Juliana Maria Costa Nuñez Pantoja

Cirurgiã-Dentista

Efeito do design da união soldada a TIG e a laser, em estruturas usinadas em liga de Ti-6Al-4V e fundidas em titânio comercialmente puro, sobre a resistência à ciclagem mecânica

Tese apresentada à Faculdade de Odontologia de Piracicaba da Universidade Estadual de Campinas, para obtenção do título de Doutor em Clínica Odontológica - Área de concentração em Prótese Dental.

Orientador: Prof. Dr. Marcelo Ferraz Mesquita

Piracicaba – SP

2011

FICHA CATALOGRÁFICA ELABORADA PELA

BIBLIOTECA DA FACULDADE DE ODONTOLOGIA DE PIRACICABA

Bibliotecária: Elis Regina Alves dos Santos – CRB-8ª / 8099

N922e

Nuñez-Pantoja, Juliana Maria Costa.

Efeito do design da união soldada a TIG e a laser, em estruturas usinadas em liga de Ti-6Al-4V e fundidas em titânio comercialmente puro, sobre a resistência à ciclagem mecânica / Juliana Maria Costa Nuñez-Pantoja. -- Piracicaba, SP: [s.n.], 2011.

Orientador: Marcelo Ferraz Mesquita.

Tese (Doutorado) – Universidade Estadual de Campinas, Faculdade de Odontologia de Piracicaba.

1. Soldagem em odontologia. 2. Radiografia dentária. 3. Próteses e implantes. 4. Fadiga. 5. Técnica de fundição odontológica. I. Mesquita, Marcelo Ferraz. II. Universidade Estadual de Campinas. Faculdade de Odontologia de Piracicaba. III. Título.

(eras/fop)

Título em Inglês: Effect of TIG and laser welded joints with different designs executed in machined Ti-6Al-4V and casted pure titanium structures on the mechanical cycling resistance

Palavras-chave em Inglês (Keywords): 1. Dental soldering. 2. Radiography, Dental. 3. Prostheses and implants. 4. Fatigue. 5. Dental casting technique

Área de Concentração: Prótese Dental

Titulação: Doutor em Clínica Odontológica

Banca Examinadora: Marcelo Ferraz Mesquita, Luis Geraldo Vaz, Caio Hermann, Mauro Antonio de Arruda Nóbilo, Rafael Leonardo Xediek Consani

Data da Defesa: 22-02-2011

Programa de Pós-Graduação em Clínica Odontológica

UNIVERSIDADE ESTADUAL DE CAMPINAS
Faculdade de Odontologia de Piracicaba

A Comissão Julgadora dos trabalhos de Defesa de Tese de Doutorado, em sessão pública realizada em 22 de Fevereiro de 2011, considerou a candidata JULIANA MARIA COSTA NUÑEZ PANTOJA aprovada.

Prof. Dr. MARCELO FERRAZ MESQUITA

Prof. Dr. LUIZ GERALDO VAZ

Prof. Dr. CAIO HERMANN

Prof. Dr. MAURO ANTONIO DE ARRUDA NÓBILO

Prof. Dr. RAFAEL LEONARDO XEDIEK CONSANI

DEDICO ESTE TRABALHO

A Deus, pela minha vida, pela minha saúde, e pela oportunidade de crescimento intelectual.

Aos meus pais, **Julio** e **Rosana**, ao meu irmão **Mora**, e ao meu marido **Cau**, pelo carinho, exemplo, dedicação, paciência, confiança, sempre me incentivando e estimulando no meu desenvolvimento profissional.

Obrigada por tudo meus queridos. Esse trabalho é para vocês.

AGRADECIMENTOS ESPECIAIS

Ao engenheiro **Julio Sérgio Nuñez Gago** (meu pai), pelo apoio constante, conhecimento e paciência durante o desenvolvimento dos parâmetros essenciais para a execução desse trabalho. Sem você, pai, esse trabalho teria sido muito mais difícil.

Ao meu orientador Prof. Dr. **Marcelo Ferraz Mesquita**, Professor Responsável da Área de Prótese Total da Faculdade de Odontologia de Piracicaba – UNICAMP, pela confiança em mim depositada e pela orientação na elaboração e execução deste trabalho. Sou eternamente grata pelas grandes oportunidades, pelos ensinamentos e pela amizade.

Ao Prof. Dr. **Luis Geraldo Vaz**, pelos conhecimentos, pelos ensinamentos, pela grande solidariedade, pelo desprendimento e pela inteira disponibilidade durante a execução dos ensaios de ciclagem mecânica. Faltam-me palavras de agradecimento, professor.

Ao Prof. Dr. **Mauro Antônio de Arruda Nóbilo**, pela paciência, disponibilidade e prestatividade, na realização das soldagens a laser desse trabalho.

Ao protético **Gelson Toledo**, responsável técnico do Laboratório de Prótese Toledo, pela enorme paciência, competência e disponibilidade, na execução dos procedimentos TIG de soldagem desse estudo. Muito obrigada por tudo, Gelson.

AGRADECIMENTOS

À Faculdade de Odontologia de Piracicaba, nas pessoas do seu diretor Prof. Dr. **Jacks Jorge Júnior**, e do seu vice-diretor Prof. Dr. **Alexandre Augusto Zaia,** pelo acolhimento durante o período de graduação, mestrado e doutorado.

À Profa. Dra. **Renata Cunha Matheus Rodrigues Garcia**, coordenadora geral do programa de pós-graduação da Faculdade de Odontologia de Piracicaba da Universidade Estadual de Campinas.

Ao Prof. Dr. **Marcio de Moraes**, coordenador do programa de pós-graduação em Clínica Odontológica da Faculdade de Odontologia de Piracicaba da Universidade Estadual de Campinas.

À empresa NEODENT, pela doação de todos os corpos de prova usinados, em especial aos colegas, Dr. **Caio Hermann,** e à Dra. **Ivete Sartori**.

À FAPESP, pela concessão da bolsa de estudos (Processo 2009/50567-1) e do auxílio de financiamento do projeto (Processo 2009/51377-1).

À Profa. Dra. **Glaucia Maria Bovi Ambrosano,** pelos ensinamentos, paciência, e auxílio nas análises estatísticas.

Aos amigos do laboratório de Prótese Total, da Faculdade de Odontologia de Piracicaba, **Jessica**, **Ana Paula, Andreza, Sabrina, Claudia, Vanessa**, **Leonardo**, **Mariana**, **Matheus**, **Izabella**, **Ataís, Ana Patrícia, Manoela, Celina, Maíra**, **Bruna**, **Gabriela** e **João Paulo** pela amizade, companheirismo e agradável convivência.

Aos Profs. Drs. **Guilherme Elias Pessanha Henriques** e **Rafael Leonardo Xediek Consani**, pelo convívio agradável durante todo esse período.

Aos demais colegas do Departamento de Prótese e Periodontia, da Faculdade de Odontologia de Piracicaba, pela amizade e companheirismo.

Às minhas avós, **Alina** e **Elvira**, pelas orações, colaborações e risadas. Amo muito vocês.

À minha sogra **Terezinha**, ao meu sogro **José Carlos**, aos meus cunhados **Carlos Henrique** e **Carlos Eduardo,** às minhas cunhadas **Ludmilla** e **Dayane**, ao meu sobrinho **João Pedro**, aos queridos Sr. **Geraldo** (*in memorian*) e Dna. **Nenê**, pelo carinho, paciência na minha ausência, e incentivo sempre.

À **Vanete**, **Tere** e **Célia**, pela ajuda a todo o momento, e pela paciência nos momentos de estresse.

À amiga e futura cunhada **Vivian**, pela amizade, carinho e atenção nesse período.

Ao engenheiro mecânico **Marco Blanco Cangini**, técnico do laboratório de Materiais Dentários, da Faculdade de Odontologia de Piracicaba, pelos conhecimentos e auxílio em diversos momentos.

Ao técnico **Adriano**, pela atenção durante o uso do Microscópio Eletrônico de Varredura.

Aos professores **Cínthia Pereira Machado Tabchoury**, **Jaime Aparecido Cury** e aos técnicos **Waldomiro** e **Alfredo**, pela ajuda na manipulação da saliva artificial utilizada nesse estudo.

Aos técnicos do laboratório de radiologia da Faculdade de Odontologia de Piracicaba, **Fernando**, **Giselda** e **Wal**, pela paciência e pela ajuda na utilização dos equipamentos do laboratório.

Aos funcionários **Fábio** e **Ester** do laboratório de Próteses Odontológicas Vaiazzi, pela atenção e excelente convivência durante a fundição dos corpos de prova da dissertação de mestrado e da tese de doutorado.

Às secretárias **Eliete** e **Mônica**, e aos técnicos **Eduardo** e **Bete** pela ajuda e boa convivência durante o curso de pós-graduação.

A todos os meus amigos que sempre estiveram torcendo por mim.

A todos que indiretamente auxiliaram na elaboração desse trabalho.

Meus sinceros agradecimentos.

EPÍGRAFE

"A mente que se abre a uma nova idéia
jamais voltará ao seu tamanho original"

Albert Einstein

RESUMO

Este estudo avaliou a resistência à ciclagem mecânica de estruturas confeccionadas em liga de Ti-6Al-4V e em Ti cp, submetidas à soldagem a laser (**L**) e ao procedimento TIG de soldagem (**TIG**), com variação de design das uniões, e correlacionar esses resultados com dados obtidos na análise radiográfica dessas soldas. Para o primeiro ensaio, foram obtidos 70 corpos de prova (halteres) usinados em liga de Ti-6Al-4V, com 3,5 mm de diâmetro de secção transversal e, para o segundo ensaio, outros 70 foram fundidos em Ti cp. Os corpos de prova foram seccionados em duas partes iguais e as partes a serem unidas foram limpas e alinhadas, segundo os designs da união: em "I", variando as distâncias de soldagem (0,0mm (**I00**) ou 0,6mm (**I06**)), ou em "X" (**X**). As variáveis design da união e tipo de soldagem foram cruzadas entre si formando um total de 6 grupos, que associadas ao grupo controle (intactos) totalizaram 7 grupos (n=10): **G1-** Intacto; **G2- L/I00**; **G3- L/I06**; **G4- L/X**; **G5- TIG/I00**; **G6- TIG/I06**; **G7- TIG/X,** para cada ensaio. Os corpos de prova foram soldados a laser utilizando-se 360V/8ms (**X**) e 390V/9ms (**I00** e **I06**), com foco e freqüência regulados em zero. As soldas TIG foram realizadas utilizando-se os seguintes parâmetros de soldagem: 2:2 (**X**) e 3:2 (**I00** e **I06**). As uniões receberam acabamento, polimento e foram submetidas à análise radiográfica (90 kV, 15 mA, 0,6 seg e 12 mm de distância), para verificação de poros no interior das soldas. Posteriormente, foi realizado o ensaio de resistência à ciclagem mecânica, e o número de ciclos até a fratura foi registrado. As superfícies de fratura foram analisadas em microscópio eletrônico de varredura. Os testes de Kruskal-Wallis e Dunn (α=0,05) mostraram que para ambos os procedimentos de soldagem, a pior maneira de se realizar a união se dá com **I00**, e a melhor, com **X** ($p<0,05$), para ambos os metais. Apenas para **L**, **I06** foi tão resistente quanto **X,** tanto para o Ti cp quanto para Ti-6Al-4V. Quando são comparados os dois procedimentos de união, para os dois materiais, o teste de Mann-Whitney ($\alpha=0,05$) mostrou haver diferença entre as soldas para **I00** e **I06**, sendo que a união soldada a laser foi a mais resistente. Quando foram comparados os grupos com seus respectivos controles, notou-se que para o Ti-6Al-4V, nenhum grupo soldado foi tão resistente quanto os intactos; já para o Ti cp, os grupos com design **X**, para as duas soldas, foram tão resistentes quanto o controle, assim como o **I06** soldado a laser. O coeficiente de correlação de Spearman ($\alpha=0,05$) indicou correlação negativa entre número de ciclos resistidos e presença de poros nas radiografias das uniões dos materiais testados. Pode-se concluir que a união soldada com design em “X” deve ser empregada quando se pretende soldar a TIG corpos de prova de 3,5 mm de diâmetro de secção transversal. Quando é empregada a solda a laser, pode ser utilizado o design em “I” com 0,6 mm de espaçamento entre as partes, além do design em “X”.

Palavras-Chave: Titânio; Soldagem em Odontologia; Radiografia Odontológica; Próteses e Implantes; Fadiga; Técnicas de Fundição Odontológica.

ABSTRACT

The aim of this study was to evaluate the fatigue strength of Ti-6Al-4V alloy and pure titanium (cp TI) structures, submitted to laser (**L**) and TIG (**TIG**) welding procedures, varying the design of the joints, and correlate these results with joints radiographic analyses data. For the first test, 70 dumbbell rods with central diameters of 3.5 mm were obtained by lost-wax casting procedure in cp Ti, and others 70 were machined in Ti-6Al-4V alloy, for the second test. The specimens were sectioned in 2 equal parts, perpendicular to the rods long axis. The parts to be weld were cleaned and aligned according to joint designs: "I" design, varying welding distances (0.0 mm (**I00**) or 0.6 mm (**I06**)), or "X" (**X**) design. The variables joint design and type of welding were crossed creating 6 groups, that associated to the control group (intact), totaled 7 groups (n=10): **G1-** Intact; **G2- L/I00**; **G3- L/I06**; **G4- L/X**; **G5- TIG/I00**; **G6- TIG/I06**; **G7- TIG/X,** to each test. Laser-welding was executed using 360V/8ms (**X**) and 390V/9ms (**I00** e **I06**) with focus and frequency regulated at zero. The TIG welding were executed using 2:2 (**X**) and 3:2 (**I00** and **I06**) as welding parameters. The joints were finished, polished and submitted to radiographic examination (90 kV, 15 mA, 0.6 second and 12 mm of distance), for pores verification within the joints. Later, the rods were submitted to mechanical cyclic tests, and the number of cycles until fracture was recorded. The fracture surface was examined with a scanning electron microscope. Kruskal-Wallis and Dunn test (α=0.05) indicated that for both welding procedures, the worst way to perform the joint occurs with **I00**, and the best, with **X** ($p<0.05$), for both metals. To **L**, **I06** was as resistant as **X**. When both type of welding are compared, to both materials, the Mann-Whitney test (α=0.05) indicates difference to **I00** and **I06**, being the laser-weld joint the better one. When groups were compared with their respective control groups, it was noticed for Ti-6Al-4V that none of the joint groups was as resistant as the intact; on the contrary to cp Ti the “X” design groups, welded by **TIG** or **L** were as resistant as control, even as **I06** laser-welded. The Spearman correlation coefficient (α=0.05) indicated a negative correlation between number of cycles and presence of porosity in radiographies of both materials. It could be concluded that the “X” design can be employed to weld 3.5 mm in diameter structures by laser or TIG welding procedure. When laser welding procedure is used, the “I” design with a space of 0.6 mm between parts can also be employed.

Key Words: Titanium; Dental Soldering; Dental Radiography; Prostheses and Implants; Fatigue; Dental Casting Technique.

