COMPARISON OF THE FRACTURE RESISTANCE OF METAL REINFORCED ACRYLIC VS. ACRYLIC ONLY DISTAL EXTENSION CANTILEVERS OF FIXED IMPLANT SUPPORTED PROSTHESES.

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DECLARATION

I, Martin Alexander Osswald, declare that this research report is my own work. It is submitted for the Degree of Master of Dentistry in the University of the Witwatersrand, Johannesburg. It has not been submitted before for any other degree or examination at this or any other university.

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ABSTRACT

Purpose: The aim of this research project was to determine the fracture resistance to linear vertical compressive forces of acrylic and metal-reinforced acrylic fixed implant supported prosthesis cantilever arms. **Methods:** Ten non-reinforced and 10 reinforced acrylic superstructures were secured on five evenly distributed implants and subjected to linear axial compressive forces utilising an Instron® materials testing machine. The reinforcement consisted of commercially available preformed titanium metal strengthener bars. Force was applied to the cantilever arms 15mm distal to the distal most implant and two tests were conducted on each sample. The first drop in load recorded was noted as a fracture of the sample. **Results:** Fracture of the acrylic was noted at the distal most implant in both non-reinforced and the reinforced samples. The mean fracture value for the non-reinforced samples was 679N and for the reinforced samples, 628N. **Conclusion:** No significant difference between the fracture resistance of the two designs was noted.

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1. INTRODUCTION

The high success rates of endosseous dental implants in terms of osseointegration and long term survival have been well documented. Rigid splinting of multiple osseointegrated implants by a superstructure is generally still recommended. Rigid splinting in early and immediate loading of implants to limit micromotion to within 150µ is also a key factor for successful osseointegration (Maniatopoulos et al 1986; Pilliar et al 1986, cited by Szmukler-Moncler 1998).

More emphasis has recently been placed on factors affecting the survival of the implant superstructure or prosthesis (Goodacre et al 2003). However, there is a general paucity in the literature of studies describing or prescribing norms and ideals in terms of materials and design of the actual prosthesis. The benefits of utilising acrylic as the restorative material for implant supported prostheses are apparent on many levels. The first is a cost saving when compared to materials such as cast metal, porcelain or milled titanium. Acrylic prostheses are also relatively easy to construct in a short period of time resulting in decreased laboratory costs. Acrylic is rigid yet easy to adjust, both extra and intra-orally. These factors make acrylic ideal as the material of choice for immediate and early loading protocols. The potential cost benefits could also make implant treatment a more affordable treatment modality for more patients.

2. LITERATURE REVIEW

Many different materials and prosthesis designs have been described in studies, particularly in the literature on immediate loading. The literature review revealed studies using acrylic only for the fixed implant supported prosthesis (ISP) (Degidi et al 2003; Nikellis, Levi & Nicolopoulos 2004; Wolfinger, Balshi & Rangert 2003; Balshi & Wolfinger 1997; Schnitman et al 1997); reinforced acrylic with stainless steel bars (Wolfinger et al 2003), cast metal (Tarnow, Emtiaz & Classi 1997; Horiuchi et al 2000; Grunder 2001), milled titanium frames (Wolfinger et al 2003; Balshi & Wolfinger 1997; Grunder 2001; Kronstrom et al 2003; Maló, Rangert & Nobre 2003), and orthodontic wire (Nikellis et al 2004). Some papers failed to describe the metal reinforcement (Glauser et al 2001; Glauser et al 2003). Fibre reinforcement of acrylic was also noted (Glauser et al 2001; Glauser et al 2003). The complete absence of any description of superstructure design was noted in some studies (Jaffin, Kumar & Berman 2000; Salama et al 1995; Rocci, Martignoni & Gottlow 2003; Misch & Degidi 2003; Romanos 2004).

