INTERNATIONAL DENTAL JOURNAL 72 (2022) 58–66

Scientific Research Report

Guidelines for the choice of circumferential wrought wire and cast clasp arms for removable partial dentures

C. Peter Owen [a*], Noland Naidoo [b]

[a] Professor Emeritus, Faculty of Health Sciences, University of the Witwatersrand, Johannesburg, South Africa
[b] Part-time Consultant Specialist, Department of Prosthodontics, School of Oral Health Science, Faculty of Health Sciences, University of the Witwatersrand, Johannesburg, South Africa

ARTICLE INFO

Article history:
Available online 20 February 2021

Key words:
Removable partial denture
Clasps
Undercuts
Forces
Safety limit

ABSTRACT

Purpose: The aim of this study was to establish guidelines for the selection of cast and wrought-wire clasps for removable partial dentures (RPDs) that would be appropriate for clinically encountered undercuts and facial curvatures.

Methods: Randomly selected discarded casts were collected and 30 premolars and 30 molars were surveyed, sectioned to a line representing the clasp and scanned using a flatbed scanner. The average clasp curvature and length for each group was determined and a three-dimensional model printed, to which wrought wire clasps of 0.9- and 1.0-mm diameter were adapted. Standard wax clasp patterns were adapted and cast in a stellite alloy. Each clasp was deformed beyond its proportional limit; and the forces exerted at that limit and at deflections of 0.25 mm, 0.5 mm, and 0.75 mm were measured, and a safety limit was calculated that would ensure elastic deformation at the required undercuts.

Results: A table was produced with guidelines for those clasps that would provide the highest retentive force within the proposed safety limit. The highest forces were provided by cast clasps in a 0.25-mm undercut. Wrought round wire of 1-mm diameter provided the next highest retentive forces, in a 0.25-mm undercut for premolar clasps arms and 0.5-mm for molar clasps.

Conclusions: The results provide valid guidelines for the use of combinations of clasp material and undercut that would exert the maximum retentive force without deformation for both short (premolar) as well as long (molar) clasps, for wrought and cast clasps.

Introduction

Tooth loss has a multifactorial aetiology and can have varying effects on an individual's quality of life.[1] The epidemiological literature suggests that there may be a decreasing number of patients afflicted with complete edentulism,[2,3] but that tooth loss continues to occur and may result in clinicians having to treat a higher incidence of patients who are partially edentulous. Treatment modalities include removable partial dentures (RPDs), fixed partial dentures (FPDs), implant supported prostheses (ISPs), or no treatment. The last option is, however, dependent on the patient's ability to adequately function in their current state or their preferences.

Globally, a variety of surveys have assessed need and usage of RPDs. A survey in the United States in 2002 found an increasing percentage of RPD wearing with age up to the >70 years cohort where there was a decrease, which was related to an increase in complete dentures.[4] The authors estimated that at least 250,000 people younger than age 40 had RPDs. A study[5] of self-reported data from 10,902 elderly subjects older than 60 years of age in 7 Latin American and Caribbean cities reported only on whether they had "bridges/dentures" and found that overall, 70% reported positively. Data from surveys in Europe[6] and Israel[7] found disparities between countries in terms of the average number of natural teeth remaining, which ranged from 15 to 27. Similarly there were country variations in the numbers of teeth

* *Corresponding author.* Faculty of Health Sciences, 7 York Road, Parktown 2193, Johannesburg, South Africa.
E-mail address: peter.owen@wits.ac.za (C.P. Owen).
https://doi.org/10.1016/j.identj.2021.01.005

replaced, related to the variable health services options available. A Chinese national oral health survey found that 27% in the 55 to 64 age group and 31% in the 65 to 74 age group wore RPDs.[8] In a suburban Nigerian community, a survey of subjects older than 65 years of age revealed that 48% were partially edentulous but only 7% had RPDs, mainly because of financial constraints.[9] In a national survey of a Polish population older than 65 years of age, 46% were partially edentulous, but 31% of those did not have RPDs.[10]

The replacement of missing teeth by means of RPDs is a well-established and cost-effective option. The responsibility for designing an RPD rests solely with the clinician and a detailed diagrammatic prescription should be provided to the dental technician.[11,12] This prescription should be based on evidence or clinical guidelines.[13] However, much of the design principles for RPDs are based on anecdotal reports or clinical experience rather than high levels of scientific evidence.[14] 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 numerous times without permanently deforming the clasp and without damaging the tooth.[11,15-19] The flexibility of a clasp is influenced by its length, diameter, cross-sectional form, and by its material.[14,18-23] If the amount of force required to overcome an undercut is beyond the proportional limit of the clasp arm, either the tooth will be affected or the clasp arm will be permanently deformed or will fracture.[14,17,24]

Davenport et al[14] reported 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 7 mm to overcome 0.5-mm undercuts without deforming. Their conclusions were, however, formulated from questionnaires that were sent to a selection of clinicians. Hence, their statements were a consensus of expert opinions and were not validated or substantiated.