SUMÁRIO

PÁGINA

1. INTRODUÇÃO 1

2. CAPÍTULOS 7

CAPÍTULO 1: Fatigue strength: Effect of welding type and joint design executed in Ti-6Al-4V structures ……... 8

CAPÍTULO 2: Fatigue performance of several joint designs executed in pure titanium structures varying the type of welding ………………... 30

3. CONSIDERAÇÕES GERAIS.. 48

4. CONCLUSÃO.. 54

REFERÊNCIAS 55

APÊNDICE 58

1. INTRODUÇÃO

Devido ao alto índice de sucesso das reabilitações protéticas implanto-suportadas, cada vez é mais freqüente a utilização de implantes para a confecção de próteses unitárias ou múltiplas. Esse tipo de reabilitação relaciona-se à preservação de estruturas dentais adjacentes sadias, em próteses parciais, e à melhora da retenção e da estabilidade das próteses totais (Neo *et al.*, 1996).

Inicialmente, as infraestruturas protéticas implanto-suportadas eram confeccionadas a partir de metais nobres, mas devido ao alto custo dos mesmos passaram a ser substituídas por metais não nobres. Estes, entretanto, apresentavam algumas desvantagens, tais como baixa resistência à corrosão, insuficiente adaptação cervical e possíveis reações alérgicas a elementos como níquel e berílio. Iniciaram-se, então, estudos a respeito do uso do titânio, e de suas ligas, também na construção das infraestruturas. Esse elemento trazia consigo diversas vantagens, tais como: biocompatibilidade, alta resistência à corrosão, baixo custo, propriedades físicas e mecânicas favoráveis, baixa densidade e baixa condutibilidade térmica. (Taira *et al.*, 1989; Craig *et al.*, 1993; Lautenschlager & Monaghan, 1993; Wang & Fenton, 1996; Chai & Chou, 1998).

A maior dificuldade na utilização do titânio deve-se a alta reatividade química com o oxigênio, hidrogênio e nitrogênio do ar, quando aquecido a temperaturas acima de 600 ^{o}C. O alto ponto de fusão desse metal ($\pm$1700 ºC), também torna difícil o processo de fundição e soldagem. Dessa forma, faz-se necessário o uso de equipamento especial que mantenha uma atmosfera inerte de gás argônio impedindo a contaminação do metal (Taira *et al.* 1989; Craig *et al.*, 1993).

Para que se obtenha sucesso em um tratamento utilizando próteses confeccionadas sobre implantes é necessário que seja feito um planejamento adequado e cauteloso, que sejam tomados os devidos cuidados durante os procedimentos clínicos e laboratoriais, que seja confeccionada uma infraestrutura metálica resistente aos esforços mastigatórios e a corrosão, e que se obtenha a melhor adaptação possível dos pilares, com passividade (Sjögren *et al.*, 1988). Próteses sobre implantes requerem maior precisão na adaptação e assentamento passivo quando comparadas às sobre dentes, devido à ausência de ligamento periodontal entre o implante e o osso. Uma desadaptação da prótese pode acarretar danos na interface osseointegrada (Sahin & Çehreli, 2001; Chai & Chou, 1998).

Os procedimentos clínicos e laboratoriais, mesmo quando realizados de maneira criteriosa, invariavelmente geram distorções na peça finalizada (Wee *et al.*, 1999). Dessa forma, é consenso na literatura mundial, que o desajuste marginal é uma realidade clínica (Spazzin *et al*., 2009). Assim, para contornar desvantagens inerentes à obtenção de um monobloco, várias estratégias têm sido propostas, visando ao aumento da precisão ou à diminuição de erros inerentes ao processo de confecção da prótese. O corte em segmentos da estrutura fundida e reunião pela técnica da soldagem é uma delas.

Quando realizada de maneira convencional (soldagem por brasagem), a técnica aquece demasiadamente a peça protética induzindo a alterações dimensionais e a modificações de propriedades, incorporando falhas e heterogeneidades (Henriques *et. al.*, 1997). Para o titânio e suas ligas, muita ênfase tem sido dada aos processos TIG (*tungsten inert gas*) e laser (*light amplification by stimulated emission of radiation*) de soldagem, pois produzem soldas de qualidade superior à obtida por brasagem a maçarico, além de gerarem pouca distorção na peça e evitarem a contaminação desse metal com elementos do ar, devido à proteção local de gás inerte, geralmente um fluxo de argônio (Bertrand & Polon-Quintin, 2010; Nuñez-Pantoja *et al.*, 2010).

A soldagem a laser e a TIG são consideradas métodos práticos, pois não necessitam da inclusão da peça em revestimento, sendo realizadas no próprio modelo (Gordon & Smith, 1970; Souza *et al.*, 2000; Bertrand & Polon-Quintin, 2010; Nuñez-Pantoja *et al.*, 2010). Além disso, acarretam em menor índice de distorções, principalmente no caso da solda a laser, já que o feixe de laser pode ser concentrado em um ponto muito pequeno, evitando o aquecimento da peça toda (Tambasco *et al.*, 1996; Wang & Chang, 1998; Bertrand & Polon-Quintin, 2010). O procedimento TIG de soldagem utiliza como fonte térmica um arco elétrico formado entre o eletrodo de tungstênio e a peça a ser soldada. É um procedimento que está sendo muito utilizado na prática clínica atual, devido ao menor custo do aparelho de soldagem quando comparados às máquinas de soldagem a laser (Rocha *et al*., 2006; Bock *et al.*, 2008).

Características intrínsecas do processo de soldagem podem alterar as propriedades mecânicas do metal na região soldada (Berg *et al.,* 1995; Liu *et al*., 2002; Zavanelli *et al.,* 2004). Dessa maneira, os protocolos de execução das técnicas de soldagem e as características das uniões soldadas despertam o interesse dos pesquisadores, que buscam avaliar as conseqüências dos procedimentos sobre a qualidade da peça finalizada, e o que isso pode representar na longevidade do tratamento protético (Gordon & Smith, 1970; Huling & Clark, 1977; Zavanelli *et al.*, 2004; Nuñez-Pantoja *et al.*, 2010).

Apesar da melhor adaptação aos pilares, a infraestrutura soldada pode não apresentar a mesma resistência mecânica quando comparada a infraestrutura fundida em monobloco, quando submetidas a esforços cíclicos presentes durante o ato da mastigação. Esses esforços estão diretamente ligados ao fenômeno de fadiga, que causa uma alteração na estrutura do material, sendo permanente, localizada e progressiva, podendo ou não levar a fratura do componente depois de um determinado número de ciclos. (Vallittu & Luotio, 1996; Henriques *et. al.*, 1997; Ferreira *et. al.*, 1998, Nuñez-Pantoja *et al*., 2010). Souza (1974) e Dieter (1981) relataram que 90% das falhas mecânicas de infraestruturas metálicas ocorrem devido a esse fenômeno, que está intimamente ligado às características superficiais das estruturas solicitadas. A presença de descontinuidades geométricas internas e irregularidades superficiais tendem a agir como nucleadores de iniciação de trincas (Henriques *et. al.,* 1997).

Ainda não estão totalmente estabelecidos, na literatura, protocolos de soldagem a laser e a TIG, em função do diâmetro a ser soldado. Sabe-se, contudo, que quanto maior é esse diâmetro, menor é a capacidade de penetração do feixe de laser (Baba & Watanabe, 2005; Bertrand & Polon-Quintin, 2010; Nuñez-Pantoja *et al*., 2010). Quando se aumenta a voltagem do aparelho de soldagem a laser, aumenta-se a profundidade de penetração da solda. Entretanto, quando a energia se torna excedente, o metal ao redor da área irradiada se funde e parte dele se evapora, formando-se uma região denominada buraco de fechadura, que diminui a resistência da união soldada (Baba & Watanabe, 2005; Bertrand & Polon-Quintin, 2010; Nuñez-Pantoja *et al.*, 2010). Sabe-se que estruturas de diâmetro reduzido devem ser soldadas, sempre que possível, com justaposição das partes, para que seja evitada a formação de poros ocorrida por inclusão de gás argônio no momento da soldagem, quando há espaçamento entre as partes (Roggensach *et al.*, 1993; Zavanelli *et al.*, 2004; Nuñez-Pantoja *et al.*, 2010); todavia, para estruturas de diâmetros superiores a 3,0 mm, a justaposição das partes dificulta a penetração da solda na porção central da secção transversal (Berg *et al*., 1995; Bertrand & Polon-Quintin, 2010; Nuñez-Pantoja *et al.*, 2010).

Na tentativa de superar esse problema, diferentes designs da união foram propostos durante o passar dos anos (Zupancic *et al*., 2006; Bertrand & Polon-Quintin, 2010; Nuñez-Pantoja *et al.*, 2010). Quando duas partes a serem unidas formam a letra "X" vistas lateralmente, elas podem ser soldadas iniciando-se na porção central, seguindo rumo à superfície, com o uso de menor energia, quando comparadas à justaposição de duas partes formadoras da letra 'I", visto que uma profundidade reduzida de penetração do laser seria requerida (Zupancic *et al*., 2006; Bertrand & Polon-Quintin, 2010; Nuñez-Pantoja *et al.*, 2010).

Existe ainda grande necessidade de novas pesquisas a cerca da soldagem do titânio e de suas ligas, principalmente no que diz respeito à resistência da união soldada. Desta forma, o objetivo neste estudo foi avaliar o efeito das soldagens TIG e a laser em estruturas usinadas em liga de Ti-6Al-4V, e em estruturas fundidas em Ti cp, com 3,5mm de diâmetro de secção transversal variando os designs das uniões; analisar radiograficamente a qualidade da solda e observar o aspecto da superfície de fratura, correlacionando os achados com os ensaios de resistência à fadiga.

2. CAPÍTULOS

Esta tese de doutorado está baseada na Resolução CCPG/002/06UNICAMP que regulamenta o formato alternativo para dissertações de Mestrado e teses de Doutorado. Dois capítulos contendo artigos científicos compõem este tese de Doutorado, conforme descrito abaixo:

Capítulo 1. Fatigue strength: Effect of welding type and joint design executed in Ti-6Al-4V structures

Artigo nas normas do periódico **Journal of Oral Rehabilitation**.

Capítulo 2. Fatigue performance of several joint designs executed in pure titanium structures varying the type of welding

Artigo nas normas do periódico **Journal Of Material Science. Materials in Medicine**

CAPÍTULO 1

Fatigue strength: Effect of welding type and joint design executed in Ti-6Al-4V structures

Running Title: Fatigue strength of Ti-6Al-4V joints

Key words: Titanium Alloys; Fatigue Performance; Laser-weld Joint; TIG Welding; Dental Implant prostheses.

Authors:

Juliana Maria Costa Nuñez Pantoja [a]

Luis Geraldo Vaz [b]

Rafael Leonardo Xediek Consani [c]

Mauro Antônio de Arruda Nóbilo [d]

Marcelo Ferraz Mesquita [d]

[a] Post Graduate Student, Department of Prosthodontics and Periodontology, Piracicaba Dental School, University of Campinas, Piracicaba, São Paulo, Brazil.
[b] Adjunct Professor, Department of Dentistry Materials and Prosthodontics, Araraquara Dental School, State University of São Paulo Julio de Mesquita Filho, Araraquara, São Paulo, Brazil.
[c] Associate Professor, Department of Prosthodontics and Periodontology, Piracicaba Dental School, University of Campinas, Piracicaba, Sao Paulo, Brazil.
[d] Chair Professor, Department of Prosthodontics and Periodontology, Piracicaba Dental School, University of Campinas, Piracicaba, São Paulo, Brazil.

This study was supported by FAPESP (Grant nº 09/50567-1), and NEODENT Indústria e Comércio de Materiais Dentários – Ltda.

Corresponding author:
Juliana Maria Costa Nuñez Pantoja
Travessa Dona Eugênia, 135 ap. 71
Piracicaba, Sao Paulo, Brazil, 13416-218
Telephone number: +55 19 8203 8158 / Fax number: +55 19 2106 5218
E-mail: julynunez@terra.com.br

Date of submission:

Abstract

This study evaluated the fatigue strength of Ti-6Al-4V machined structures submitted to laser (**L**) and TIG (**TIG**) welding procedures, varying the design of joints. Seventy dumbbell rods were machined in Ti-6Al-4V alloy with central diameters of 3.5 mm. The specimens were sectioned and welded, using **TIG** or **L**, and 3 joint designs ("I" design, varying welding distances (0.0 mm (**I00**) or 0.6 mm (**I06**)), or "X" (**X**) design). The combinations of variables created 6 groups, which, when added to the intact group, made a total of 7 groups (n=10). **L** was executed as follows: 360 V/8 ms (**X**) and 390 V/9 ms (**I00** and **I06**) with focus and frequency regulated to zero. **TIG** was executed using 2:2 (**X**) and 3:2 (**I00** and **I06**) as welding parameters. Joints were finished, polished, and submitted to radiographic examination to be analyzed visually for presence of porosity. The specimens were then subjected to mechanical cyclic tests, and the number of cycles until failure was recorded. The fracture surface was examined with a scanning electron microscope. The Kruskal-Wallis and Dunn test (α=0.05) indicated the number of cycles resisted for fracture was higher to **X** for both welding procedures. To **L**, **I06** was as resistant as **X**. The Mann-Whitney test (α=0.05) indicated that **L** joints were more resistant than **TIG** to **I00** and **I06**. Spearman correlation coefficient (α=0.05) indicated a negative correlation between number of cycles and presence of porosity. Thus, to weld Ti-6Al-4V structures, the best condition is **X**, independent of welding type.

Introduction

For many years, conventional removable dentures were the only treatment available for non-toothed elderly people although these people, in most cases, suffered due to instability of mandibular dentures (1,2). With the implantology approach, an alternative for these people treatment appeared (2,3).

The implant prostheses frameworks are made, many times, from titanium because of its biocompatibility, excellent fatigue and corrosion resistance, low density, low thermal conductibility, and low relativity cost, when compared with gold dental alloys (2,4-6). Several titanium alloys are used for this reason. The Ti-6Al-4V alloy is the most common alloy used due to its resistance and high performance (5-7).

Implant prostheses require more accurate fabrication criteria than conventional ones due to the lack of periodontal tissues. Marginal misfit of implant-supported frameworks may cause biologic complications of the surrounding tissues or mechanical failure of the prostheses (8). However, potential distortions can be created at any step of the framework- fabrication process (9,10). These distortions can be corrected by employing welding procedures (6,9,11-13).

Titanium alloys are characteristically difficult to cast and weld because of the high melting point and strong affinity to gases such as oxygen, hydrogen, and nitrogen (12,14,15). For this reason, it is necessary to use a special machine that employs argon shielding in the welding chamber to solder titanium frameworks (12,15,16). The laser-welding technique and TIG welding procedure are the most-used welding processes when it comes to titanium (6,2,15-17).

These processes create reduced joining distortions, lower damage to veneering coverage, and close yield strength to the base metal under static conditions (11,12,15,18). However, some reports on laser welding of titanium and titanium alloys have shown the presence of large pores in the laser-repaired joints (6,13).

Despite the better fit, welded frameworks must not present the same fatigue strength as compared to intact structures (6,13). It is usual to find voids, inclusions, and notches inside the joints that act as stress raisers in the fatigue process, which occurs when these joints are submitted to cyclic loadings (6,15).

Welding protocols are not totally established in scientific literature, although, it is known that the joint's design directly affects its mechanical resistance (6,15). In structures with diameters larger than 3.0 mm, the laser beam cannot penetrate to the centre of cross section if there is an "I" design with juxtaposition of the parts (6,15,18,19). In thin diameters, the resistance is lower when there is space between the joint parts, when compared with juxtaposition of parts, due to gas arrestment during the addition of metal used to fill the gap (6,19). Thus, aiming to establish the best joint design for structures 3.5 mm in diameter, depending on the type of welding used, this study evaluated the fatigue strength of Ti-6Al-4V welded structures, using several joint designs and two types of welding.