However, not one of these studies described the design of the superstructure in terms of dimensions nor had any been related to or based on any evidence-based testing such as simple fracture resistance to compressive forces. In addition, no reference was found in the literature between the various designs, fracture resistance and *in-vivo* clinically measured occlusal loading. The effect of distal extension cantilevers on fixed ISP strength has also not been described.

The studies reported on the success rates of osseointegration with immediate or early loading but the influence of the chosen prosthetic material or design was not made clear. It is possible that the material and design used may not have been important as long as it offered a rigid splinting effect. If this is the case then non-reinforced acrylic may be sufficient and have the potential to considerably reduce costs.

3. AIM

The aim of this research project was to determine the maximum compressive load or break resistance of acrylic and metal-reinforced acrylic implant supported prostheses when subjected to linear vertical compressive forces on a distal cantilever.

The null hypothesis was that a cantilevered standardised acrylic superstructure will resist fracture at clinically encountered forces and would not improve with metal reinforcing.

4. MATERIALS AND METHODS

All implant components were from the current catalogue of and manufactured by Southern Implants (Irene, South Africa). Five 3.75mm standard diameter IBS implants were secured in an aluminium block (Fig 1). Their distribution was such as to mimic an optimal distribution for the maxilla as described by Skalak (1983).

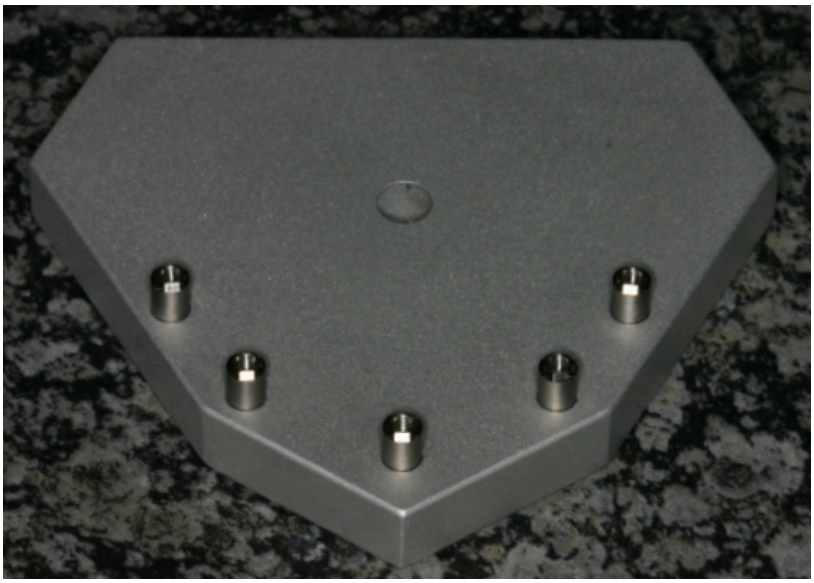

Fig 1. Aluminium block housing the implants

The acrylic super-structures were all replicated from a master template. The template dimensions were standardised to 10x10mm. These measurements were based on average measurements of actual prostheses placed for patients treated in the Department of Prosthodontics, School of Oral Health Sciences, University of the Witwatersrand. Abutments (TCB1nh) were secured to the aluminum model and the template was constructed by hand in self-curing acrylic and carefully adjusted to conform to the chosen dimensions. The template was cantilevered to 20mm beyond the distal implants. Passivity of fit of the template was confirmed on completion.

Laboratory analogues (LS12) were attached to the template with brass screws and embedded in stone, in one half of a flask, filled up to the lower border of the acrylic template (Fig 2). The top half of the flask was filled with laboratory silicone putty and was tightly clamped to the lower half until the silicone had set (Fig 3).