Warr[25] used a mathematical analysis of clasp behaviour to predict loading forces relative to the proportional limit and suggested that the load exerted by moving in and out of an undercut should allow for a margin of safety and, in 1961,[26] proposed that the reason why clasps deformed or fractured was that they function too close to their proportional limit. However this ‘safety factor,’ or margin from the proportional limit, was never identified or tested. Bates[15] approached this 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 2 standard deviations and termed this the ‘realistic limit.’ It is logical that some form of limit should be applied to all clasp materials to avoid breakage, which is the likely outcome in cast metals, and permanent deformation, which is more likely in the more flexible wrought metals.

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.[16,19,20-22,27,28] It was therefore decided to conduct a study on clasps that reflected clinically encountered tooth curvatures and lengths to ascertain the force exerted on deflection at clinically encountered undercuts. Additionally, the proportional limit in relation to this deflection was calculated, to test whether Bates’s realistic limit was feasible and had clinical validity, or whether some other more clinically valid factor could act as a safety limit. The hypothesis was that there would be combinations of clasp materials, clasp lengths, and undercuts that would exceed the realistic limit as well as the proportional limit.

Methods and materials

Experimental design

This was an in vitro study designed to test a variety of clasp materials in terms of the forces they would exert when flexed at a deflection corresponding to the accepted undercuts present on premolar and molar teeth. The undercuts used were 0.25, 0.5, and 0.75 mm. Thirty premolars and molars were used and the clasp materials are shown in Table 1.

Sample size calculation

The relative margin of error was calculated and found to be acceptable for 30 premolars and 30 molars. The sample size for the different clasp materials and diameters was 10 per group. This was chosen in line with previous research, where the differences between the materials tested provided acceptable precision for the use of 10 samples per material [17,29,30]

Determining the average curvatures

Discarded casts from the dental laboratory attached to our prosthetics clinic were randomly and anonymously collected. Inclusion criteria were that they had undamaged first premolars and molars with no evidence of cavitation or fracture and excluded were casts where the premolars or molars showed any sign of damage. The casts were placed onto a dental surveyor and the maximum facial curvature (survey line) of each tooth was established in the normal manner. A line was then drawn to represent the position of a normal clasp arm.

The teeth were then sectioned into separate dies and each die was trimmed to the surveyed line representing the clasp

Table 1 – Clasp materials tested in the study.

Material	Diameter (mm)	Number of specimens	
		Premolar	Molar
Wrought wire			
Leowire*	0.9	10	10
	1.0	10	10
Remanium Hard†	0.9	10	10
	1.0	10	10
Noninium‡,§	0.9	10	10
Cast			
Vitallium¶		10	10

* Leowire (Leone), stainless steel.
† Remanium Hard (Dentaurum), stainless steel.
‡ Noninium (Dentaurum), nickel-free stainless steel.
§ Only 0.9 mm available.
¶ Vitallium (Dentsply), stellite alloy.

arm such that it would stand on a flat-bed scanner (Canon MG3540). The midpoint was marked to line up with the midpoint of a scanning template to establish a scale reference to ensure that magnification errors did not occur. Each die was scanned at 600 dpi. The images of the premolar and molar groups were then imported individually into the software package CorelDraw (Corel) and converted to mathematically derived vector graphics. The curves were then placed over each other, each onto a separate virtual layer. The homogeneity of the curves was such that an average curvature could be mapped, representing the determined average curvature for the teeth in that group.

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) and three-dimensional (3D) models of each curvature were created (Figure 1).

To adapt the wire samples, a solid model was required; so in the software, additional primitive objects were added, and the combined shape was extruded to allow adequate space for 10 wire samples to be adapted to the curve. A 5-mm ledge was created on the curve corresponding to the clasp tip to ensure that each sample terminated at the same point (Figure 2). Standard Tessellated Language (stl) files of these models were created and exported to a 3D printer (Objet 350v, Objet Inc.) and printed using the material Objet FullCure720 RGD720.

Constructing the clasp samples

The various wrought wire and cast clasp samples were adapted to the 3D-printed models by a single dental laboratory technician. The wrought wires were adapted in the same manner as they would be on a cast for use in a normal RPD. The cast clasps were all from a standard wax pattern with identical taper and cross-sectional shape. They were invested in the same flask, and each clasp sample was analysed under magnification by a single operator to identify any defects.

The clasps were then incorporated into acrylic resin blocks measuring 38 mm x 25 mm x 6 mm using a prefabricated mould; 10 samples were prepared at the same time. 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.

Tensile testing

The clasps embedded in acrylic blocks were tested in a tensile testing machine (Instron Corporation) using the Bluehill Lite software program (Instron). The clasp tips were engaged in a custom-made jig and displaced at a cross-head speed of 0.5 mm/min using a 2kN load cell, which was auto-balanced before each test to eliminate any measurement caused by the clasp tip touching the jig. Each clasp arm was then displaced until its proportional limit was exceeded. All the testing was performed by a single operator; all 120 samples were tested on the same day under the same conditions.