Material and Methods

Seventy (70) dumbbell-shaped rods, with 3.5 mm in the central segment, were machined, based on norm ASTM E8M-04 (20). Forty (40) of these rods were sectioned in half, perpendicularly to the long axis, and 20 of them were sectioned and bevelled to form an "X" (chamfer of 1mm/30^o) when the lined-up of parts were seen by side-view (Fig. 1a). The metal adjacent to the gap was blasted with abrasive particles of aluminium oxide of 100 μm at a pressure of 0.55 MPa (13). The parts were lined up in a metal matrix in order to maintain distances of 0.0 and 0.6 mm between them (Figs. 1b and 1c) for the "I" design and just with juxtaposition of parts to "X" design (Fig. 1a). For the laser welding procedure (**L**), the welding was carried out using energy of 360 V/8 ms (**X**) and 390 V/9 ms (**I00** and **I06**), with the focus and frequency calibrated at zero, with a Desktop–F (Dentaurum, Pforzhein, Germany) laser machine. For the TIG procedure (**TIG**), 3:2 (**I00** and **I06**) and 2:2 (**X**) were used as welding parameters with the ratio numerator representing the power used, and the denominator using the programmed time. These procedures were conducted in the NTY 60C machine (Kernit, Indústria Mecatrônica Ltda., Indaiatuba, Brazil). The joints were welded on opposite sides of the cross section to stabilize parts of the specimen aligned, and then the welding was completed by a trained and competent professional. The "X" designs were welded, starting the procedure in the centre of the specimens to mark a full contact in the centre of each half piece; the welding was then completed using commercially pure titanium grade II metal filler (Dentaurum, Pforzhein, Germany) to provide a total joint between the two parts.

The combination between the variables (welding design and type of welding) created a total of 6 groups (n=10) which, together with the control group (intact), made a total of 7 groups: **G_1- L/X**, **G_2- L/I06**, **G_3- L/I00**, **G_4- TIG/X**, **G_5- TIG/I06**, **G_6- TIG/I00**, **G_7**- Intact.

After the welding processes, the joints were finished and polished with a rubber nº 5001 polisher (Dedeco Dental, New York, USA) and titanium polishing paste (Tiger Brilliant Polier Paste, Dentaurum, Pforzheim, Alemanha). During the finishing process, the specimens' diameters were constantly verified with electronic caliper (Starret Tools Co. Ltd., Shanghai, China), assuring accuracy of within 0.01 mm (6).

Radiographs of the joints were taken with radiographic film (Ektaspeed Plus, Eastman Kodak, Rochester, NY) to verify internal defects. The radiographic exam consisted of the exposure of the specimens to radiation (90 kV, 15 mA, 0.6 seconds and 12 mm of distance), using a periapical film (2, 6, 13). The radiographies (Fig. 2) were visually analyzed for the presence of porosity in the joints.

Before the fatigue test, 3 extra randomly-intact specimens were submitted to tensile tests to establish the mean yield strength at 0.2% permanent strain. The stress established for the fatigue test (291 MPa) was calculated as 30% of the mean yield strength (970 MPa) (6). The loading frequency in the fatigue test was 15 Hz (6), and the loading used was 2800 N. The specimens were then submitted to fatigue strength tests using a testing machine (Test Star II, Material Testing System, MTS Systems Corp, Minneapolis, Minn.) and the number of cycles required to cause the fatigue fracture were registered. The specimens were tested submerged in synthetic saliva (1.5 mM Ca, 3.0 mM P, 20.0 mM $NaHCO_3$, pH 7.0) (21) at room temperature to 100 000 cycles (6).

Fatigue fracture surfaces of representative specimens were examined using a scanning electron microscope (Electron Probe Microanalyser, Jeol model JXA 840 A, Jeol Ltd, Tokyo, Japan).

An exploratory analysis of the number of cycles was done, and the nonparametric test was requested. Kruskal-Wallis and Dunn tests (α=0.05) were applied to compare designs (3 to 3) inside the level of type of welding. The Mann-Whitney test (α=0.05) was applied to compare type of welding (2 to 2) inside the level of design, and to also compare (2 to 2), the control group with all of the groups. The data from the radiographic analyzes were correlated with the number of cycles by the Spearman correlation coefficient (α=0.05) to each type of welding/ joint design, and to all grouped data.

Results

Table I shows the median values of the number of cycles until fracture in the experimental groups. The number of cycles required until fracture was seen to be higher to **X** for both welding procedure and **I06** for the laser welding ($p<0.05$). The lower number of cycles data were found to **I00** ($p<0.05$). When both types of welding were compared, the Mann-Whitney test (α=0.05) indicates difference to **I00** (p=0.0032) and **I06** (p=0.0002), with the laser weld joints being the most resistant of them ($p<0.05$). All the specimens included in the control groups achieved over 100 000 cycles, and this group was the most resistant of all the groups.

The Spearman correlation coefficient (α=0.05) indicated there was a negative correlation (p=0.0141/rs= -0.3152) between the number of cycles and the presence of porosity (Table II). In these cases, specimens that had pores in the joint, as observed by radiographies (Fig 2), were less resistant to mechanical cycling.

Figs. 3 and 4 show SEM photomicrography of the fractured surfaces of laser-weld joints and TIG-weld joints, respectively. Figs. 3a and 4a show the surface of X designs. Figs. 3b and 4b show the 0.6 mm-joint opening with "I" design. Figs. 3c and 4c show the 0.0 mm-joint opening with "I" design. The presence of internal pores or voids caused by insufficient weld penetration (arrows) was higher when the joint opening was 0.0 mm for 'I" design specimens, for both types of welding. For **I06**, the TIG procedure showed less welding penetration than laser welding, and the internal void was smaller in **X**.

Discussion

To achieve passive fit of the prostheses, it is often necessary to cut the framework and repair it later with a welding procedure (8). However, a welded prosthesis often does not present the same fatigue strength as the non-welded (6,13).

Fatigue, which is the process of progressive localized permanent structural change occurring in a material subjected to cyclic loading, is generally responsible for 90% of all material service failures (7). This process is initiated by stress-raisers such as voids, inclusions, notches, surface roughness modifiers, and metallurgical variables. Structural defects within the joints assume greater significance under cyclic loading than under static tensile loading (22).

Porosities and voids can occur as a result of gas arrestment, insufficient deep welding penetration, or by the capture of inclusions such as acrylic resin used to join in the mouth the parts to be welded (15). Gas-arrestment formation is related to rapid solidification process, to energy excess released during the welding process, or just to vaporization of volatile alloying elements (6,15).

According to some authors, full contact of base metals is associated with successful laser joints (13,18), although this is not an absolute truth, since the degree of weld penetration depth is dependent on the diameter of the region to be welded. Some studies show that the laser beam penetration is limited to 1.5 mm in depth (3,18,19,23). In a previous study, the weld penetration depth was insufficient to 3.5 mm specimens in diameter (6), which agrees with the results of this study (Table I). Figs. 2b, 3c and 4c showed the insufficient welding penetration in the centre of the specimens.

When more than 2 mm in depth is required, an "X" shape preparation should be used, and welding always requires the use of a metal filler, essential to optimize laser welding, minimize tensions occurring in the work piece, and fulfill the joint (6,15). In this study, this was true for the two welding procedures studied (Table I / Figs. 3a and 4a); however, greater emphasis should be given to the TIG welding procedure due to the fact that ,for laser welding, the **I06** was a favorable design (Table I / Fig. 3b).

The TIG welding is a punctual welding procedure, since the energy supplied to the joint is given through the contact of the tungsten electrode with the metal part to be welded, which generates an electric arc (16). Thus, the welding penetration deep inside the cross section of thick specimens is hampered, even when there is a gap between parts, due to the fact that the electrode does not reach the central portion of the cross section, especially in delicate gaps (0.6 mm). On the contrary, the laser beam can reach this region and melt the metal filler interposed between the two parts. The laser beam can be considered a monochromatic and directional light beam, high energy, able to be focused on very small areas (15,18). This may explain the higher results to laser weld joints in **I06** (Table I), when compared to TIG welding joints ($p=0.0002$).

In a previous study (6) that used similar methods, the "I" design with a gap between parts, did not create dumbbells as tough as those obtained in this study. Probably a methodology improvement was responsible for these differences. This study used a metallic device for alignment of the parts to be welded, which avoided the union of the parts with acrylic resin. In this way, it was possible to perform the cleaning and oxide blasting of the parts with less probability of causing inclusions originating from microscopic remnants of acrylic resin, solid vaseline, or oxide particles.

It is important to pay attention to the fact that none of the treated groups had equivalent results with the control groups, which shows that Ti-6Al-4V welded specimens of 3.5 mm in diameter are less resistant to mechanical cycling than are intact specimens (Table I). This is probably due to the fact that the intact specimens were free from any internal or external irregularities, being very much resistant specimens. They were machined from pre-fabricated bars of Ti-6Al-4V grade V, the same that are used in prosthetic component and in CAD-CAM frameworks. This, along with the reduced weldability of Ti-6AL-4V, can explain these results (24,25). Titanium weldability decreases with increased levels of β stabilizers (24). Because vanadium is a β stabilizer, the α-β alloy, Ti-6Al-4V, presents weldability inferior to that of cp Ti (24,25).

Nuñez-Pantoja and collaborators (2010) (6) obtained results regarding some welded specimens similar to those of intact specimens. These different results may be due to the difference in dumbbell cross section diameters or due to the different lots used to machine the specimens in both studies. The difference in lots can lead to different mean yield strengths.

The radiographic analyses of the joints demonstrated a negative correlation, as revealed by the Spearman correlation coefficient (rs=-0.3152/p=0.0141), when the groups were analyzed together. It was observed that when there was porosity inside the joints in radiographic analyses (Fig. 2), the number of cycles decreased, which corroborates with Nuñez-Pantoja *et al.* (2010) (6). This correlation confirms that internal porosity really acts as an stress raiser (6,13,19,22), decreasing the fatigue life of the prostheses.

An important detail to be discussed is that this study did not aim to quantify welded framework survival. This study aimed to evaluate the best design to be used to weld 3.5 mm in diameter, depending on the type of welding used. This trial of materials resistance aimed to compare the welded joints. As such, no conclusions can be drawn with regard to the number of years that welded frameworks will survive in the buccal cavity. It is suggested, in this manner, for future studies, the mechanical cycling of welded joints performed in prosthetic frameworks, using a simulated masticatory load, to verify the longevity of these joints in simulated clinical situations.

Within the limitations of this study, it may be concluded that the use of the “X” design is necessary when it is intended to TIG weld structures 3.5 mm in diameter. To laser weld these cross sections, the “X” design or the “I” design can be used, without juxtaposition of the parts. Moreover, radiographs of welded joints can be considered a way to perform quality control of TIG and laser welded joints in clinical practice.

Acknowledgements

This study was supported by FAPESP (Grant nº 09/50567-1), and NEODENT Indústria e Comércio de Materiais Dentários – Ltda.

References

1. Assunção WG, Zardo GG, Delben JA, Barão VA. Comparing the efficacy of mandibular implant-retained overdentures and conventional dentures among elderly edentulous patients: satisfaction and quality of life. Gerodontology. 2007;24(4):235-8.
2. Nuñez JM, Takahashi JM, Henriques GE, Nóbilo MA, Consani RL, Mesquita MF. Radiographic inspection of porosity in pure titanium dumbbell castings. Gerodontology. 2010 May 26. doi: 10.1111/j.1741-2358.2010.00376.x. [Epub ahead of print]
3. Neo TK, Chai J, Gilbert JL, Wozniak WT, Engelman MJ. Mechanical properties of titanium connectors. Int J Prosthodont. 1996;9(4):379-93.
4. Blackman R, Barghi N, Tran C. Dimensional changes in casting titanium removable partial denture frameworks. J Prosthet Dent. 1991;65(2):309-15.
5. Wang RR, Fenton A. Titanium for prosthodontic applications: a review of the literature. Quintessence Int. 1996;27(6):401-8.
6. Nuñez-Pantoja JM, Vaz LG, Nóbilo MA, Henriques GE, Mesquita MF. Effects of laser-weld joint opening size on fatigue strength of Ti-6Al-4V structures with several diameters. J Oral Rehabil. 2010 Jul 29. [Epub ahead of print]
7. Zavanelli RA, Pessanha Henriques GE, Ferreira I, De Almeida Rollo JM. Corrosion-fatigue life of commercially pure titanium and Ti-6Al-4V alloys in different storage environments. J Prosthet Dent. 2000;84(3):274-9.
8. Spazzin AO, Henriques GE, de Arruda Nóbilo MA, Consani RL, Correr-Sobrinho L, Mesquita MF. Influence of prosthetic screw material on joint stability in passive and non-passive implant-supported dentures. Open Dent J. 2009 Dec 30;3:245-9.
9. Wee AG, Aquilino SA, Schneider RL. Strategies to achieve fit in implant prosthodontics: a review of the literature. Int J Prosthodont. 1999;12(2):167-78.
10. Lee H, Ercoli C, Funkenbusch PD, Feng C. Effect of subgingival depth of implant placement on the dimensional accuracy of the implant impression: an in vitro study. J Prosthet Dent. 2008;99(2):107-13.
11. Chai T, Chou CK. Mechanical properties of laser-welded cast titanium joints under different conditions. J Prosthet Dent. 1998;79(4):477-83.

12. Watanabe I, Topham DS. Laser welding of cast titanium and dental alloys using argon shielding. J Prosthodont. 2006;15(2):102-7.
13. Zavanelli RA, Guilherme AS, Pessanha-Henriques GE, de Arruda Nóbilo MA, Mesquita MF. Corrosion-fatigue of laser-repaired commercially pure titanium and Ti-6Al-4V alloy under different test environments. J Oral Rehabil. 2004;31(10):1029-34.
14. Wang RR, Welsch GE. Joining titanium materials with tungsten inert gas welding, laser welding, and infrared brazing. J Prosthet Dent. 1995;74(5):521-30.
15. Bertrand C, Poulon-Quintin A. Proposals for optimization of laser welding in prosthetic dentistry. J Prosthodont. 2010;19(1):69-76
16. Rocha R, Pinheiro AL, Villaverde AB. Flexural strength of pure Ti, Ni-Cr and Co-Cr alloys submitted to Nd:YAG laser or TIG welding. Braz Dent J. 2006;17(1):20-3.
17. Liu J, Watanabe I, Yoshida K, Atsuta M. Joint strength of laser-welded titanium. Dent Mater. 2002;18(2):143-8.
18. Tambasco J, Anthony T, Sandven O. Laser welding in the dental laboratory: an alternative to soldering. J Dent Technol. 1996; 13(4):23-31.
19. Roggensack M, Walter MH, Böning KW. Studies on laser- and plasma-welded titanium. Dent Mater. 1993;9(2):104-7.
20. ASTM International. Designation: E 8M – 04: Standard Test Methods for Tension Testing of Metallic Materials [Metric]. West Conshohocken; 2004. 24p.
21. Birkeland JM. The effect of pH on the interaction of fluoride and salivary ions. Caries Res. 1973;7(1):11-8.
22. Henriques GE, Consani S, Rollo JM, Andrade e Silva F. Soldering and remelting influence on fatigue strength of cobalt-chromium alloys. J Prosthet Dent. 1997;78(2):146-52.
23. Baba N, Watanabe I. Penetration depth into dental casting alloys by Nd:YAG laser. J Biomed Mater Res B Appl Biomater. 2005;72(1):64-8.
24. Wiskott HW, Doumas MT, Scherrer SS, Susz C, Belser UC. Microstructures of brazings and welds using grade 2 commercially pure titanium. Int J Prosthodont. 2001;14(1):40-7.