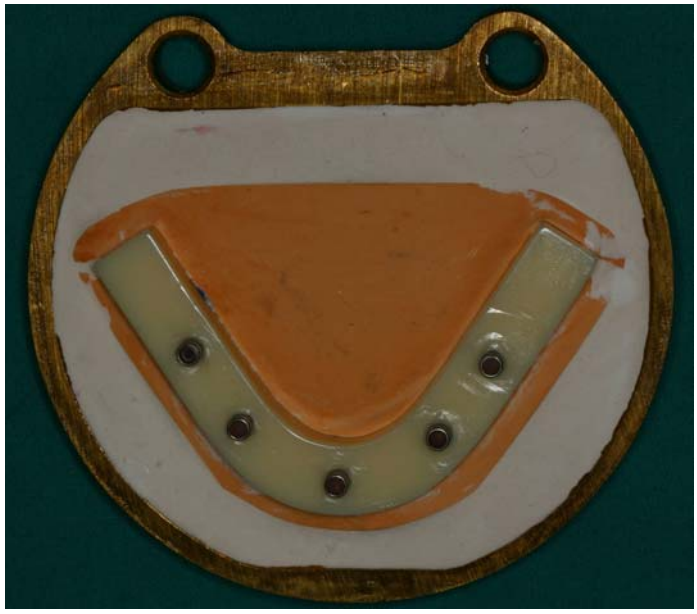

Fig 2. Template replication

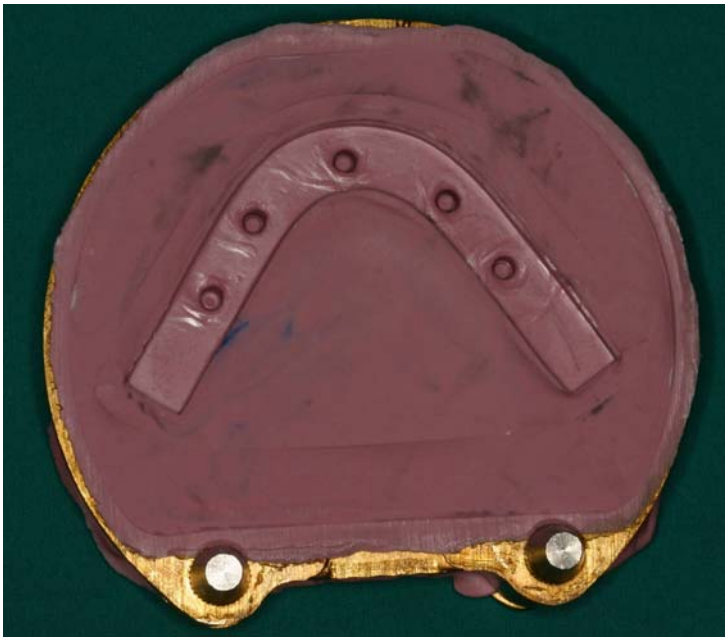

Fig 3. Silicone template replication

The template was then removed and new TCB1nh abutments were secured with brass screws on the embedded laboratory implant analogues. Acrylic superstructures were then constructed using Vertex Rapid Simplified (Vertex-Dental B.V., Zeist, The Netherlands). The monomer and polymer were mixed and cured strictly according to the manufacturer's instructions and placed in the recess in the silicone created by the

template. The ambient air temperature and humidity at time of flasking were recorded for each batch of samples (Table 1).

Table 1. Flasking conditions

Batch Number	Reinforcement	Air Temperature	Air Humidity
1	NR	25°C	50%
1	R	27°C	40%
2	NR	23°C	65%
2	R	24°C	82%

NR = non-reinforced, R = reinforced

The acrylic was heat-cured as per the manufacturer's instructions. The cured acrylic superstructures were subjected to minimal finishing: the acrylic flash was removed but no further trimming or polishing was performed to ensure that the dimensions of the samples did not deviate from the template (Fig 4). Degradation of the silicone during the curing process was observed during the pilot studies. It was therefore decided to prepare a new silicone flask for each sample.