The software was programmed to record the proportional limit for each specimen as well as the load exerted at deflections of 0.25 mm, 0.5 mm, and 0.75 mm and was able to account for any slippage that occurred 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 sample (Figure 3).

Data analysis

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 calculated. The realistic limit (mean proportional limit minus 2 standard deviations)[15] was

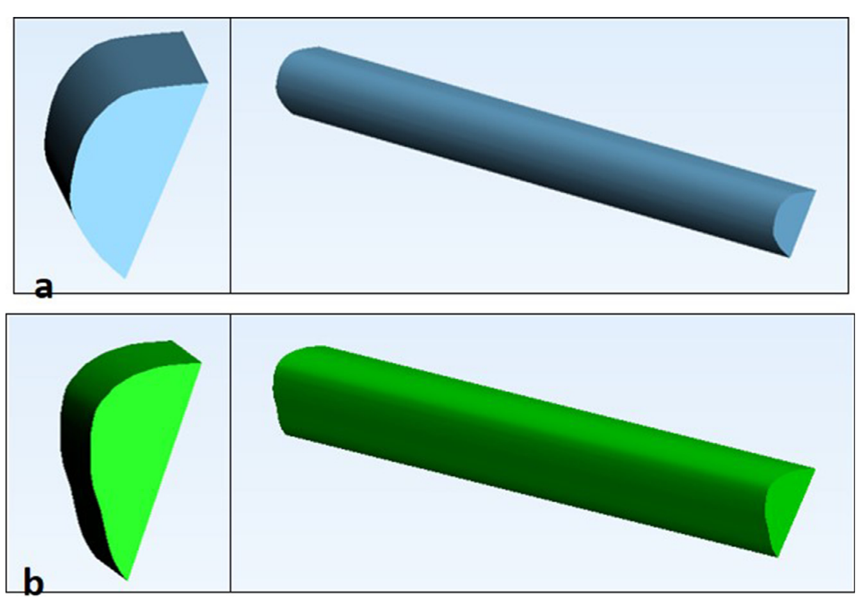

Fig. 1 – Three-dimensional models of the average curvatures. (A) premolar and (B) molar.

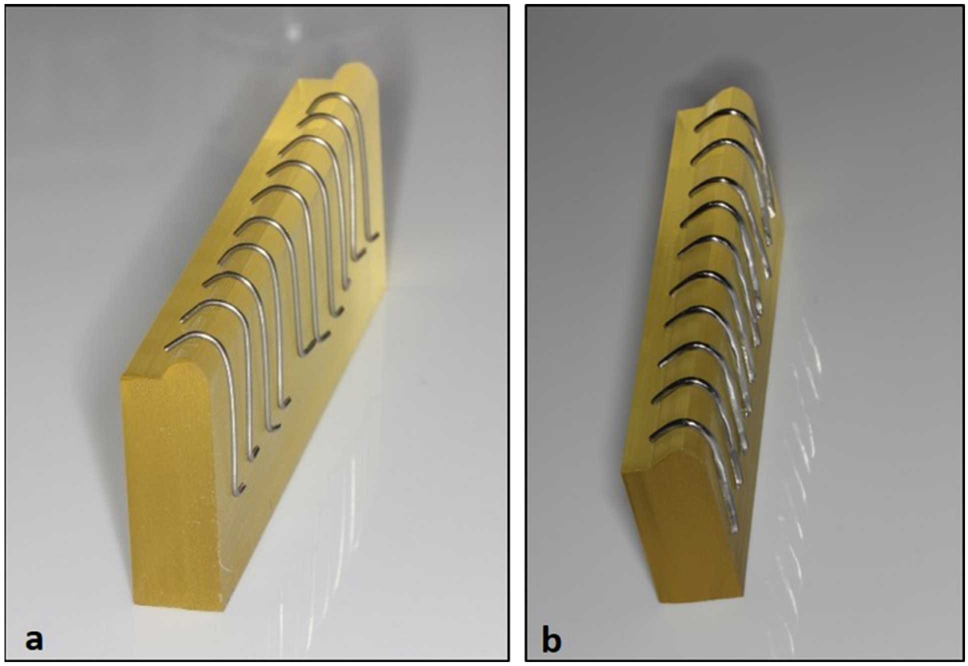

Fig. 2 – Examples of (A) wrought wire clasps and (B) cast clasps adapted to the three-dimensional models.

calculated for each experiment and expressed as a percentage of the proportional limit. Similarly, a factor of the proportional limit less 1 standard deviation was calculated and, again, expressed as a percentage of the proportional limit. A 1-sample *t*-test was used to determine the extent of the difference between the mean loads at the different deflections and these limits.

Results

The lengths of the determined average curvature for the premolar and molar dies selected were 9 mm and 14.5 mm, respectively. None of the wrought wire samples fractured during testing, but 1 molar cast clasp and 5 of the premolar cast clasps fractured.