25. Lampman S. Wrought titanium and titanium alloys. In: Metal handbook. 10 ed. Metals Park: ASM; 1990. V.2, p. 592 – 633.

Table 1. Median (standard deviation) values of number of cycles resisted in the experimental groups.

Designs	Type of welding	
	Laser	TIG
"X"	* 32 376.5 (8 009.8) Aa	*40 573.3 (10 512.35) Aa
"I"/distance 0.6 mm	* 31 422 (8 121.557) Aa	*8 866 (2 439.08) Bb
"I"/distance 0.0 mm	*2 720 (1 173.58) Ba	*1 082.5 (635.08) Cb

Intact group = 100 000 (0,0). * Differs from intact group (Mann-Whitney, alfa=0.05).

Median values followed by the same letter are not significantly different (Kruskal-Wallis / Dunn, alfa=0.05). Capital letters compare designs inside the level of the factor type of welding and small letters compare type of welding inside the level of design factor.

Table 2. Association of number of cycles survived and presence of internal voids in radiography exam.

		Spearman Coefficient	(p)=
Laser	"X"	-0.7817	0.0075
	"I" distance 0.6 mm	-0.6963	0.0252
	"I" distance 0.0 mm	-0.6963	0.0252
TIG	"X"	-0.6963	0.0252
	"I" distance 0.6 mm	*	*
	"I" distance 0.0 mm	-0.6963	0.0252
All type of welding / All designs		-0.3152	0.0141

* All specimens' present pores in radiographies, thus correlation was impossible.

Spearman Coefficient ranges between -1 and 1.

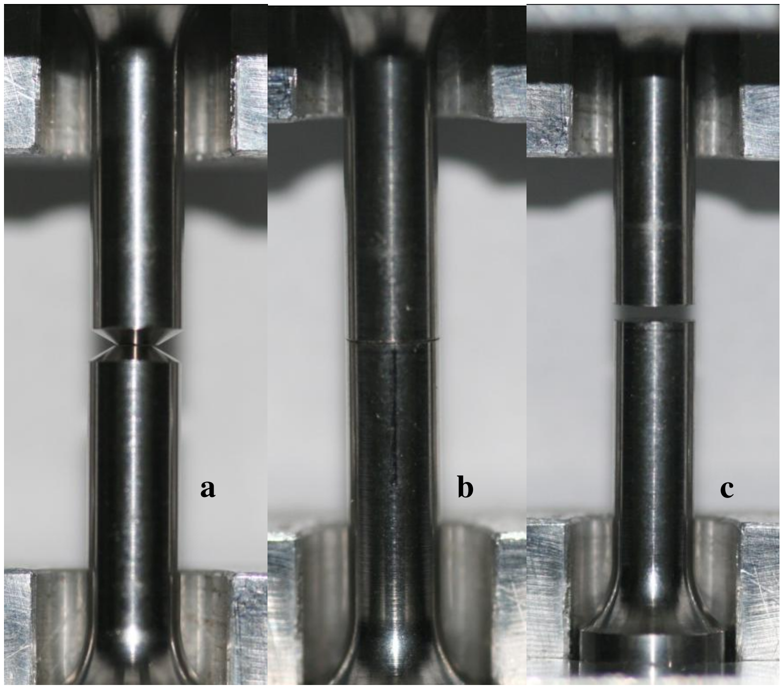

Fig. 1: Alignment of the specimens: **a. X/ b. I00/ c. I06**.

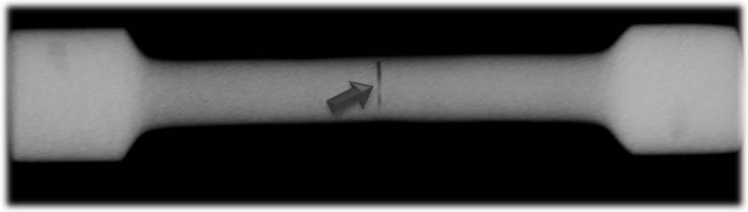

Fig. 2: Radiograph of **TIG/I00** specimen (***Arrow*** indicate void).

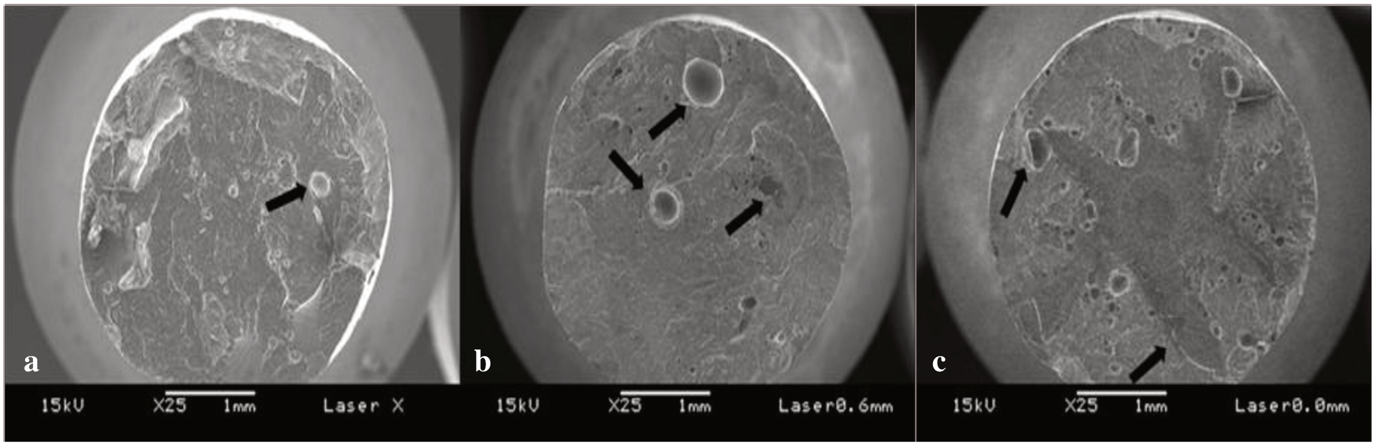

Fig. 3: SEM photomicrograph of fracture of laser-weld specimens: **a. X / b. I06 / c. I00**
(***Arrows*** indicate voids).

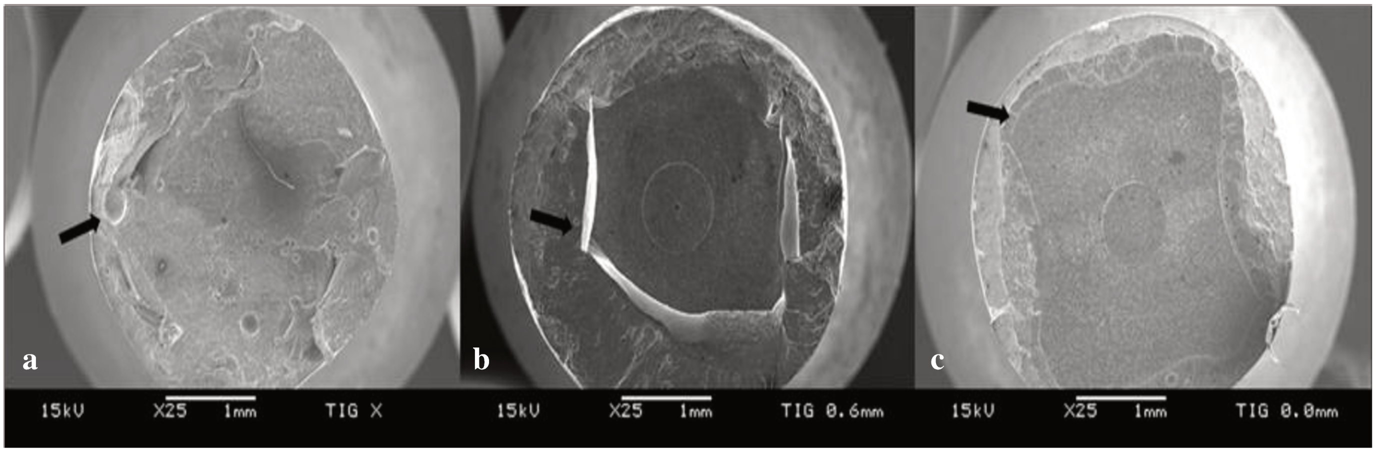

Fig. 4: SEM photomicrograph of fracture of TIG-weld specimens: **a. X / b. I06 / c. I00** (***Arrows*** indicate voids).

Figures Legends

Fig. 1: Alignment of the specimens: **a. X/ b. I00/ c. I06**.

Fig. 2: Radiograph of **TIG/I00** specimen (***Arrow*** indicate void).

Fig. 3: SEM photomicrograph of fracture of laser-weld specimens: **a. X / b. I06 / c. I00** (***Arrows*** indicate voids).

Fig. 4: SEM photomicrograph of fracture of TIG-weld specimens: **a. X / b. I06 / c. I00** (***Arrows*** indicate voids).

CAPÍTULO 2

Fatigue performance of several joint designs executed in pure titanium structures varying the type of welding

*Juliana Maria Costa Nuñez Pantoja, Post Graduated Student

Mauro Antônio de Arruda Nóbilo, PhD

Rafael Leonardo Xediek Consani, PhD

Luis Geraldo Vaz, PhD

Marcelo Ferraz Mesquita, PhD

Department of Prosthodontics and Periodontology, Piracicaba Dental School, UNICAMP, Av. Limeira, 901, 13.414-903, Piracicaba, Sao Paulo, Brazil. (**JMCNP**), (**MFM**), (**MAAN**), (**RLXC**) Department of Dentistry Materials and Prosthodontics, Araraquara Dental School, State University of Sao Paulo Julio de Mesquita Filho, Humaitá street, 1680, 14.801-903, Araraquara, Sao Paulo, Brazil (**LGV**)

This study was supported by FAPESP (Grant no 09/50567-1), and NEODENT Indústria e Comércio de Materiais Dentários – Ltda.

***[1] Corresponding author:**

Travessa Dona Eugênia street, 135, ap. 71, Sao Dimas, Piracicaba, Sao Paulo, Brazil, 13416-218.

Phone: +55 19 8203 8158; Fax: +55 19 2106 5218; e-mail: julynunez@terra.com.br

Abstract

This study evaluated the fatigue strength of pure titanium joints. Seventy dumbbell rods 3.5 mm in diameter were cast in pure titanium. Sixty specimens were sectioned and welded using three joint designs ("I," varying the welding distances (0.0 mm [*I00*] or 0.6 mm [*I06*]), or "X" [*X*]) and two welding types: *TIG* and Laser (*L*). *L* was executed as follows: 360 V/8 ms (*X*) and 390 V/9 ms (*I00* and *I06*) with the focus and frequency regulated to zero. *TIG* was executed using 2:2 (*X*) and 3:2 (*I00* and *I06*) as welding parameters. Joints were subjected to radiographic examination to visually analyze internal porosity. Specimens were subjected to mechanical cyclic tests, and the number of cycles until failure was recorded. The results were better for *X* for both welding types. *L* is more resistant than *TIG* to *I00* and *I06*. It can be concluded that the best design is *X*.

Key words: Titanium; Casting; Laser; Fatigue.

Introduction

Titanium, either commercially pure or alloyed, has been widely used as an orthopedic and dental implant material [1, 2]. It represents a nearly perfect combination of tissue compatibility and mechanical resistance [3]. The combination of high fatigue strength and high corrosion resistance makes titanium an excellent choice for many critical applications [1, 4, 5]. In dentistry, this metal was initially used in implant manufacture, but, nowadays, due to the high cost of precious metals, it has been replaced by semi-precious metals [1, 4-6]. Titanium is cheaper than gold alloys and has high corrosion resistance, sufficient cervical fit, low density, and low thermal conductibility and does not cause allergic reactions to some elements like some semi-precious metals do [1, 5, 7-11].

The absence of passivity in implant-supported prostheses may cause biological complications of the surrounding tissues due to a lack of periodontal tissues or mechanical failures of the prosthetic or implant systems [10, 12]. However, distortions can be created at any step of the fabrication process of implant-supported prostheses, making framework misfits a clinical reality [12, 13]. To avoid misfit, in some cases, it is necessary to section the framework and join it later [1, 14]. Titanium reacts to air gases when heated above 600^{o}C [8, 15]. Thus, to weld this metal, an argon inert atmosphere is necessary to prevent contamination [14, 16, 17]. The titanium welding procedure can be performed using “laser” or electric arc welding (TIG welding) [18, 19].

The welding process has great potential to affect materials’ properties, usually increasing hardness and decreasing tensile strength and ductility [14]. It has been shown that laser welding results in less distortion when compared to conventional methods [20, 21]. This technique also has demonstrated a smaller heat-affect zone (HAZ) and close yield strength to the base metal under static conditions [10, 14, 20]. It allows the procedure to be executed in the proper plaster model while avoiding several steps that are needed when conventional techniques are used [4, 6, 21]. Laser welding may be performed close to the aesthetic veneering coverage of a dental prosthesis without risk of damage because heating does not spread too far away from the welded site [4, 6, 20, 21].

Tungsten inert gas (TIG) welding is also carried out under an argon environment, and uses, as a heat source, an electric arc formed between the tungsten electrode and the metal to be welded [3, 4, 22]. During this process, heating is highly concentrated at the point where there is the contact with the electrode [4, 19, 22]. This type of welding is being widely used in current clinical practice due to the lower cost of the welding equipment compared to that of laser welding machines [4, 22].

Despite better fit, the laser-welded frameworks must not present the same fatigue strength as compared to intact structures [6, 11]. Porosities, cracks, and voids can cause significant reduction of fatigue behavior [6, 11, 21]. Thus, the protocols for implementing welding techniques and the characteristics of welded joints attract the interest of researchers who seek to evaluate the consequences of the procedures with respect to the quality of the finished joint and what it means for the longevity of prosthetic treatment [6, 11, 21].

This study was carried out to evaluate the fatigue strength of cp Ti welded structures of 3.5 mm in diameter. It evaluates the design of the joint the gives the best fatigue strength for each of the two welding techniques studied.

Material and Methods

Dumbbell-shaped rods were made in acrylic resin – Duralay II (Duralay, Reliance Dental Mfg Co, Chicago, USA) – on a metal matrix, with 3.5 mm in the central segment, based on standard ASTM E8M-04 [23]. These acrylic specimens were visually observed to identify any structural flaw and cast in a vacuum injection titanium casting machine (Rematitan System, Dentaurum, Pforzheim, Germany) with commercially pure titanium (cp Ti). To accomplish this, the acrylic resin patterns of each specimen were invested (Rematitan Plus, Dentaurum, Germany) with a 250g powder and 40mL liquid ratio prepared in a vacuum in accordance with the manufacturer's instructions. After setting, the invested specimens were heated by means of a slow heat cycle [24] to remove the wax. The specimens were cast using 36g CP Ti ingots, following identical manufacturing procedures. Upon casting, the molds were immediately immersed in cold running water.