All the prostheses were cantilevered to 20mm beyond the distal implants on either side. A mark was placed on the occlusal surface of each cantilevered section 15mm distal to the centre of the distal implant abutment, marking the point of force application. Fifteen millimeters is the maximum recommended cantilever extension for ISPs (Shackleton et al 1994).

Fig 4. Finished acrylic superstructure

The reinforced samples were manufactured with the same process as for the non-reinforced acrylic samples except that metal strengtheners were incorporated into the cured acrylic (Fig 5a). Titanium strengtheners (BAS1) were cut to size and made to interlock with ‘L’-shaped cut-aways (Fig 5b), as per the manufacturer’s recommendations. The strengtheners were secured to the abutments with a small amount of self-cure acrylic before packing the heat cured acrylic.

Fig 5a. Metal strengtheners

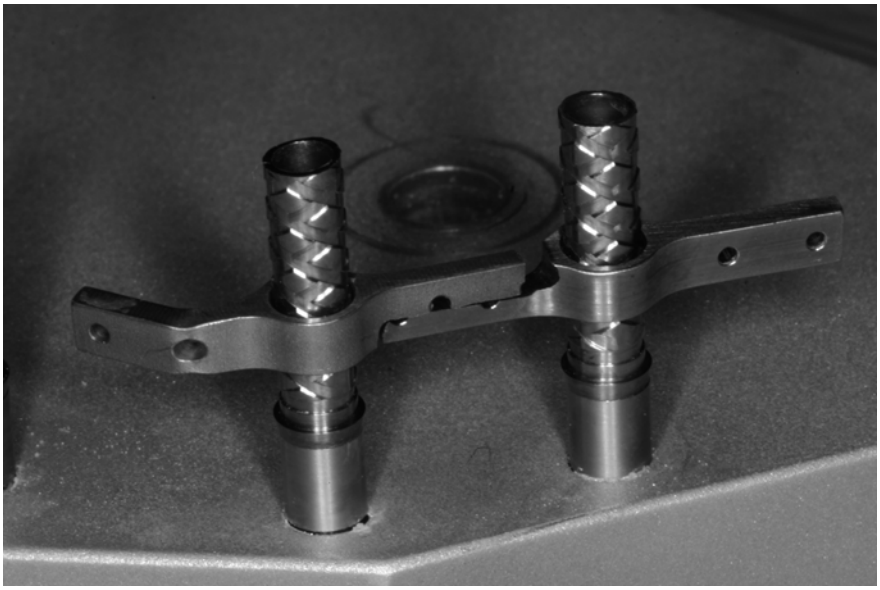

Fig 5b. L-shaped cut-aways

All the finished samples were again checked for passivity of fit on the aluminium model, and temperature and humidity also recorded (Table 1).

Ten reinforced and ten non-reinforced acrylic samples were constructed in four different batches. Each batch contained five samples all packed with a single mix of acrylic.

A cone-shaped stainless steel probe with a 2mm radius tip was constructed to mimic the dimension of the palatal cusp of a maxillary first molar (Fig 6). A stainless steel model mount was custom manufactured to fit the tensile testing machine. The platform was designed to allow the prosthesis to be rotated so that the test point would be directly axial to the line of force application thereby eliminating any off-axis force vectors (Fig 7).

Fig 6. Cone-shaped steel probe

Fig 7. Platform allowing axial alignment

All testing was performed on an Instron® (Model 3366, Grove City, Pennsylvania, USA) materials testing unit. The load cell was electronically adjusted by the testing machine and was capable of delivering loads from 1-10 000N. A pilot study had established that a cross-head speed of 2mm/minute was sensitive enough to accurately record the maximum compressive strength of the samples.

Testing was conducted over two days. Each sample was secured to the implants in the aluminium base using titanium screws (TSS2). These were tightened to 32Ncm as per the manufacturer's instructions using a calibrated torque wrench.
The aluminium base and attached sample was secured to the mounting jig and the probe was lowered to contact the mark made on the distal cantilever. The Instron machine was set to apply a 10N preload before calibrating to zero and commencing with the test. This was done to eliminate readings from any play in the test set up. The probe was set to travel a 2mm distance before terminating the test. The pilot study indicated that the samples would fracture within this limit.