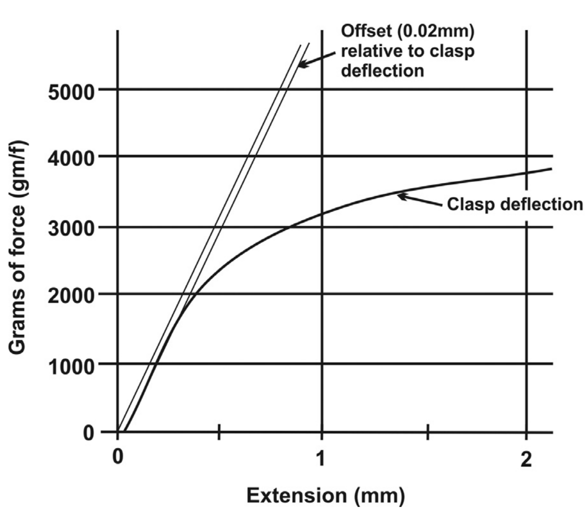

Fig. 3 – Example of a graph generated during the tensile testing for a premolar cast clasp.

Table 2 shows the results of the load tests for each combination of tooth type and clasp type. Table 3 shows the results for the calculations of Bates's realistic limit and is expressed as a percentage of the proportional limit for each wire. The mean percentage of the proportional limit was 63%, and it can be seen that at this limit that no cast clasps (Vitallium, Table 2) would be below this limit. As it is known clinically that cast clasps have been placed successfully on both premolars and molars, this realistic limit would appear to be too restrictive. Therefore, Table 3 also shows a proposed safety limit of 1 standard deviation from the proportional limit. The average difference of this limit from the proportional limit is 82%, and so it would seem reasonable to suggest that this now defines the safety limit. Table 4 then shows the effect of applying this safety limit to identify those materials that would fall within this limit and, therefore, could be used clinically. Figures 4 and 5 illustrate the force exerted at undercuts of 0.25 and 0.5 mm relative to the proportional limit and this safety limit.

Two of the values were not statistically significantly different from the mean proportional limit: the deflection of the 1.0-mm Leowire at an undercut of 0.5 mm for a molar clasp and the deflection of the Vitallium (cast) clasp at an undercut of 0.25 mm for a premolar clasp. However the former was within 5% of the safety limit and the latter within 1% of the safety limit. Therefore, it is possible to construct a table as in Table 5 that provides the clinical guidelines for the highest forces exerted within the safety limit. In addition, although nickel sensitivity is rare, it nevertheless does exist, and so the nickel-free wire Noninium is included in Table 5 at the appropriate diameter for premolar and molar clasps.

Table 2 – Tabulated results from the load tests for each experiment.

Clasp type	Tooth type	Proportional limit (g)				Deflection 0.25 mm (g)				Deflection 0.5 mm (g)				Deflection 0.75 mm (g)			
		n	mean	SD	RSD(%)	*n*	mean	SD	RSD(%)	*n*	mean	SD	RSD (%)	*n*	mean	SD	RSD (%)
0.9 mm Leowire	premolar	10	1003	215	21	10	303	45	15	10	583	96	16	10	849	151	18
	molar	10	1219	388	32	10	106	28	27	10	203	49	24	10	288	67	23
1.0 mm Leowire	premolar	10	1130	233	21	10	676	118	17	10	1213	305	25	10	1755	424	24
	molar	10	856	81	10	10	359	62	17	10	657	116	18	10	916	157	17
0.9 mm Remanium Hard	premolar	10	896	212	24	10	420	56	13	10	800	121	15	10	1162	172	15
	molar	10	699	74	11	10	124	21	17	10	235	38	16	10	334	54	16
1.0 mm Remanium Hard	premolar	10	858	140	16	10	535	66	12	10	1027	114	11	10	1475	173	12
	molar	10	915	142	16	10	219	27	12	10	417	51	12	10	604	72	12
0.9 mm Noninium	premolar	10	574	54	9	10	360	65	18	10	619	109	18	10	818	136	17
	molar	10	705	123	17	10	160	38	24	10	274	47	17	10	363	54	15
Vitallium	premolar	10	1457	333	23	10	1179	287	24	10	1927	429	22	10	2364	422	18
	molar	10	1310	271	21	10	773	96	12	10	1464	167	11	10	1988	245	12

RSD = relative standard deviation; SD = standard deviation.

Discussion

Permanent deformation and fatigue fracture are among the most common mechanical complications that can affect RPD clasps.[31-33] 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.