Specimens were divested with the help of a pneumatic hammer (M320, Flli Manfredi, Sofia, Italy) and airborne particles abraded at a pressure of 0.55 MPa with 100 μm aluminum oxide particles. The specimens were finished and polished using special titanium drills (Rematitan, Dentaurum, Pforzhein, Alemanha), a rubber nº 5001 polisher (Dedeco Dental, New York, USA), and titanium polishing paste (Tiger Brilliant Polier Paste, Dentaurum, Pforzheim, Alemanha). The diameters of the specimens were constantly verified using an electronic caliper (Starret Tools Co. Ltd., Shanghai, China), during the finishing process, assuring accuracy within 0.01 mm [24].

Radiographies of the specimens were then taken using radiographic film (Ektaspeed Plus, Eastman Kodak, Rochester, USA) to identify internal defects. The radiographic exam consisted of exposing the specimens to radiation (90 kV, 15 mA, 0.6 s, and 12 mm of distance) using a periapical film [6]. The specimens that indicated voids in the dumbbell rods' central diameter were excluded.

Forty specimens were sectioned in half, perpendicularly to the long axis, and twenty of them were sectioned and beveled to form an "X" (chamfer of 1mm/30^{o}) when the parts lined up when seen from a side view (Fig 1C). The metal adjacent to the gap was blasted with abrasive particles of aluminum oxide of 100 μm at a pressure of 0.55 MPa [6]. The parts were lined up in a metal matrix to maintain distances of 0.6 and 0.0 mm between them (Fig 1A and 1B) for the "I" design. The parts were juxtaposed to form the "X" design (Fig 1C). For the laser welding procedure (*L*), the welding was carried out using energy of 360V/8ms (*X*) and 390V/9ms (*I00* and *I06*), with the focus and frequency calibrated at zero, using a Desktop –F (Dentaurum, Pforzhein, Germany) laser machine. For the TIG procedure (*TIG*), 3:2 (*I00* and *I06*) and 2:2 (*X*) were used as welding parameters, with the ratio numerator representing the power used and the denominator indicating the programmed time. These procedures were conducted in an NTY 60C machine (Kernit, Indústria Mecatrônica Ltda., Indaiatuba, Brazil). The joints were welded on opposite sides of the cross section to stabilize the aligned parts of the specimen, and, then, the welding was completed by a trained and competent professional. The "X" designs were welded starting in the center of the specimens to mark a full contact in the center of each half-piece, and, then, the welding was completed using pure titanium grade II metal filler (Dentaurum, Pforzhein, Germany) to provide a total joint between the two parts.

The number of variables (welding design and type of welding) dictated a total of 6 groups (n=10), which, together with the control group (intact), made a total of 7 groups: G_1- *L/X*, G_2- *L/I06*, G_3- *L/I00*, G_4- *TIG/X*, G_5- *TIG/I06*, G_6- *TIG/I00*, and G_7- Intact.

After the welding process, the joints were finished and polished as it was done after the casting process, and radiographies of the joints were taken in the same way. These radiographies were visually analyzed for the presence of internal voids in the joint (Fig. 2).

Three randomly extra intact specimens at the same diameter were subjected to tensile tests, before fatigue tests, to establish the mean yield strength at 0.2% permanent strain. The stress established for the fatigue test (124 MPa) was calculated as 30% of the mean yield strength [6]. The loading frequency in the test was 15 Hz, and the loadings calculated were 1193 N. The specimens were then subjected to fatigue strength tests using a testing machine (Test Star II, Material Testing System, MTS Systems Corp, Minneapolis, Minn), and the number of cycles required to cause the fatigue fracture was registered. The specimens were tested submerged in synthetic saliva (1.5 mM Ca, 3.0 mM P, 20.0 mM $NaHCO_3$, pH 7.0) [25] at room temperature for 100 000 cycles [6].

The fatigue fracture surfaces of representative specimens were examined using a scanning electron microscope (Electron Probe Microanalyser, Jeol model JXA 840 A, Jeol Ltd, Tokyo, Japan).

An exploratory analysis of the number of cycles was done, and a nonparametric test was requested. The Kruskal-Wallis and Dunn tests (α=0.05) were applied to compare the designs (3 to 3) with respect to the type of welding. The Mann-Whitney test (α=0.05) was applied to compare the type of welding (2 to 2) with respect to the design and to compare (2 to 2) the control group with the other groups. The data from the radiographic analyses were correlated with the number of cycles, using the Spearman correlation coefficient (α=0.05), to each type of welding/design and to all grouped data.

Results

Table 1 shows the median values of the number of cycles until fracture in the experimental groups. The number of cycles required until specimen fracture was higher for *X* with respect to both the welding procedure and *I06* for the laser welding ($p<0.05$). A lower number of cycles was found for *I00* for both welding types and for *I06* for *TIG* ($p<0.05$). When the two types of welding were compared, the Mann-Whitney test (α=0.05) indicated a difference only for *I06* (p=0.0002), and the laser weld joints behaved better.

In the control group, the specimens resisted, mostly, over 100 000 cycles. The specimens included in G_1, G_2, and G_4 behaved similarly to the control group specimens ($p>0.05$).

The Spearman correlation coefficient ($\alpha=0.05$) suggested a negative correlation ($p<0.0001$/rs=-0.7276) between the number of cycles and the presence of porosity in the radiographies (Table 2), except when G_5 data were correlated. In these cases, specimens that had pores inside the joints, observed in the radiographies (Fig. 2), were less resistant to mechanical cycling.

Figures 3 and 4 show SEM photomicrographs of fractured laser-welded and TIG-welded joint surfaces, respectively. Figures 3A and 4A show the surfaces of X designs. Figures 3B and 4B show the 0.6 mm-joint opening with the "I" design. Figures 3C and 4C show 0.0 mm joint opening with the "I" design. The presence of internal pores or voids caused by insufficient weld penetration (arrows) was higher for *I00* for both type of welding. For *I06*, the *TIG* procedure showed less welding penetration than laser welding, and the internal void was small in *X*.

Discussion

Passive fit to osseointegrated abutments and modifications or repair of fractured prostheses can be achieved with welding procedures [4, 6, 11, 14, 17, 21, 26, 27]. However, weak joint strength can cause failure of the metal framework during habitual use [6, 11, 17]. A low percentage of welded areas can affect the cyclic strength of joint structures, and the welding design has a significant effect on the penetration depth of the welded cross section [6, 17, 21].

All of the concepts discussed here corroborate the results of this study. The design of the union clearly influenced the quality of the welded joint. SEM photomicrographs (Fig 3C and 4C) of fractured surfaces show large voids for *I00* inside the welded zone, indicating insufficient weld penetration in the cross section of 3.5 mm in diameter when there is juxtaposition of the parties, which agrees with the results of Nuñez-Pantoja *et al.* (2010) [6]. According to some authors, the laser beam penetrates to a limited depth of 1.5 mm [16, 20, 26].

Even knowing that commercially pure titanium's weldability is considered good [19, 28], it is also known not only that the joint quality is influenced by characteristics of welded metal but also that peculiar characteristics of the various welding procedures directly influence the outcome of the welded joint.

TIG welding procedure uses, as an energy source, an electric arc formed by the contact of a tungsten electrode with the metal that will be welded [4, 19]. Thus, this procedure has a punctual characteristic, making it difficult to penetrate the solder in cross sections of large diameters, even when there is a gap between the parts [4, 19]. The thickness of the electrode makes difficult the contact of the tungsten electrode with the filler metal interposed between the parts to be welded, mainly in small gaps. Laser welding, in turn, uses, as heat source, a high-power light beam that is capable of penetrating thin thicknesses, or reaching the filler metal present in the central portion of the gap to be welded [20, 29]. These statements can explain the better results for the *I06* laser-welded procedure (Table 1) compared to the *I06* welded using the TIG procedure.

The *I06* laser-welded joint obtained in this study was as resistant to mechanical cycling as the control group, which disagrees the results of Nuñez-Pantoja *et al.* (2010) [6]. In that study, all of the "I" design joints were weaker than intact dumbbells. Although the two studies were conducted using similar methods, the methodology used during the execution of the alignment of the specimens to be welded was not the same. In this study, a welding device was used. This device allowed the alignment of the parts according to different designs, without to the necessity for creating the joint of the aligned parts with acrylic resin. Thus, the metal to be welded could be cleaned better, with less likelihood of introducing inclusions originating from microscopic remnants of acrylic resin, solid vaseline, or oxide particles.

The "X" design was necessary to obtain TIG-welded joints that were as strong as the specimens in the control group. Some studies have indicated that, if two adjacent joint-forming surfaces are ground so as to form the shape of the letter "X", they can be welded starting from the center, and the joint is built toward the surface of the object [21, 30], which my enable the creation of higher-quality joints, especially when welding is performed on thick structures [21]. This design can be also considered a great design to laser-weld cross sections of 3.5 mm in thickness, according to the results of this study.

With regard to the radiographic analysis of the joints, the Spearman correlation coefficient ($\alpha=0.05$) demonstrated a negative correlation ($rs=-0.7276$; $p<0.0001$) when all groups were correlated. It was observed that, when there was porosity inside the joints in radiographic analyses (Fig. 2), the number of cycles dropped, which corroborates the work of Nuñez-Pantoja *et al*. (2010) [6]. This correlation indicates that internal porosity acts to raise stress [6, 11], decreasing the fatigue life of the prostheses. These imperfections lead to the development of microcracks, which coalesce and, ultimately, result in microscopic cracking and failure [6, 11, 21].

With the intention of simulating the joints executed in structures made by dental prosthesis laboratories, the specimens were cast in this study. To isolate the variable of internal porosity inherent to the casting process [11, 15], radiographies of the specimens were taken to exclude faulty cast specimens. However, some specimens in the control group fractured, and some of the others did not fracture in joint or near the joint (HAZ). Perhaps some small pores cannot be seen in a radiographic exam, or, perhaps, the manual finishing and polishing after the casting process, despite being standardized, influenced crack initiation and propagation [15].

It should be pointed out that this study did not aim to quantify welded framework survival [3]. The aim of this study was to determine which welded joint design gives the best fatigue strength for each of the two welding techniques studied. As such, no conclusions can be made with regard to the number of years that welded frameworks will survive in a buccal cavity. Thus, analysis of the fatigue strength of welded prosthetic frameworks is suggested, using a simulated masticatory load, to verify the longevity of these joints in simulated clinical situations.

Conclusions

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

1. For TIG-welded structures of 3.5 mm cross sections, it is necessary to employ the "X" design.
2. For laser-welded structures, these joints can be made using the "X" design or the "I" design without juxtaposition of the parts.
3. Radiographs of the welded joints can be considered a way to perform quality control for TIG- and laser-welded joints in clinical practice.

Acknowledgements

FAPESP Sao Paulo Reserch Foundation (Grant no 09/550567-1), and **NEODENT**.

References

1. Blackman R, Barghi N, Tran C. Dimensional changes in casting titanium removable partial denture frameworks. J Prosthet Dent. 1991;65(2):309-15.
2. Wang RR, Fenton A. Titanium for prosthodontic applications: a review of the literature. Quintessence Int. 1996;27(6):401-8.
3. Wiskott HW, Doumas MT, Scherrer SS, Susz C, Belser UC. Microstructures of brazings and welds using grade 2 commercially pure titanium. Int J Prosthodont. 2001;14(1):40-7.
4. Rocha R, Pinheiro AL, Villaverde AB. Flexural strength of pure Ti, Ni-Cr and Co-Cr alloys submitted to Nd:YAG laser or TIG welding. Braz Dent J. 2006;17(1):20-3.
5. Wang RR, Fenton A. Titanium for prosthodontic applications: a review of the literature. Quintessence Int. 1996;27(6):401-8.
6. Nuñez-Pantoja JM, Vaz LG, Nóbilo MA, Henriques GE, Mesquita MF. Effects of laser-weld joint opening size on fatigue strength of Ti-6Al-4V structures with several diameters. J Oral Rehabil. 2010 Jul 29. [Epub ahead of print]
7. Jones TK, Hansen CA, Singer MT, Kessler HP. Dental implications of nickel hypersensitivity. J Prosthet Dent. 1986;56(4):507-9.
8. Taira M, Moser JB, Greener EH. Studies of Ti alloys for dental castings. Dent Mater. 1989;5(1):45-50.
9. Lautenschlager EP, Monaghan P. Titanium and titanium alloys as dental materials.Int Dent J. 1993;43(3):245-53.
10. Chai T, Chou CK. Mechanical properties of laser-welded cast titanium joints under different conditions. J Prosthet Dent. 1998;79(4):477-83.
11. Zavanelli RA, Guilherme AS, Pessanha-Henriques GE, de Arruda Nóbilo MA, Mesquita MF. Corrosion-fatigue of laser-repaired commercially pure titanium and Ti-6Al-4V alloy under different test environments. J Oral Rehabil. 2004;31(10):1029-34.
12. Spazzin AO, Henriques GE, de Arruda Nóbilo MA, Consani RL, Correr-Sobrinho L, Mesquita MF. Influence of prosthetic screw material on joint stability in passive and non-passive implant-supported dentures. Open Dent J. 2009 Dec 30;3:245-9.
13. Spazzin AO, Henrique GE, Nóbilo MA, Consani RL, Correr-Sobrinho L, Mesquita MF. Effect of retorque on loosening torque of prosthetic screws under two levels of fit of implant-supported dentures. Braz Dent J. 2010;21(1):12-7.
14. Sjögren G, Andersson M, Bergman M. Laser welding of titanium in dentistry. Acta Odontol Scand. 1988;46(4):247-53.
15. Guilherme AS, Henriques GE, Zavanelli RA, Mesquita MF. Surface roughness and fatigue performance of commercially pure titanium and Ti-6Al-4V alloy after different polishing protocols. J Prosthet Dent. 2005;93(4):378-85.

16. Roggensack M, Walter MH, Böning KW. Studies on laser- and plasma-welded titanium. Dent Mater. 1993;9(2):104-7.
17. Watanabe I, Topham DS. Laser welding of cast titanium and dental alloys using argon shielding. J Prosthodont. 2006;15(2):102-7.
18. Berg E, Wagnere WC, Davik G, Dootz ER. Mechanical properties of laser-welded cast and wrought titanium. J Prosthet Dent. 1995;74(3):250-7.
19. Wiskott HW, Doumas MT, Scherrer SS, Susz C, Belser UC. Microstructures of brazings and welds using grade 2 commercially pure titanium. Int J Prosthodont. 2001;14(1):40-7.
20. Tambasco J, Anthony T, Sandven O. Laser welding in the dental laboratory: an alternative to soldering. J Dent Technol. 1996; 13(4):23-31.
21. Bertrand C, Poulon-Quintin A. Proposals for optimization of laser welding in prosthetic dentistry. J Prosthodont. 2010;19(1):69-76
22. Bock JJ, Bailly J, Gernhardt CR, Fuhrmann RA. Fracture strength of different soldered and welded orthodontic joining configurations with and without filling material. J Appl Oral Sci. 2008;16(5):328-35.
23. ASTM International. Designation: E 8M – 04: Standard Test Methods for Tension Testing of Metallic Materials [Metric]. West Conshohocken; 2004. 24p.
24. Nuñez JM, Takahashi JM, Henriques GE, Nóbilo MA, Consani RL, Mesquita MF. Radiographic inspection of porosity in pure titanium dumbbell castings. Gerodontology. 2010 May 26. doi: 10.1111/j.1741-2358.2010.00376.x. [Epub ahead of print]
25. Birkeland JM. The effect of pH on the interaction of fluoride and salivary ions. Caries Res. 1973;7(1):11-8.
26. Baba N, Watanabe I. Penetration depth into dental casting alloys by Nd:YAG laser. J Biomed Mater Res B Appl Biomater. 2005;72(1):64-8.
27. Tiossi R, Falcão-Filho H, Aguiar Júnior FA, Rodrigues RC, Mattos Mda G, Ribeiro RF. Modified section method for laser-welding of ill-fitting cp Ti and Ni-Cr alloy one-piece cast implant-supported frameworks. J Oral Rehabil. 2010;37(5):359-63.
28. Lampman S. Wrought titanium and titanium alloys. In: Metal handbook. 10 ed. Metals Park: ASM; 1990. V.2, p. 592 – 633.
29. Fornaini C, Bertrand C, Rocca JP, Mahler P, Bonanini M, Vescovi P, Merigo E, Nammour S. Intra-oral laser welding: an in vitro evaluation of thermal increase. Lasers Med Sci. 2010;25(4):473-7.
30. Zupancic R, Legat A, Funduk N. Tensile strength and corrosion resistance of brazed and laser-welded cobalt-chromium alloy joints. J Prosthet Dent. 2006;96(4):273-82.