First the left and then the right cantilever arm of each sample was subjected to compression forces. This allowed for the left of the reinforced and left of the un-reinforced samples to be compared, and also for the possibility of a cross-over study between the left and right within each sample.

The testing room temperature was controlled and constant at 22°C with the humidity varying between 60 and 62%.

The results were statistically analysed using Arcus Quickstat (Biomedical Version 1.0) software.

5. RESULTS

Fracture resistance values in Newtons (N) for all the samples are listed in tables 2-5. The first drop in load was noted and recorded as a fracture of the prosthesis.

Table 2. Test Results Non-reinforced Samples – Batch One, values in Newtons

Sample	Left Side	Right Side
1	729	781
2	690	600
3	571	490
4	664	499
5	757	545

Table 3. Test Results Non-reinforced Samples – Batch Two, values in Newtons

Sample	Left Side	Right Side
1	817	622
2	549	412
3	562	512
4	909	406
5	541	381

Table 4. Test Results Reinforced Samples – Batch One, values in Newtons

Sample	Left Side	Right Side
1	987	895
2	424	921
3	1035	647
4	681	695
5	781	701

Table 5. Test Results Reinforced Samples – Batch Two, values in Newtons

Sample	Left Side	Right Side
1	506	391
2	497	403
3	426	-*
4	505	407
5	435	460

* this reading was lost due to a breakage of the right cantilever arm during the experimental set-up and alignment.

The mean fracture resistance for all the left side non-reinforced samples was 679N (SD 126N) and for the left side reinforced samples was 628N (SD 233N).

For all the samples, both reinforced and non-reinforced, fracture of the acrylic occurred at the distal implant abutment. The distal cantilevers of the non-reinforced samples all fractured off the abutment (Fig 8). The reinforced samples all displayed a fracture line through the acrylic but the distal cantilevers did not fracture off the remaining acrylic structure due to the embedded support from the metal strengtheners (Fig 9). The reinforced samples could therefore be loaded beyond the first drop in load/fracture but for comparative purposes the test was stopped at this point.