Studies have shown that clasps tend to lose their efficiency over time.[30] In 1 study, clasps exposed to repeated stress underwent permanent deformation and fatigue over a 36-month period.[29] This raises the question of whether constant deflection of the clasp during insertion and removal causes fatigue of the clasp [34] or whether the clasp functioned too close to its proportional limit.[15] Deformation of the clasp arm may lead to unfavourable stresses on the abutments and the RPD itself.[31]

Vallittu and Kokonen[34] used cyclic fatigue testing on straight wires by deflecting them 0.6 mm. They found a decrease in load (but gave no figures) and fatigue fracture occurring at about 25 loading cycles. However, the clinical applicability of such tests is questionable. Saito et al,[32] in a retrospective clinical evaluation of a variety of RPDs, found that clasp-retained dentures showed an increase in clasp fracture after 6 years but did not quantify this or identify the clasps. Cheng et al[35] tested 3 different cobalt-chromium cast alloys in a circumferential clasp assembly configured on a metal molar tooth with 0.25- and 0.5-mm undercuts and subjected them to an insertion and removal test for 7200 cycles to simulate 5 years of clinical use. They found a "marked decrease" in retentive force after 360 cycles and, thereafter, a gradual decrease. However, the amount of decrease was not provided, and the load at deflection was only determined by displacing a straight wire and not a wire corresponding to a circumferential clasp's curve. Nevertheless, the load at an undercut of 0.5 mm exceeded the proportional limit but not that at an undercut of 0.25 mm

Previous studies have made assumptions as to the length of the facial curvature of teeth. It has been reported that wrought wire clasps need to be at least 7 mm in length to overcome 0.5-mm undercuts without deforming, and that a cast clasp needs to be at least 15 mm[14] because anything shorter would be too rigid to disengage 0.25-mm undercuts

Table 3 – Mean PL less 2 SD (Bates's 'RL'), and the mean PL less 1 SD, both expressed as percentages of the PL. The last column creates a 'safety limit' based on the mean percentage of the PL less 1 SD as being clinically valid.

Clasp type	Tooth type	Bates RL (g) (= mean PL less 2 SD)	RL as % of PL	Mean PL less 1 SD	PL less 1 SD as % of PL	Safety limit: 82% of mean PL
0.9-mm Leowire	premolar	572	57	787	79	822
	molar	443	36	831	68	1000
1.0-mm Leowire	premolar	664	59	897	79	927
	molar	693	81	774	90	702
0.9-mm Remanium Hard	premolar	473	53	685	76	735
	molar	550	79	625	89	573
1.0-mm Remanium Hard	premolar	578	67	718	84	703
	molar	631	69	773	84	751
0.9-mm Noninium	premolar	467	81	520	91	470
	molar	458	65	582	83	578
Vitallium	premolar	792	54	1125	77	1195
	molar	768	59	1039	79	1074
		Mean	**63**	**Mean**	**82**	

PL = proportional limit; RL = realistic limit; SD = standard deviation.

Table 4 – Analysis of force at deflection relative to the safety limit. Forces higher than the safety limit are shaded.

Clasp Type	Tooth Type	PL	Deflection (g)			SL: 82% of PL	P value for mean Defl < SL		
			0.25-mm mean	0.5-mm mean	0.75-mm mean		Defl .25 mm	Defl .50 mm	Defl .75 mm
0.9-mm Leowire	premolar	1003	303	583	849	822	.0000	0.0000	Defl > SL
	molar	1219	106	203	288	1000	.0000	.0000	.0000
1.0-mm Leowire	premolar	1130	676	1213	1755	927	.0001	Defl > SL	Defl > SL
	molar	856	359	657	916	702	.0000	.2486*	Defl > SL
0.9-mm Remanium Hard	premolar	896	420	800	1162	735	.0000	Defl > SL	Defl > SL
	molar	699	124	235	334	573	.0000	.0000	.0000
1.0-mm Remanium Hard	premolar	858	535	1027	1475	703	.0000	Defl > SL	Defl > SL
	molar	915	219	417	604	751	.0000	.0000	.0001
0.9-mm Noninium	premolar	574	360	619	818	470	.0005	Defl > SL	Defl > SL
	molar	705	160	274	363	578	.0000	.0000	.0000
Vitallium	premolar	1457	1179	1927	2364	1195	.8625†	Defl > SL	Defl > SL
	molar	1310	773	1464	1988	1074	.0000	Defl > SL	Defl > SL

Defl = deflection; PL = proportional limit; SL = safety limit.

* Not significant: 77% of PL.
† Not significant: 81% of PL.

without permanently deforming the clasp 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 simulated the clinical length and curvature to which clasps for premolars and molars would be adapted. The average lengths were 9 mm for premolars and 14.5 mm for molars, and both length and diameter proved to be significant factors in the flexibility of the clasps. Not surprisingly, the retentive force of the clasps for a given undercut were also affected by the length of the clasp because the shorter the clasp and stiffer the clasp, the greater the force required for its flexure.

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 of these materials. Therefore, a margin of safety does need to be applied, but the concept of the realistic limit[8,15] was found to be clinically unrealistic. For this reason, it is proposed that a safety limit of 82% of the mean proportional limit be adopted as being clinically appropriate.