Table 1. Median (standard deviation) values of number of cycles resisted in the experimental groups

Designs	Type of welding	
	Laser	TIG
"X"	93452 (8284.377) Aa	100000 (10260.05) Aa
"I"/distance 0.6 mm	100000 (8368.399) Aa	*26567 (4745.308) Bb
"I"/distance 0.0 mm	* 22734 (7002.697) Ba	*19262 (7041.156) Ba

Intact group = 100 000 (3659.202). * Differs from intact group (Mann-Whitney, alfa=0.05).

Median values followed by the same letter are not significantly different (Kruskal-Wallis / Dunn, alfa=0.05). Capital letters compare designs inside the level of the factor type of welding and small letters compare type of welding inside the level of design factor.

Table 2. Association of number of cycles survived and presence of internal voids in radiography exam

		Spearman Coefficient	(p)=
Laser	"X"	-0.8231	0.0034
	"I" distance 0.6 mm	-0.6425	0.0451
	"I" distance 0.0 mm	-0.8528	0.0017
TIG	"X"	-0.7845	0.0072
	"I" distance 0.6 mm	*	*
	"I" distance 0.0 mm	-0.7977	0.0057
All type of welding / All designs		-0.7276	< 0.0001

* All specimens' present pores in radiographies, thus correlation was impossible.

Spearman Coefficient ranges between -1 and 1.

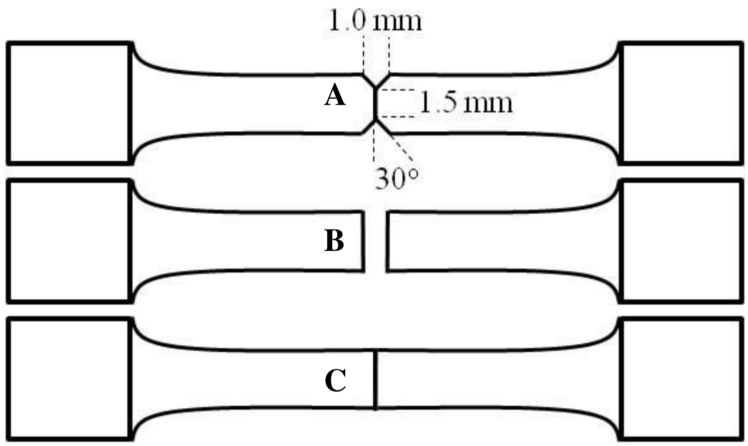

Fig 1. Joint designs: **A) X**; **B) I06**; **C) I00.**

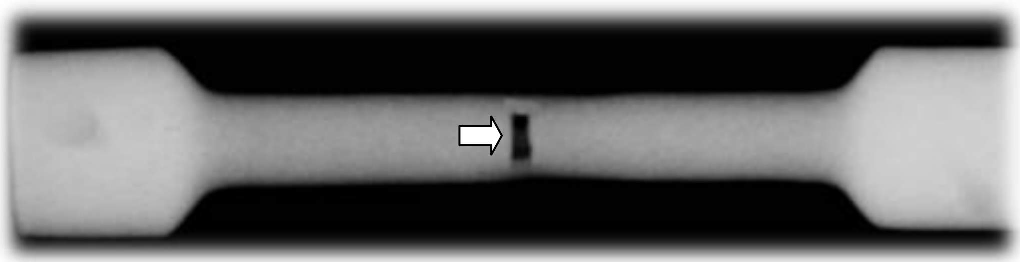

Fig 2. Radiograph of **TIG/I06** specimen (***Arrow*** indicate void).

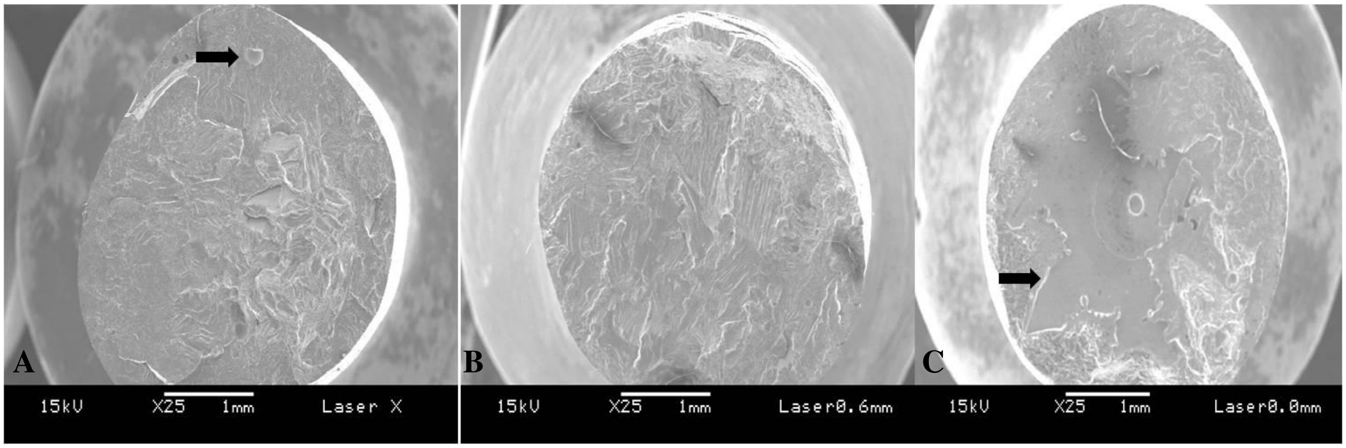

Fig. 3: SEM photomicrograph of fracture of laser-weld specimens: **A. X / B. I06 / C. I00** (*Arrows* indicate voids).

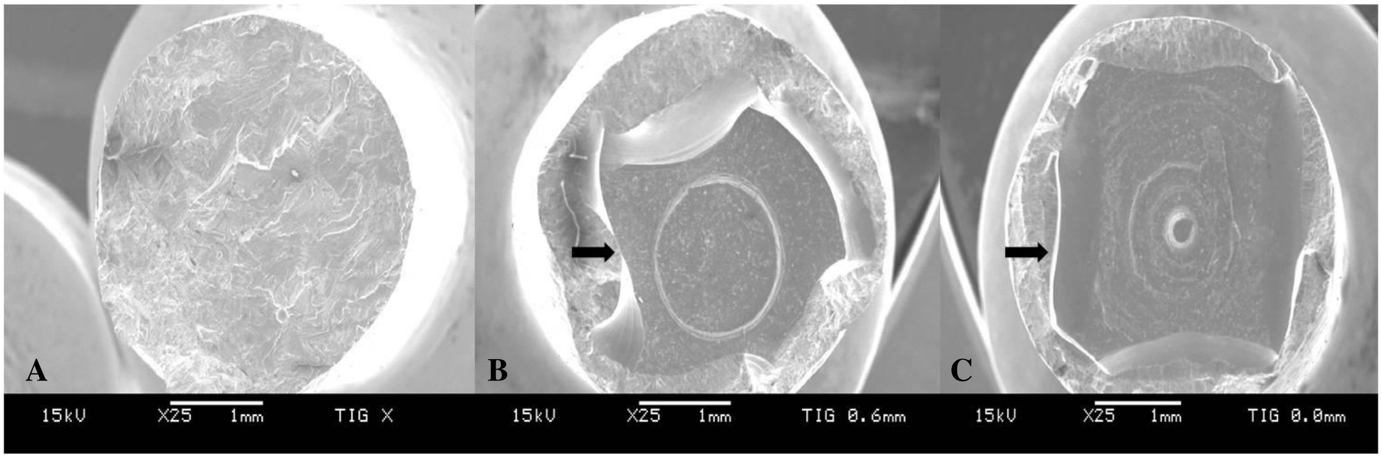

Fig. 4: SEM photomicrograph of fracture of TIG-weld specimens: **A. X / B. I06 /** C. **I00** (*Arrows* indicate voids).

3. CONSIDERAÇÕES GERAIS

Diante dos resultados desses estudos, pode-se notar uma grande diferença nos valores de número de ciclos resistidos até a fratura para o Ti cp quando comparado a liga de Ti-6Al-4V. Para os corpos de prova usinados em liga de Ti-6Al-4V, todos os halteres soldados resistiram menos que os intactos. Já para os fundidos em Ti cp, os corpos de prova soldados com design em "X", para os dois tipos de solda, e os com design em "I" e distância 0.6 mm, soldados a laser, resistiram da mesma maneira que o grupo controle (intactos). Além disso, ao serem analisadas as tabelas dos valores originais (Tabelas 1, 2, 3, 4 5 e 6 / Apêndice), percebe-se que os valores médios resistidos pelos espécimes usinados em liga de Ti-6Al-4V soldados foram numericamente inferiores aos valores resistidos pelos espécimes fundidos em Ti cp, e soldados.

Isso pode ser explicado primeiramente pelo fato de que os halteres obtidos em Ti-6Al-4V foram usinados, ou seja, são isentos de porosidades internas e irregularidades externas, o que não acontece com regra, para os halteres fundidos. Assim, quando se comparam os grupos intactos de ambos os materiais, mesmo que não seja feita uma análise estatística dos valores obtidos, sabe-se que, naturalmente, os corpos de prova usinados resistirão a um maior número de ciclos que os corpos de prova fundidos. Isto, associado ao fato de que as propriedades de resistência a tração do Ti-6Al-4V são superiores às do Ti cp (Wang & Fenton, 1996; Wiskott *et al.*, 2001; Watanabe & Topham, 2004; Guilherme *et al.*, 2005), fizeram com que o levantamento das curvas tensão x deformação levassem a valores muito menores de tensão de escoamento para o Ti cp (413 MPa), e consequentemente à carga de ciclagem mecânica também menor (1193 N), quando comparado ao Ti-6Al-4V (2800 N).

Além disso, sabe-se que a soldabilidade do titânio decresce à medida que aumentam os níveis de β-estabilizadores nas ligas (Wiskott *et al.*, 2001). Sabe-se, também, que o vanádio é um β-estabilizador (Lampman, 1990). Assim, pode-se afirmar que a soldabilidade do Ti cp é maior que a da liga de Ti-6Al-4V. Este fato pode, também, ter motivado a diferença numericamente maior dos valores obtidos para o Ti cp quando comparados aos obtidos para a liga de Ti-6Al-4V.

Outro argumento que complementa a justificativa dessa diferença apóia-se na estrutura cristalina desses dois metais. A estrutura cristalina do titânio pode ser α (hexagonal de corpo fechado) ou β (cúbica de corpo centrado) (Lampman, 1990). O Ti cp, em temperatura ambiente apresenta estrutura cristalina α, e a liga de Ti-6Al-4V, α-β (Lampman, 1990; Wiskott *et al.,* 2001). A incompatibilidade dos cristais na estrutura cristalina α-β gera falhas internas, as quais irão originar tensões residuais internas (Lampman, 1990; Wiskott *et al.,* 2001). Dessa maneira, uniões soldadas em ligas α-β têm maior probabilidade de apresentar falhas no interior da solda, e de fraturar com pouca deformação plástica (Lampman, 1990; Wiskott *et al.,* 2001). Esse problema pode ser minimizado através do tratamento térmico das partes a serem unidas antes da realização do procedimento de união para reduzir essas tensões internas previamente à soldagem, ou após

(Lampman, 1990; Wiskott *et al.*, 2001; Bertrand & Polon-Quintin, 2010). Todavia, este tratamento infelizmente não é feito nos laboratórios de prótese na prática clínica atual.

Seria interessante, a realização de estudos comparativos utilizando-se corpos de prova de Ti cp e de Ti-6Al-4V, obtidos da mesma maneira, ou usinados, ou fundidos, para que a comparação da soldabilidade desses materiais seja feita com mais propriedade. Mas este não foi o objetivo desse estudo.

Os resultados dos dois trabalhos concordam que as uniões de secções transversais de 3,5 mm são mais resistentes quando são executadas fazendo-se uso do design em "X". Isso ocorre, principalmente, quando se trata do procedimento TIG de soldagem, visto que a fonte de energia para a realização dessa união é pontual, e se dá pelo contato do eletrodo de tungstênio com a peça a ser soldada, que origina um arco elétrico (Wiskott *et al.*, 2001; Cardoso, 2007). Dessa forma, a penetração da solda no interior de corpos de prova espessos fica dificultada, mesmo quando há abertura da junta a ser soldada, já que o eletrodo não alcança a porção central da união, especialmente em espaçamentos pequenos (0.6 mm). Ao contrário, o feixe de laser pode alcançar essa região e fundir o metal de adição interposto entre as duas partes. O feixe de laser pode ser considerado um feixe de luz monocromática e direcional, de alta energia, capaz de ser focado em áreas muito pequenas (Cardoso, 2007). Isso pode explicar os resultados superiores da união soldada a laser quando comparada à soldada a TIG para o design em "I" com espaçamento de 0,6 mm entre as partes.

Em estudos anteriores (Nuñez *et al.,* 2009; Nuñez-Pantoja *et al.*, 2010), que utilizaram metodologia semelhante, o espaçamento entre as partes não originou halteres tão resistentes quanto os obtidos nesse estudo. Isso ocorreu, provavelmente, devido a um aprimoramento da metodologia empregada. O atual estudo fez uso de um dispositivo de alinhamento das partes a serem soldadas (Fig. 1 e 2 / Apêndice). O que evitou o uso de vaselina sólida e resina acrílica para a padronização dos espaçamentos e fixação das peças alinhadas. Dessa maneira, foi possível realizar a limpeza e jateamento das partes a serem soldadas com menor probabilidade de ocorrem inclusões originadas de restos microscópicos desses materiais empregados.

Para os corpos de prova soldados com justaposição das partes e design em "I", pode-se observar que a qualidade da união gerada é ruim, para ambos os materiais e para os dois procedimentos de soldagem testados. Isso ocorreu devido à insuficiente penetração do feixe de laser nas uniões com justaposição das partes (Sjögren *et al.*, 1988; Wang & Fenton, 1996; Nuñez-Pantoja *et al.*, 2010), e à dificuldade de se soldar a porção central, independente do espaçamento das partes a serem unidas, quando se realiza a união por meio da solda TIG, pela característica pontual dessa soldagem. Quanto menor a profundidade alcançada pelo feixe de laser, maior a quantidade de vazios internos (Baba & Watanabe, 2005; Nuñez-Pantoja *et al.,* 2010). Segundo diversos autores, essa penetração é limitada a 1,5 mm de profundidade (Roggensach *et al.*, 1993; Neo *et al.*, 1996; Tambasco *et al.*, 1996; Baba & Watanabe, 2005).