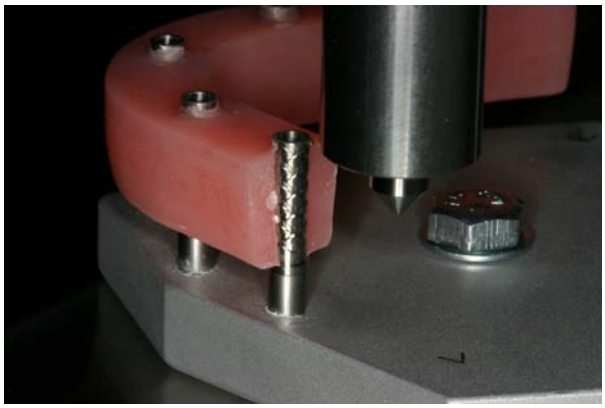

Fig 8. Fracture of non-reinforced sample

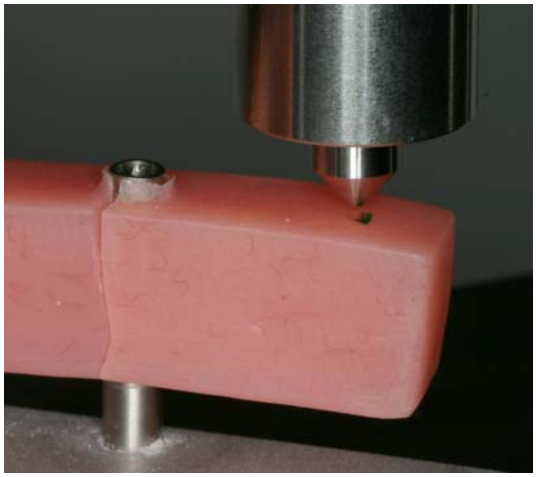

Fig 9. Fracture of reinforced sample

The study was designed so that two tests could be carried out on each sample. First the left and then the right cantilever arm of each sample was tested. This was done to determine whether a crossover study could be performed thereby doubling the sample size and decreasing the cost for similar studies in future. A two sided t-test showed that there was a significant difference between the fracture resistance of the left and right non-reinforced acrylic cantilever arms ($p=0.0085$) A cross over study was therefore not possible and only the first (left) cantilever arm fracture values of all the samples were considered for all subsequent analyses.

As there was a significant difference in the variance of the two reinforced batches the Mann-Whitney test was used. The analyses were as follows:

1. There was no significant difference in fracture resistance between batches one and two for the non-reinforced samples (p=0.9395)
2. There was no significant difference between the two reinforced batches (p=0.1508).
3. There was no significant difference in fracture resistance between the non-reinforced and reinforced samples (p=0.5481).

6. DISCUSSION

The flasking process emerged as a potential confounding factor in this experiment, as the laboratory conditions during flasking could not be controlled for constant ambient air temperature and humidity. The effects of employing silicone for flasking the acrylic superstructure could also not be determined. Silicone was utilised to allow for easy retrieval and avoiding damage to the acrylic template. In reality the prosthesis would be flasked using plaster in both halves of the flask. Although a new mix of silicone putty was used in the manufacture of each sample, no comparison of acrylic strength in relation to using plaster, could be made.

The effects of the changes in ambient air temperature and humidity may have been a factor affecting the difference in mean fracture resistance values for the reinforced samples. The mean fracture values for the non-reinforced batches were 682 N and 676N for batches one and two, flasked at ambient air temperature and humidity of 25°C/50% and 23°C/65% respectively. The mean fracture values for the reinforced

batches were 782N and 474N for batches one and two, flasked at ambient air temperature and humidity of 27°C/40% and 24°C/82% respectively. The laboratory conditions for this experiment were beyond the author's control but reflect a realistic scenario during the flasking of acrylic prostheses in normal practice. The increased air humidity recorded during the manufacture of the second reinforced batch may have resulted in a higher water content of the acrylic. However, it was not possible to establish whether the differences between the two reinforced batches were due to the differing environmental conditions or to the incorporation of the metal strengtheners and their possible effect on the acrylic curing process.

The standard deviation of fracture values for all the non-reinforced samples was 126N and for all the reinforced, 233N. The variance may also be attributable to inconsistencies and/or un-controlled variables during the flasking process. The greater variance between the reinforced samples may have been due to an influence of the metal strengtheners on the curing process. Clinically the differing variances may be of relevance as the non-reinforced acrylic would offer a more predictable design in terms of consistency in strength.

Only the values for the first (left) cantilever arms were analysed for statistical purposes due to significant differences in the fracture values between the left and right sides of the reinforced samples. It could not be established whether the fracture of the first arm affected the structural integrity of the acrylic at the second arm or whether the difference was due to flasking variables.

The dimensions of the prepared samples were kept uniform and artificial teeth were not embedded in the acrylic bases. This was done to control variables. The uniform 10mmx10mm dimensions of the samples would not be found in clinical cases. Clinically the prosthesis would have varying dimensions depending on the anatomical factors influencing the available restorative space. These factors could affect the strength of a prosthesis used clinically.

This study tested the fracture resistance of the two prosthesis designs to axial linear forces over a small unit of area (2mm tip diameter). The forces were applied with a constant increase in magnitude until the material fractured. The experiment was set up to determine absolute fracture values and did not consider the effects of material fatigue due to repetitive loading below fracture magnitudes, so caution should be exercised in extrapolating these results to an *in-vivo* situation. *In vivo,* the forces are more likely to be applied intermittently, often off-axis, at differing magnitudes, over a larger area and not concentrated on one point. Future research should therefore include subjecting the two different prosthetic designs in this experiment to fatigue testing.