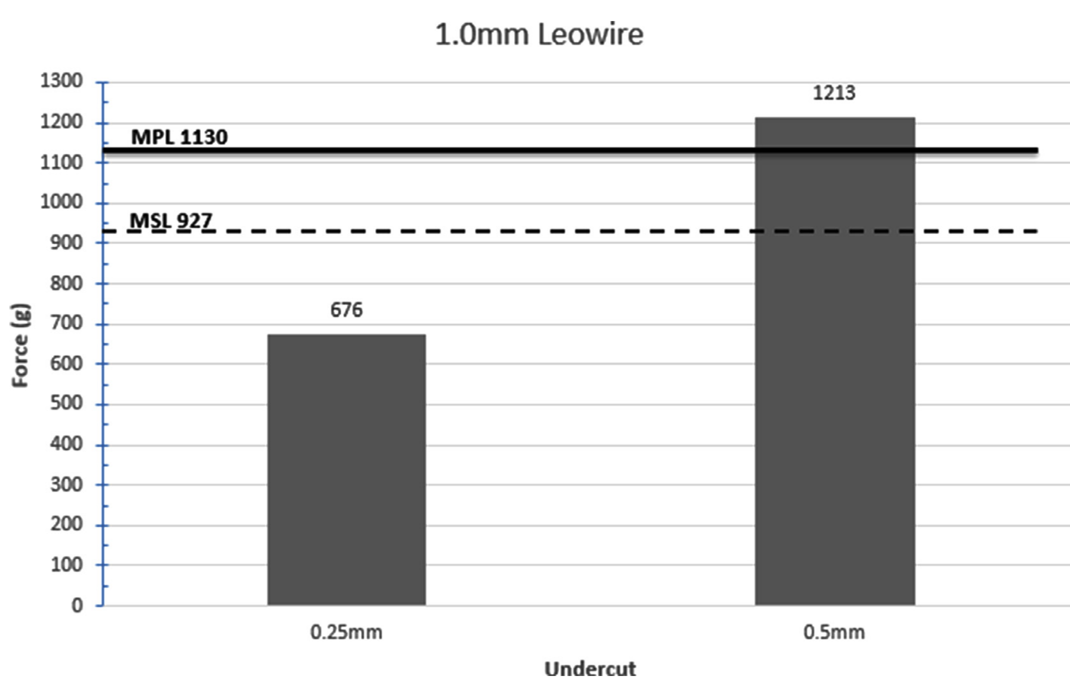

Fig. 4 – The force exerted by a short (premolar) 1.0-mm diameter stainless steel wire at undercuts of 0.25 and 0.5 mm relative to the MPL of the wire and the proposed MSL as being 82% of the MPL. MPL= mean proportional limit; MSL = mean safety limit.

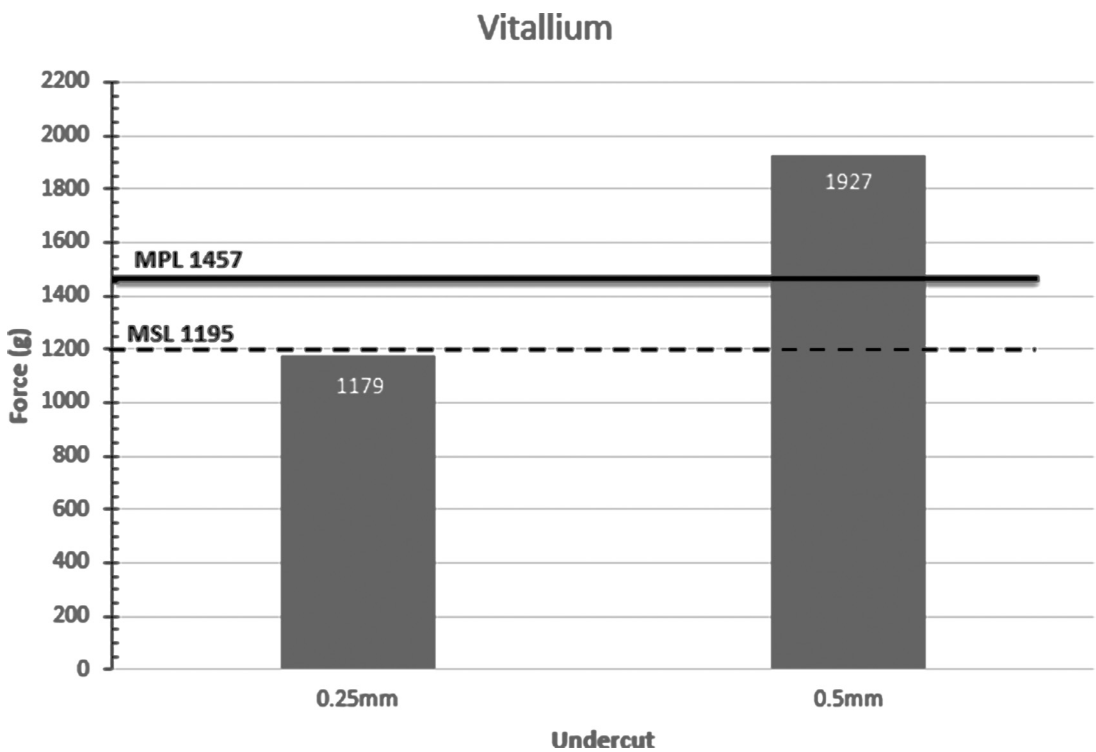

Fig. 5 – The force exerted by a short (premolar) cast wire at undercuts of 0.25 and 0.5 mm relative to the MPL of the wire and the proposed MSL as being 82% of the MPL. MPL= mean proportional limit; MSL = mean safety limit.

