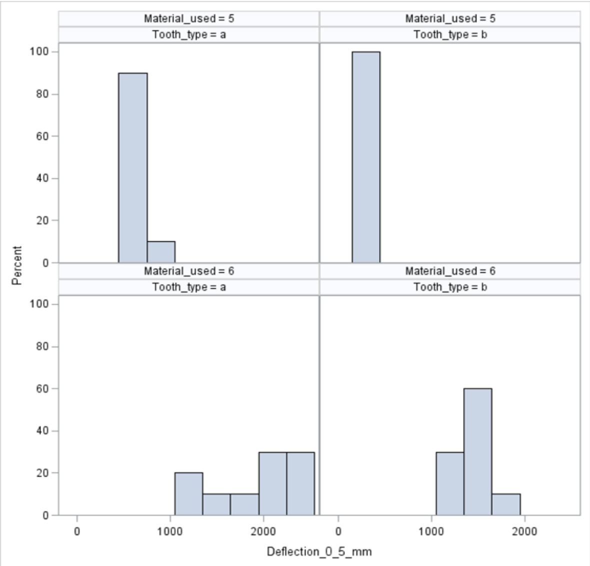



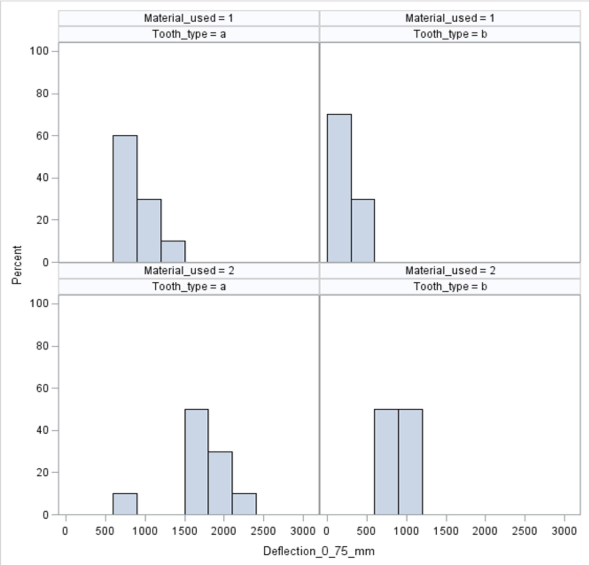
*Histograms inspected to identify and correct/exclude any outliers for the deflection at 0.50mm*