The fracture site for both designs was at the distal most implant abutment. The non-reinforced cantilever arms all fractured off at the first recorded drop in load. The reinforced cantilever arms did not fracture off the remaining prosthesis at the first drop in load but remained attached, and an increase in load resistance was observed after the initial drop. The clinical implications of this are unknown. It is possible that incomplete fracture of the reinforced cantilevers might result in movements and unfavourable force distribution across the distal (and remaining) implant/s. The

complete fracture of the non-reinforced cantilevers may act as a safety feature in the clinical scenario, as this would remove further force and play across the distal implant until the patient can have the prosthesis repaired. This point might be relevant when cantilevers are utilised in immediate loading, such as the “All-on-Four” protocol (Maló et al 2003), where osseointegration depends on the rigid splinting of implants to minimise micro-movements across the implants.

Cantilevers should be minimised to optimise force distribution across the implants and bone interface. Within the limitations of this study, cantilevers up to 15mm, with well distributed implants, can be justified in terms of mechanical prosthesis design.

In vivo occlusal forces are cyclical and range from 200-450N on average during normal function (Worthington, Land and Rubenstein 2003). The mean fracture resistance values of the first (left) cantilever arms of both the non- and the reinforced samples were 679N and 628N respectively and well beyond the 450N level. The lowest values recorded for the first arms were 541N and 424N respectively. From a clinical perspective the absolute lowest fracture resistance values should be considered, and these are at the higher range recorded clinically.

Most importantly, there was no statistical difference between the fracture resistances of the non-reinforced and reinforced samples in this study. Based on this, within the limitations of this study and the obvious increased cost implications of utilising metal strengtheners, the use of metal strengtheners cannot be justified.

7. CONCLUSION AND RECOMMENDATIONS

Within the limitations of this study, metal reinforcement of an acrylic implant-supported prosthesis seems to be unnecessary.

This study has established a standardised design to facilitate further material testing of ISPs. Future testing of acrylic with different reinforcements such as fibre or cast structures housing the acrylic, are recommended. The dimensions of the acrylic frame can be altered and artificial teeth can be included in the framework. Fatigue testing of the designs described in this study and testing of different materials may also be relevant. Combining the absolute fracture resistance with fatigue values could help with the establishment of protocols for optimal material and design for fixed implant supported prostheses.

11. REFERENCES

Balshi T & Wolfinger G (1997). Immediate Loading of Brånemark Implants in Edentulous Mandibles: A Preliminary Report. Implant Dentistry. 6:83-88.

Degidi M, Petrone G, Iezzi G & Piatelli A (2003). Histologic Evaluation of 2 Human Immediately Loaded and 1 Submerged Titanium Implants Inserted in the Posterior Mandible and retrieved After 6 Months. J Oral Impl. VolXXX(5):223-229.

Glauser R, Lungren A, Gottlow J, Sennerby L, Portmann M, Ruhstaller P & Hämmerle C (2003). Immediate Occlusal Loading of Brånemark TiUnite? Implants Placed Predominantly in soft Bone: 1-Year Results of a Prospective Clinical Study. Clin Implant Relat Res. 5(S1):47-56.

Glauser R, Rée A, Lundgren A, Gottlow J, Hämmerle C & Schärer P (2001). Immediate Occlusal loading of Brånemark Implants Applied in Various Jawbone Regions: A Prospective, 1-Year Clinical Study. Clin Implant Relat Res. 3(4):204-213.

Goodacre C, Bernal G, Rungcharassaeng K & Kan J (2003). Clinical complications with implants and implant prostheses. J Prosthet Dent. 90(2):31-41.

Grunder U (2001). Immediate Functional Loading of Immediate Implants in edentulous Arches; Two-Year Results. Int J Perio Rest Dent. 21(6):545-551.