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. However, these forces have not been measured clinically, so it would seem wise to choose clasp arms that will provide the greatest retention at all times, provided that those clasp arms can function elastically, within the safety limit proposed here. The clasp materials and the undercuts that result from this analysis are shown in Table 5 as a clinical guideline for clasp selection. The design referred to throughout is that of a circumferential clasp arm, so these guidelines would not apply to gingivally approaching clasps.

The hypothesis was accepted in that it was found that several material, clasp length, and undercut combinations produced forces that exceeded the realistic limit and the proportional limit. The realistic limit was found to be too stringent, and a new safety limit was proposed as the average of the proportional limit minus 1 standard deviation.

This study has shown that cast clasps should not exceed a 0.25-mm undercut and that even that was close to the proposed safety limit for premolars (or teeth of equivalent size). Casting the cast clasps together in 1 casting was a limitation of this study because this did not allow for natural variations that occur during casting. A follow-up, although labor-intensive study, where each clasp is part of a cast framework would allow for the natural variations that occur during the casting process and may be more clinically applicable. In addition, this study should be replicated with other makes of wrought and cast materials.

The flexibility of wrought wire is influenced by its alloy type, diameter, length, and depth of undercut. In this study, the highest force exerted by a wrought wire and within the safety limit, was 676 g from a stainless steel wire of 1.0-mm diameter for a 0.25-mm undercut. For a molar clasp length, the highest force within the safety limit was 657 g from a stainless steel wire of 1.0-mm diameter engaging a 0.5-mm undercut.

These results are difficult to compare with other studies, which have reported either clinician opinions,[14] removal force without recording deformation,[27] standardised curvatures, or straight wires[19,28] and often just to determine the proportional limit.[16,20-22] This is the first study to use clasps conforming to the natural curvatures of premolars and molars. In addition, wrought wire free of nickel was tested

Table 5 – Clinical guidelines for the clasp selection for molars and premolars based on the highest loads exerted within the safety limit and on the wires tested.

Undercut	Premolars 0.25 mm	Molars 0.25 mm	0.5 mm	0.75 mm
Clasp material (mean force in grams)	Vitallium (1179) Leowire 1.0 mm (676)	Vitallium (773)	Leowire 1.0-mm (657)	
For nickel-sensitive patients (mean force in grams)	Noninium 0.9 mm (360)			Noninium 0.9 mm (363)

and found to be applicable but was more flexible than stainless steel and, therefore, exerted less retentive force.

Finally, acrylic resin base RPDs are often regarded as interim prostheses, but with the provision of both anterior tooth support through acrylic resin rests, posterior tooth support through the use of, for example, half-round wire, and the use in particular of guide plane retention, wrought wire clasps need provide additional retention only when required.[36] In a world characterised by increasing inequality and an economic order where the majority of patients who are partially edentulous are likely to have limited access to the more expensive treatment methods and even to cast metal framework RPDs, this type of RPD could then become a more permanent and cost-effective prosthesis.[12]

Conclusion

Several combinations of clasp material, length, and undercut were found to be unsuitable because they resulted in clasp deformation at the required undercut that exceeded the proportional limit as well as the originally suggested realistic limit. The hypothesis that this would be the case was therefore accepted. A proposed safety limit of 82% of the proportional limit was proposed to ensure that clasp deflection would remain elastic, although a cast clasp at an undercut of 0.25 mm for a premolar clasp was within 1% of the safety limit and should be used with caution. The combinations of clasp material and undercut that would exert the maximum retentive force within this safety limit were found for both short (eg, premolar) and long (molar) wrought wire and cast clasps. These were produced in a summary table to provide clinical guidelines for clasp selection for RPDs.

Conflict of interest

None disclosed.

Acknowledgements

This study was developed from previous unpublished studies but using only standardised nonclinical curvatures of clasp arms; we are grateful for Drs L. M. Naidoo, M-R. Lekay-Adams, and R. Goolam for these previous studies.

Funding

None disclosed.