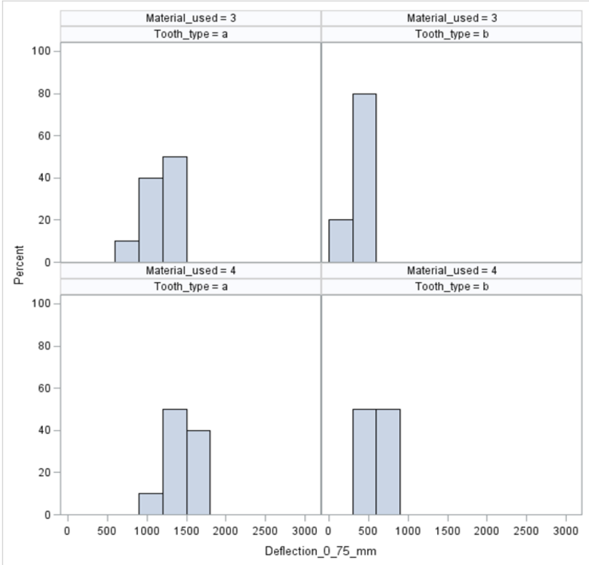


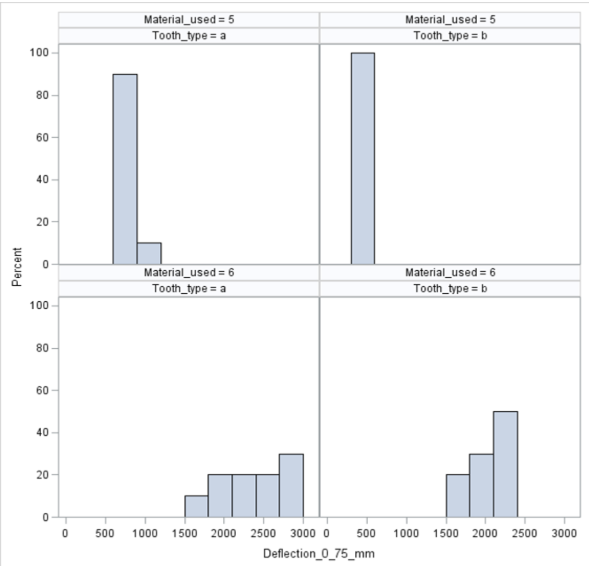



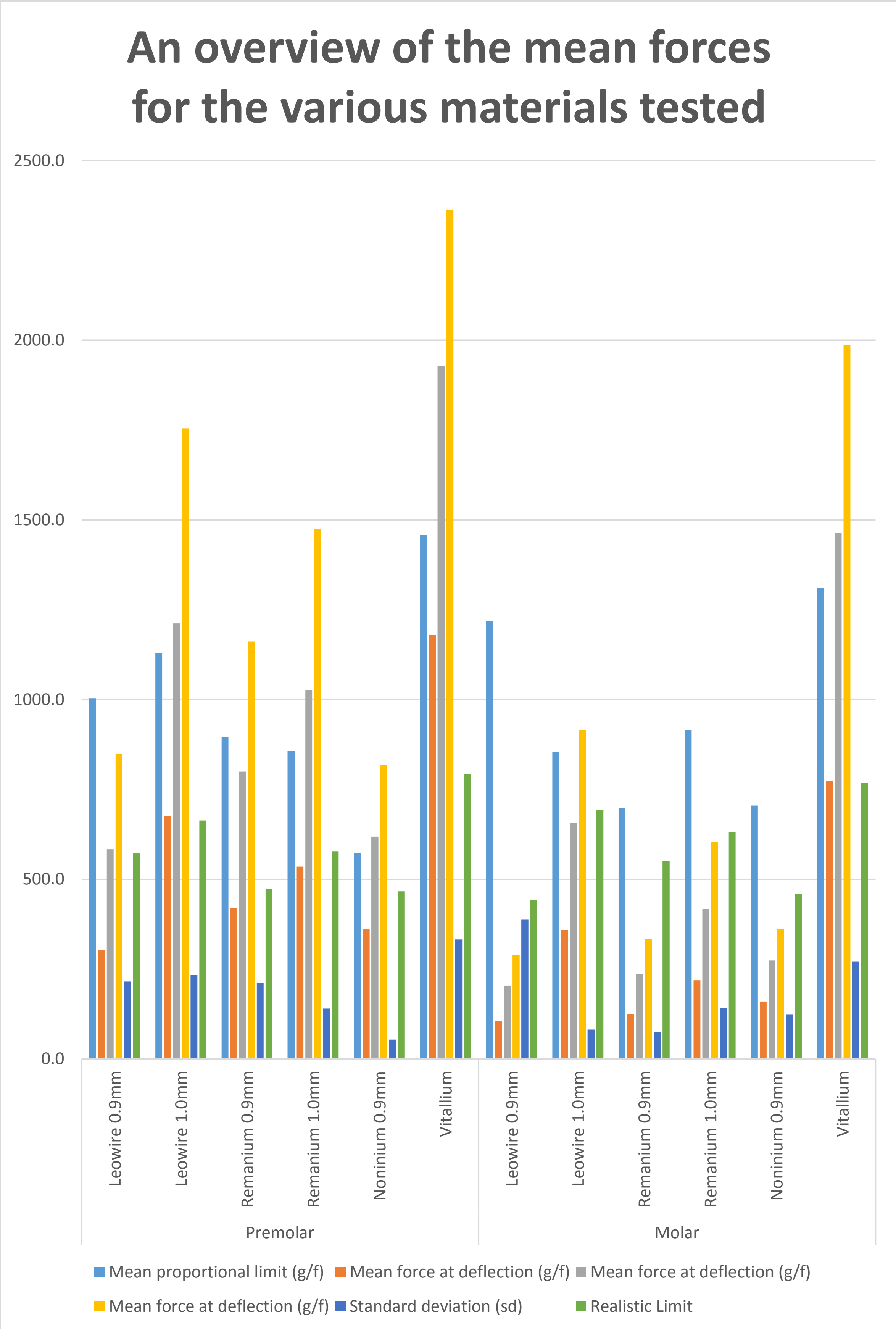
*Histograms inspected to identify and correct/exclude any outliers for the deflection at 0.75mm*



*An overview of the mean forces exerted for the various materials tested*



# APPENDIX 2: PLAGIARISM DECLARARTION

UNIVERSITY OF THE WITWATERSRAND, JOHANNESBURG

FACULTY OF HEALTH SCIENCES

PLAGIARISM DECLARATION TO BE SIGNED BY ALL HIGHER DEGREE STUDENTS

SENATE PLAGIARISM POLICY: APPENDIX ONE

I, Noland Naidoo (Student number: 586244) am a student

registered for the degree of Master of Dentistry (Prosthodontics) in the academic year 2016

I hereby declare the following:

- I am aware that plagiarism (the use of someone else's work without their permission and/or without acknowledging the original source) is wrong.
- I confirm that the work submitted for assessment for the above degree is my own unaided work except where I have explicitly indicated otherwise.
- I have followed the required conventions in referencing the thoughts and ideas of others.
- I understand that the University of the Witwatersrand may take disciplinary action against me if there is a belief that this is not my own unaided work or that I have failed to acknowledge the source of the ideas or words in my writing.

Signature: ________________________ Date: May 12, 2016