Além disso, notou-se haver correlação negativa entre número de ciclos e presença de porosidades internas nas radiografias das uniões realizadas em ambos os materiais estudados. Quanto menor a incidência de vazios verificados nas radiografias analisadas, maior o número de ciclos, o que concorda com Nuñez-Pantoja e colaboradores (2010). Esse achado reforça a idéia de que a análise radiográfica das uniões pode ser considerada uma excelente maneira de se realizar o controle de qualidade das soldas. Desse modo, o cirurgião dentista pode realizá-la na sessão de aprovação da infraestrutura soldada, e como consequência, irá instalar a prótese analisada com mais segurança. Todavia, é muito importante ressaltar que nesse estudo, não foram avaliadas as dimensões dos vazios e nem a quantidade de poros.

Nesse estudo, pode-se perceber também, que quando se pretende avaliar a qualidade da solda, exclusivamente, o emprego de corpos de prova usinados é recomendado. Alguns halteres fundidos e soldados falharam em regiões distantes da união soldada; já os usinados romperam sempre na região próxima à solda. Isso comprova a dificuldade de se isolar as variáveis externas inerentes ao processo de fundição, e ao processo de acabamento e polimento manual, que interferem diretamente na resistência à fadiga dos materiais. Contudo, os corpos de prova fundidos em Ti cp foram empregados nesse estudo com a finalidade de haver uma aproximação do trabalho com a realidade clínica atual dos laboratórios de prótese. Quando são confeccionadas estruturas metálicas implanto-suportadas de titânio, essas estruturas são geralmente fundidas em Ti cp pelos laboratórios de prótese; todavia, devido a dificuldade de se usinar o Ti cp (Lampman, 1990; Ohkubo *et al.*, 2000), quando essas estruturas são feitas por meio da técnica de CAD-CAM, elas são usinadas em Ti-6AL-4V grau V (Ohkubo *et al.*, 2000; Nuñez-Pantoja *et al.*, 2010).

Existe ainda a necessidade de realização de novos estudos a cerca do efeito do design da união e do tipo de procedimento de soldagem sobre o assentamento e longevidade das próteses. É importante que sejam obtidas informações quanto à melhor maneira de soldar infraestuturas fundidas em Ti cp ou usinadas em liga de Ti-6Al-4V, para que sejam adaptadas aos pilares com mais precisão, causando menor tensão aos implantes, sem que a prótese frature após a instalação por falha mecânica na resistência da união. Contudo, pode-se dizer que apesar das limitações desse estudo, puderam ser fornecidos dados muito importantes para o esclarecimento do comportamento da união soldada a TIG ou a laser, em função dos diferentes designs de união sugeridos na literatura.

4. CONCLUSÃO

Dentro das limitações desses estudos e com base nos resultados obtidos, pode-se concluir que:

1. Quando se pretende soldar pela técnica TIG estruturas de 3,5 mm de diâmetro de secção transversal, faz-se necessário o uso do design em “X”;

2. Para estruturas soldadas a laser, essa união pode ser feita utilizando-se o design em “X”, ou o design em “I” com 0,6 mm de espaçamento entre as partes;

3. As radiografias das uniões soldadas podem ser consideradas uma maneira de se realizar o controle de qualidade das junções soldadas a TIG e a laser na prática clínica;

REFERÊNCIAS

Baba N, Watanabe I. Penetration depth into dental casting alloys by Nd:YAG laser. J Biomed Mater Res B Appl Biomater. 2005 ;72(1):64-8.

Berg E, Wagnere WC, Davik G, Dootz ER. Mechanical properties of laser-welded cast and wrought titanium. J Prosthet Dent. 1995;74(3):250-7.

Bertrand C, Poulon-Quintin A. Proposals for optimization of laser welding in prosthetic dentistry. J Prosthodont. 2010;19(1):69-76.

Bock JJ, Bailly J, Gernhardt CR, Fuhrmann RA. Fracture strength of different soldered and welded orthodontic joining configurations with and without filling material. J Appl Oral Sci. 2008;16(5):328-35.

Cardoso, LM. Avaliação da resistência à flexão de liga de titânio submetida a soldagens a laser e TIG, visando à prótese dentária. 2007. Dissertação de Mestrado – Instituto de Pesquisas Energéticas e Nucleares.

Chai T, Chou CK. Mechanical properties of laser-welded cast titanium joints under different conditions. J Prosthet Dent. 1998;79(4):477-83.

Craig RG, Hanks CT, Kohn III A, O'Brien WJ, Peuton FA, Powers JM *et al*. Restorative Dental Materials. 10. ed. Saint Louis:Mosby; 1997. Cap 15: Cast and wrought base metal alloys, p.428-432.

Dieter GE, Metalurgia mecânica 2.ed. Rio de Janeiro: Guanabara Koogan, 1981. Cap 12: Fadiga dos metais, 345-384.

Ferreira I, Costa NG, Tokimatsu RC. Biomateriais: Fadiga dos metais. 1998; p. 32-34.

Gordon TE, Smith DL. Laser welding of prostheses--an initial report. J Prosthet Dent. 1970;24(4):472-6.

---

* De acordo com a norma da UNICAMP/FOP, baseada na norma do International Committee of Medical Journal Editors – Grupo de Vancouver. Abreviatura dos periódicos em conformidade com Medline.

Guilherme AS, Henriques GE, Zavanelli RA, Mesquita MF. Surface roughness and fatigue performance of commercially pure titanium and Ti-6Al-4V alloy after different polishing protocols. J Prosthet Dent. 2005;93(4):378-85.

Henriques GE, Consani S, Rollo JM, Andrade e Silva F. Soldering and remelting influence on fatigue strength of cobalt-chromium alloys. J Prosthet Dent. 1997;78(2):146-52.

Huling JS, Clark RE. Compratative distortion in three-unit fixed prostheses joined by laser welding, conventioanl soldering, or casting in one piece. J Dent Res. 1977;56(2):128-34.

Lampman S. Wrought titanium and titanium alloys. In: Metal handbook. 10 ed. Metals Park: ASM; 1990. V.2, p. 592 – 633.

Lautenschlager EP, Monaghan P. Titanium and titanium alloys as dental materials. Int Dent J. 1993;43(3):245-53.

Liu J, Watanabe I, Yoshida K, Atsuta M. Joint strength of laser-welded titanium. Dent Mater. 2002 Mar;18(2):143-8.

Neo TK, Chai J, Gilbert JL, Wozniak WT, Engelman MJ. Mechanical properties of titanium connectors. Int J Prosthodont. 1996;9(4):379-93.

Nuñez, JMC. Estudo da qualidade da fundição do Titânio comercialmente puro e da união soldada a laser em estruturas de Ti cp e em liga de Ti-6Al-4V. 2009. Dissertação de Mestrado – Faculdade de Odontologia de Piracicaba.

Nuñez-Pantoja JM, Vaz LG, Nóbilo MA, Henriques GE, Mesquita MF. Effects of laser-weld joint opening size on fatigue strength of Ti-6Al-4V structures with several diameters. J Oral Rehabil. 2010 Jul 29. [Epub ahead of print]

Ohkubo C, Watanabe I, Ford JP, Nakajima H, Hosoi T, Okabe T. The machinability of cast titanium and Ti-6Al-4V. Biomaterials. 2000;21(4):421-8.

Rocha R, Pinheiro AL, Villaverde AB. Flexural strength of pure Ti, Ni-Cr and Co-Cr alloys submitted to Nd:YAG laser or TIG welding. Braz Dent J. 2006;17(1):20-3.

Roggensack M, Walter MH, Böning KW. Studies on laser- and plasma-welded titanium. Dent Mater. 1993;9(2):104-7.

Sahin S, Cehreli MC. The significance of passive framework fit in implant prosthodontics: current status. Implant Dent. 2001;10(2):85-92.

Sjögren G, Andersson M, Bergman M. Laser welding of titanium in dentistry. Acta Odontol Scand. 1988;46(4):247-53.

Souza PCRD,Dinato JC, Beatrice CRS, Guastaldi AC, Bottino MA. Soldagem na Odontologia: estudo das uniões soldadas empregando laser e brazagem. Revista da APCD. 2000; 54(6):470-475.

Spazzin AO, Henriques GE, de Arruda Nóbilo MA, Consani RL, Correr-Sobrinho L, Mesquita MF. Influence of prosthetic screw material on joint stability in passive and non-passive implant-supported dentures. Open Dent J. 2009;3:245-9.

Taira M, Moser JB, Greener EH. Studies of Ti alloys for dental castings. Dent Mater. 1989;5(1):45-50.

Tambasco J, Anthony T, Sandven O. Laser welding in the dental laboratory: an alternative to soldering. J Dent Technol. 1996; 13(4):23-31.

Vallittu PK, Luotio K. Effects of cobalt-chromium alloy surface casting on resistance to deflection fatigue and surface hardness of titanium. Int J Prosthodont. 1996;9(6):527-32.

Wang RR, Chang CT. Thermal modeling of laser welding for titanium dental restorations. J Prosthet Dent. 1998;79(3):335-41.

Wang RR, Fenton A. Titanium for prosthodontic applications: a review of the literature. Quintessence Int. 1996;27(6):401-8.

Watanabe I, Topham DS. Tensile strength and elongation of laser-welded Ti and Ti-6AL-7NB. J Biomed Mater Res B Appl Biomater. 2004;71(1):46-51.

Wee AG, Aquilino SA, Schneider RL. Strategies to achieve fit in implant prosthodontics: a review of the literature. Int J Prosthodont. 1999;12(2):167-78.

Anselm Wiskott HW, Doumas T, Scherrer SS, Belser UC, Susz C. Mechanical and structural characteristics of commercially pure grade 2 Ti welds and solder joints. J Mater Sci Mater Med. 2001;12(8):719-25.

Zavanelli RA, Guilherme AS, Pessanha-Henriques GE, de Arruda Nóbilo MA, Mesquita MF. Corrosion-fatigue of laser-repaired commercially pure titanium and Ti-6Al-4V alloy under different test environments. J Oral Rehabil. 2004;31(10):1029-34.

Zupancic R, Legat A, Funduk N. Tensile strength and corrosion resistance of brazed and laser-welded cobalt-chromium alloy joints. J Prosthet Dent. 2006;96(4):273-82.

APÊNDICE

Tabela 1. Valores originais do número de ciclos até a fratura para os corpos de prova intactos usinados em liga de Ti-6Al-4V.

CP \ Design	Intacto
1	100 000
2	100 000
3	100 000
4	100 000
5	100 000
6	100 000
7	100 000
8	100 000
9	100 000
10	100 000
Mediana	100 000
Média	100 000
Desv. Padrão	0

Tabela 2. Valores originais do número de ciclos até a fratura para os corpos de prova usinados em liga de Ti-6Al-4V e soldados a laser.

CP \ Design	X	I06	I00
1	25602	35156	4000
2	38581	27688	1540
3	18467	36165	2361
4	22484	16412	2424
5	31013	35798	3069
6	33740	35397	1037
7	24739	24230	3336
8	38372	25264	5076
9	43144	44456	3016
10	34567	26007	2217
Mediana	32376,5	31422	2720
Média	31070,9	30657,3	2807,6
Desv. Padrão	8009,803	8121,557	1173,583

Tabela 3. Valores originais do número de ciclos até a fratura para os corpos de prova usinados em liga de Ti-6Al-4V e soldados por meio de solda TIG.

Design / CP	X	I06	I00
1	37398	4564	973
2	57142	12120	2166
3	33033	9114	1872
4	42655	8618	1268
5	24095	6858	897
6	51786	10337	991
7	32043	9392	624
8	50989	5147	969
9	32823	9985	2577
10	43769	6285	1174
Mediana	40026,5	8866	1082,5
Média	40573,3	8242	1351,1
Desv. Padrão	10512,35	2439,085	635,0864

Tabela 4. Valores originais do número de ciclos até a fratura para os corpos de prova intactos fundidos em Ti cp.

Design / CP	Intacto
1	100 000
2	88678
3	100 000
4	100 000
5	95890
6	100 000
7	100 000
8	100 000
9	100 000
10	99734
Mediana	100 000
Média	98430,2
Desv. Padrão	3659,202

Tabela 5. Valores originais do número de ciclos até a fratura para os corpos de prova fundidos em Ti cp e soldados a laser.

Design / CP	X	I06	I00
1	92372	100000	18922
2	88291	85938	20607
3	79887	100000	24334
4	100000	100000	11681
5	79249	100000	25776
6	86334	100000	34762
7	100000	93456	28990
8	100000	100000	21134
9	94532	75678	14765
10	100000	100000	28988
Mediana	93452	100000	22734
Média	92066,5	95507,2	22995,9
Desv. Padrão	8284,377	8368,399	7002,697

Tabela 6. Valores originais do número de ciclos até a fratura para os corpos de prova fundidos em Ti cp e soldados por meio de solda TIG.

Design / CP	X	I06	I00
1	67639	25307	13666
2	100000	26787	23909
3	100000	26347	12048
4	94987	21619	22455
5	100000	22976	14206
6	100000	31839	21057
7	88948	35525	31693
8	100000	25487	12398
9	98488	34576	17467
10	100000	30087	29287
Mediana	100000	26567	19262
Média	95006,2	28055	19818,6
Desv. Padrão	10260,05	4745,308	7041,156

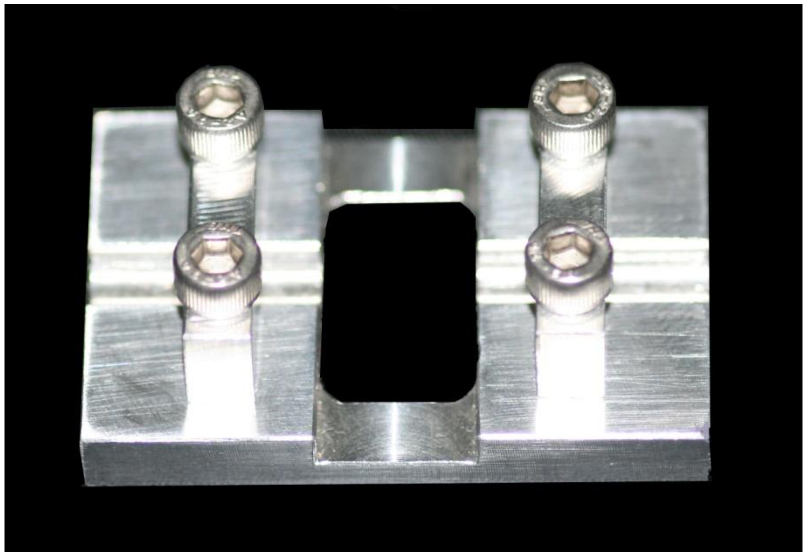

Figura 1. Dispositivo metálico utilizado para o alinhamento das partes a ser soldadas.