REFERENCES

1. 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;24:109–17.
2. 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;63:21–5.
3. Campbell SD, Cooper L, Craddock H, et al. Removable partial dentures: the clinical need for innovation. J Prosthet Dent 2017;118:273–80.
4. Hummel SK, Wilson MA, Marker VA, Nunn ME. Quality of removable partial dentures worn by the adult U.S. population. J Prosthet Dent 2002;88:37–43.
5. Singh H, Maharaj RG, Naid R. Oral health among the elderly in 7 Latin American and Caribbean Cities, 1999-2000: a cross-sectional study. BMC Oral Health 2015;15:46.
6. Börsch-Supan A, Brandt M, Hunkler C, et al. Data resource profile: the survey of health, ageing and retirement in Europe (SHARE). Int J Epidemiol 2013;42:992–1001.
7. Stock C, Jürges H, Shen J, Bozorgmehr K, Listl S. A comparison of tooth retention and replacement across 15 countries in the over-50s. Community Dent Oral Epidemiol 2016;44:223–31.
8. Guo J, Ban JH, Li G, et al. Status of tooth loss and denture restoration in Chinese adult population: findings from the 4th national oral health survey. Chin J Dent Res 2018;21:249–57.
9. Ibiyemi O, Lawal FB. A short report on tooth replacement in an older suburban population in Nigeria. Gerodontology 2017;34:508–11.
10. Mehr K, Olszanecka-Glinianowicz M, Chudek J, et al. Dental status in the polish senior population and its correlates-results of the national survey polSenior. Gerodontology 2018;35:398–406.
11. Sharry J. Principles, concepts, and practices in prosthodontics - 1976. J Prosthet Dent 1977;37:204–21.
12. Owen CP. Fundamentals of removable partial dentures. 4th ed. Available from: www.appropriatech.com. Accessed January 14, 2020.
13. Firtell DN, Muncheryan AM, Green AJ. Laboratory accuracy in casting removable partial denture frameworks. J Prosthet Dent 1985;54:856–62.
14. Davenport JC, Basker RM, Heath JR, et al. Clasp design. Br Dent J 2001;190:71–81.
15. Bates JF. The mechanical properties of the cobalt-chromium alloy and their relation to partial denture design. Br Dent J 1965;119:389–95.
16. Frank RP, Nicholls JI. A study of the flexibility of wrought wire clasps. J Prosthet Dent 1981;45:259–67.
17. Matheson GR, Brudvik JS, Nicholls JI. Behavior of wrought wire clasps after repeated permanent deformation. J Prosthet Dent 1986;55:226–31.
18. Brockhurst PJ. A new design for partial denture circumferential clasp arms. Aus Dent J 1996;41:317–23.
19. Waldmeier MD, Grasso JE, Norberg GJ, et al. Bend testing of wrought wire removable partial dentures alloys. J Prosthet Dent 1996;76:559–65.
20. 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 behavior. J Prosthet Dent 1981;46:133–41.
21. Morris HF, Asgar K, Roberts EP, et al. Stress-relaxation testing. Part II: comparison of bending profiles, microstructures, microhardnesses and surface characteristics of several wrought wires. J Prosthet Dent 1981;46:256–62.
22. Morris HF, Asgar K, Brudvik JS, et al. Stress-relaxation testing. Part IV: clasp pattern dimensions and their influence on clasp behavior. J Prosthet Dent 1983;50:319–26.
23. Stade EH, Stewart GP, Morris HF, et al. Influence of fabrication technique on wrought wire clasp flexibility. J Prosthet Dent 1985;54:538–42.
24. Bates JF. Retention of partial dentures. Br Dent J 1980;149:171–4.
25. Warr JA. An analysis of clasp design in partial dentures. Phys. Med Biol 1959;3:212–32.
26. Warr JA. Numerical system of clasp design. J Prosthet Dent 1961;11:1105–11.
27. Firtell DN. Effect of clasp design upon retention of removable partial dentures. J Prosthet Dent 1968;20:43–52.

28. VandenBrink JP, Wolfaardt JF, Faulkner MG. 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.
29. Arda T, Arikan A. An in vitro comparison of retentive force and deformation of acetal resin and cobalt-chromium clasps. J Prosthet Dent 2005;94:267–74.
30. Davenport JC, Basker RM, Heath JR, et al. Retention. Br Dent J 2000;189:646–57.
31. Keltjens HM, Mulder J, Käyser AF, et al. Fit of direct retainers in removable partial dentures after 8 years of use. J Oral Rehab 1997;24:138–42.
32. Saito M, Notani K, Miura Y, et al. Complications and failures in removable partial dentures: a clinical evaluation. J Oral Rehab 2002;29:627–33.
33. 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;23:317–24.
34. Vallittu PK, Kokkonen M. Deflection fatigue of cobalt-chromium, titanium and gold alloy denture clasp. J Prosthet Dent 1995;74:412–9.
35. Cheng H, Xu M, Zhang H, et al. Cyclic fatigue properties of cobalt-chromium alloy clasps for partial removable dental prostheses. J Prosthet Dent 2010;104:389–96.
36. Mothopi-Peri MM, Owen CP. Guide-plane retention in designing removable partial dentures. Int J Prosthodont 2018;31:145–8.