Horiuchi K, Uchida H, Yamamoto K & Sigimura M (2000). Immediate Loading of Brånemark System Implants Following Placement in Edentulous Patients: A Clinical Report. Int J Oral Maxillofac Impl. 15:824-830.

Jaffin R, Kumar A & Berman C (2000). Immediate Loading of Implants in Partially and Fully Edentulous Jaws: A Series of 27 Case Reports. J Periodontol. 71(5):833-838.

Kronstrom M, Widbom T, Lofquist L, Henningson C, Widbom C & Lundberg T (2003). Early functional loading of conical Branemark implants in the edentulous mandible: A 12-month follow-up clinical report. J Prosthet Dent. 89(4):335-340.

Maló P, Rangert B & Nobre M (2003). "All-on-Four" Immediate-Function Concept with Brånemark System® Implants for Completely Edentulous Mandibles: A Retrospective Clinical Study. Clin Implant Relat Res. 5(S1):2-9.

Maniatopoulos C, Pilliar R & Smith D (1986). Threaded versus porous-surfaced designs for implant stabilization in bone-endodontic implant model. J Biomed Mat Res. 20:1309-1333.

Misch C & Degidi M (2003). Five-Year Prospective Study of Immediate/early Loading of Fixed Prostheses in Completely edentulous Jaws with a Bone Quality-Based Implant System. Clin Impl Dent Rel Res. 5(1):17-28.

Nikellis I, Levi A & Nicolopoulos C 2004. Immediate Loading of 190 Endosseous Dental Implants: A Prospective Observational Study of 40 Patient Treatments with up to 2-year Data. Int J Oral Maxillofac Implants. 19:116-123.

Rocci A, Martignoni M & Gottlow J (2003). Immediate Loading in the Maxilla Using Flapless Surgery, Implants Placed in Predetermined positions, and Prefabricated Restorations: A Retrospective 3-Year Clinical Study. Clin Implant Relat Res. 5(S1):29-36.

Romanos G (2004). Present Status of Immediate Loading of Oral Implants. J Oral Impl. Vol XXX(3):189-197.

Salama H, Rose L, Salama M & Betts N (1995). Immediate Loading of Bilaterally Splinted Titanium Root-form Implants in Fixed Prosthodontics – A Technique Reexamined: Two Case Reports. Int J Periodont Rest Dent. 15:345-361.

Schnitman P, Wöhrle P, Rubenstein J, DaSilva J & Wang N (1997). Ten Year Results for Brånemark Implants Immediately Loaded With Fixed Prostheses at Implant Placement. Int J Oral Maxillofac Implants. 12:495-503.

Shackleton JL, Carr L, Slabbert JCG, Becker PJ (1994). Survival of fixed implant-supported prostheses related to cantilever lengths. J Prosthet Dent. 71:23-6.

Skalak R (1983). Biomechanical considerations in osseointegrated prostheses. J Pros Dent. 49:843-848.

Szmukler-Moncler S, Salama H, Reingewirtz Y & Dubruille J (1998). Timing of Loading and Effect of Micromotion on Bone-Dental Implant Interface: Review of Experimental Literature. J Biomed Mater Res. 43:192-203.

Tarnow D, Emtiaz S & Classi A (1997). Immediate Loading of Threaded Implants at Stage 1 Surgery in Edentulous Arches: Ten Consecutive Case Reports With 1-to 5-Year Data. Int J Oral Maxillofac Impl. 12:319-324.

Wolfinger G, Balshi T & Rangert B (2003). Immediate Functional Loading of System Implants in Edentulous Mandibles: Clinical Report of the Results of Developmental and Simplified Protocols. Int J Oral Maxillofac Impl. 18:250-257.

Worthington P, Lang B & Rubenstein J (2003). Osseointegration in Dentistry. An Overview. Second Edition. Quintessence Books Illinois, USA.