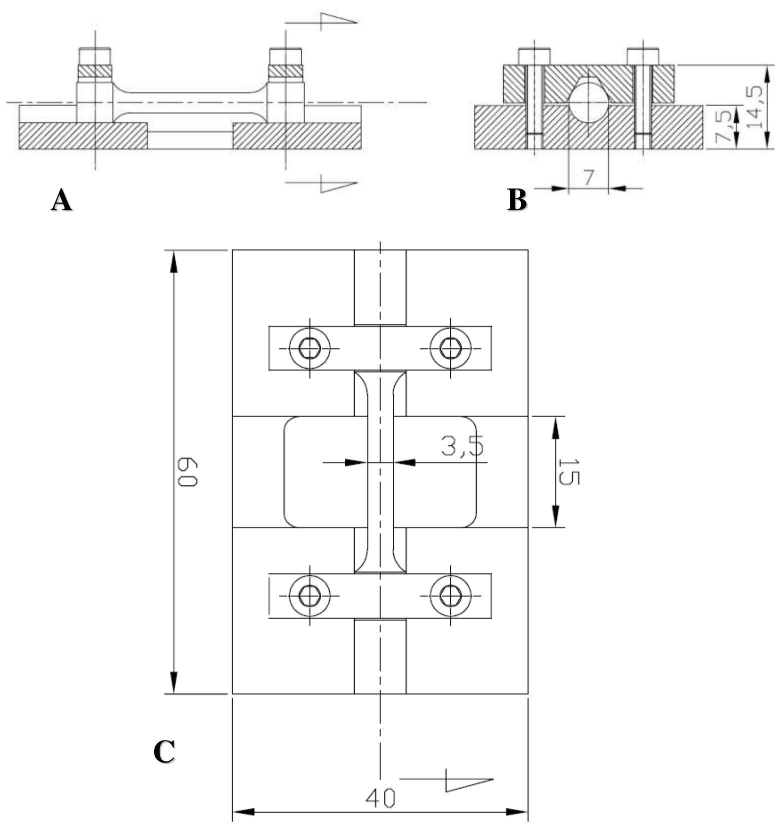

Figura 2: Dispositivo metálico para soldagem com o posicionamento do corpo de prova: **(A)** Vista frontal; **(B)** Vista lateral; **(C)** Vista superior

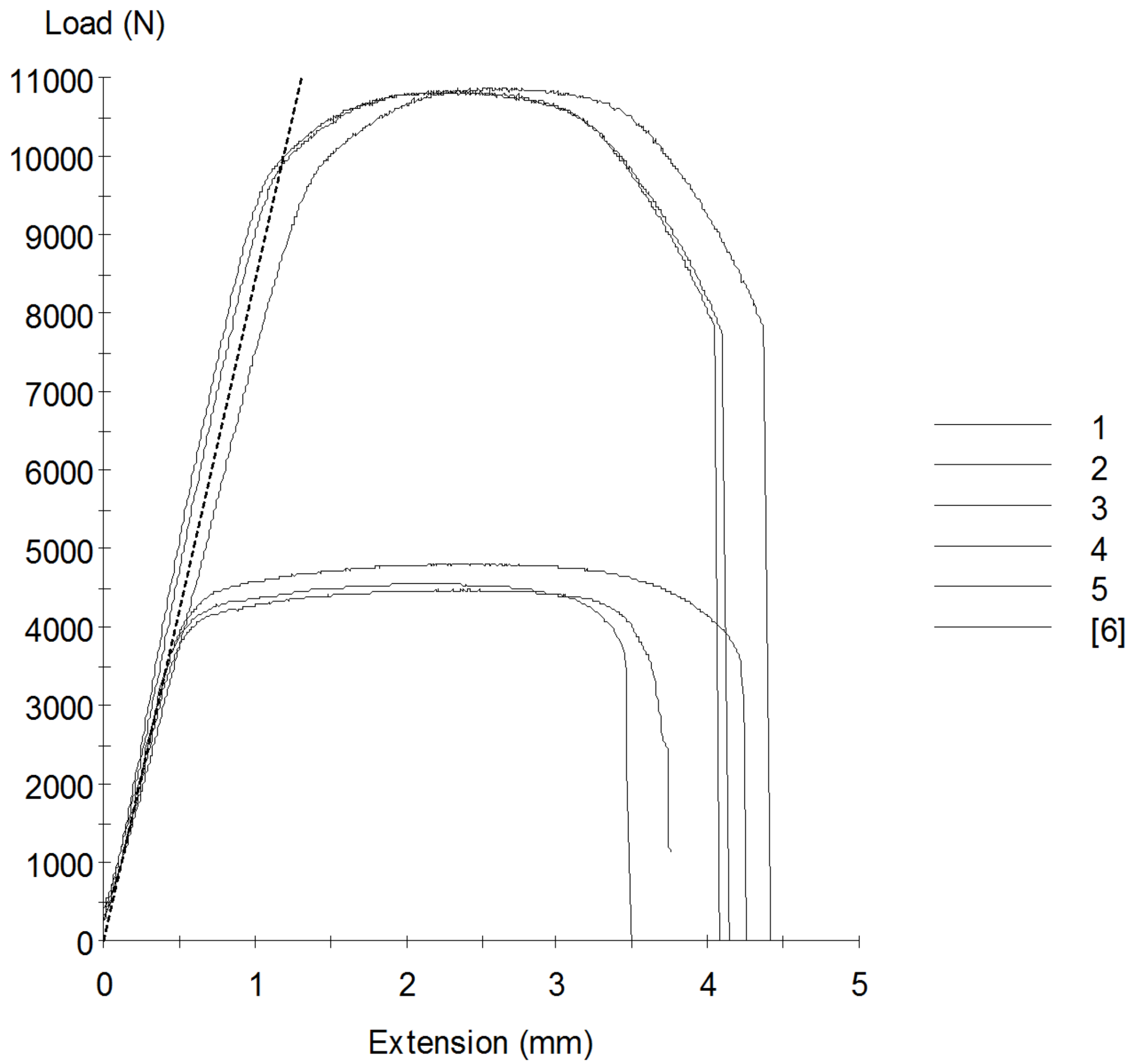

Gráfico 1. Curvas Força (N) x Deformação (mm) para os 6 corpos de prova tracionados. 1, 2 e 3) Corpos de prova usinados em liga de Ti-6Al-4V / 4, 5 e 6) Corpos de prova fundidos em Ti cp.

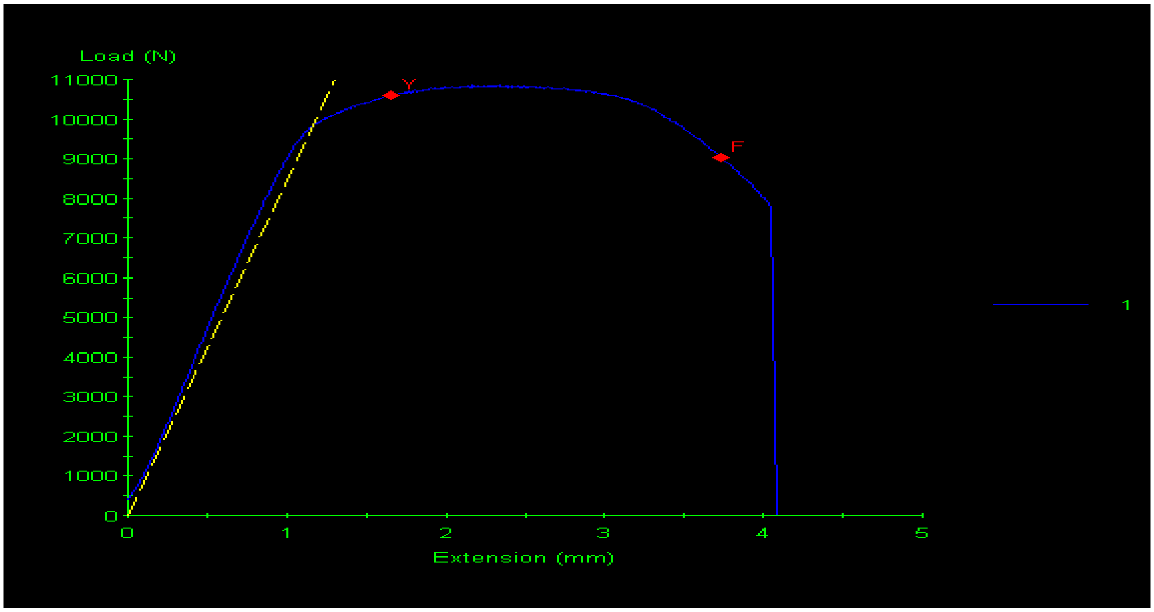

Gráfico 2. Curva Força (N) x Deformação (mm) do primeiro corpo de prova usinados em liga de Ti-6Al-4V, tracionado.

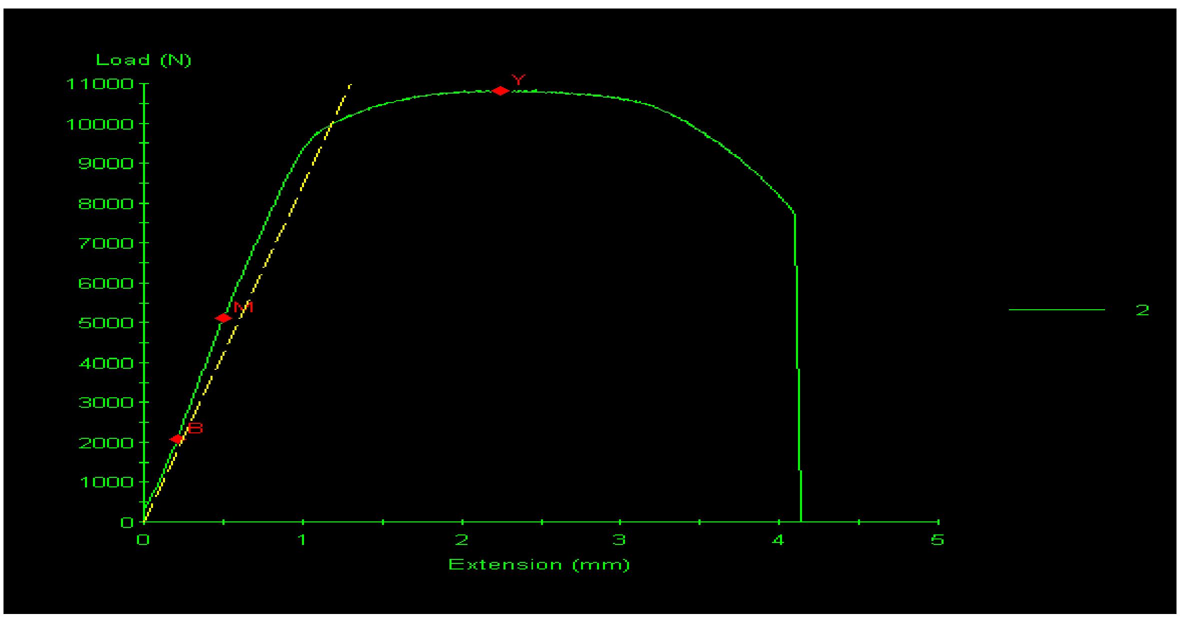

Gráfico 3. Curva Força (N) x Deformação (mm) do segundo corpo de prova usinados em liga de Ti-6Al-4V, tracionado.

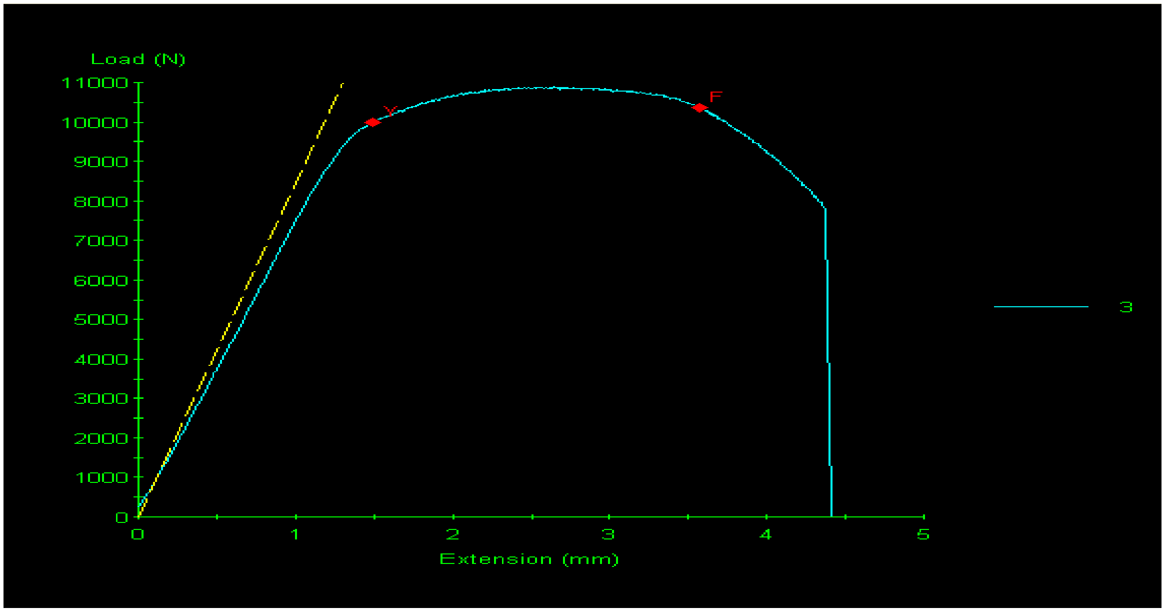

Gráfico 4. Curva Força (N) x Deformação (mm) do terceiro corpo de prova usinados em liga de Ti-6Al-4V, tracionado.

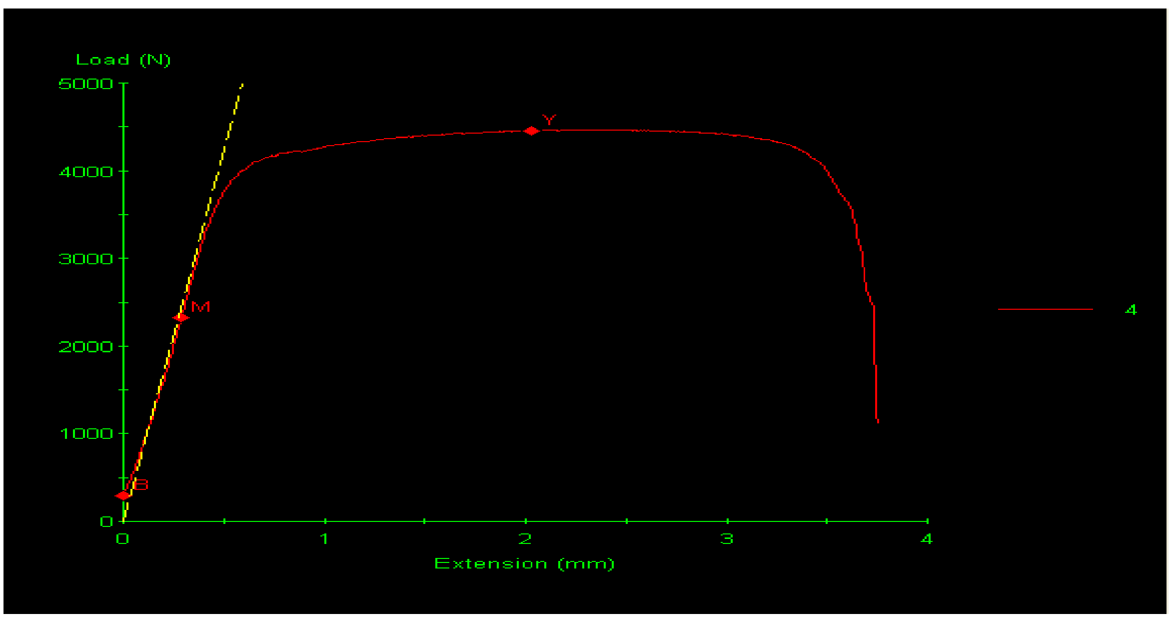

Gráfico 5. Curva Força (N) x Deformação (mm) do primeiro corpo de prova fundido em Ti cp, tracionado.

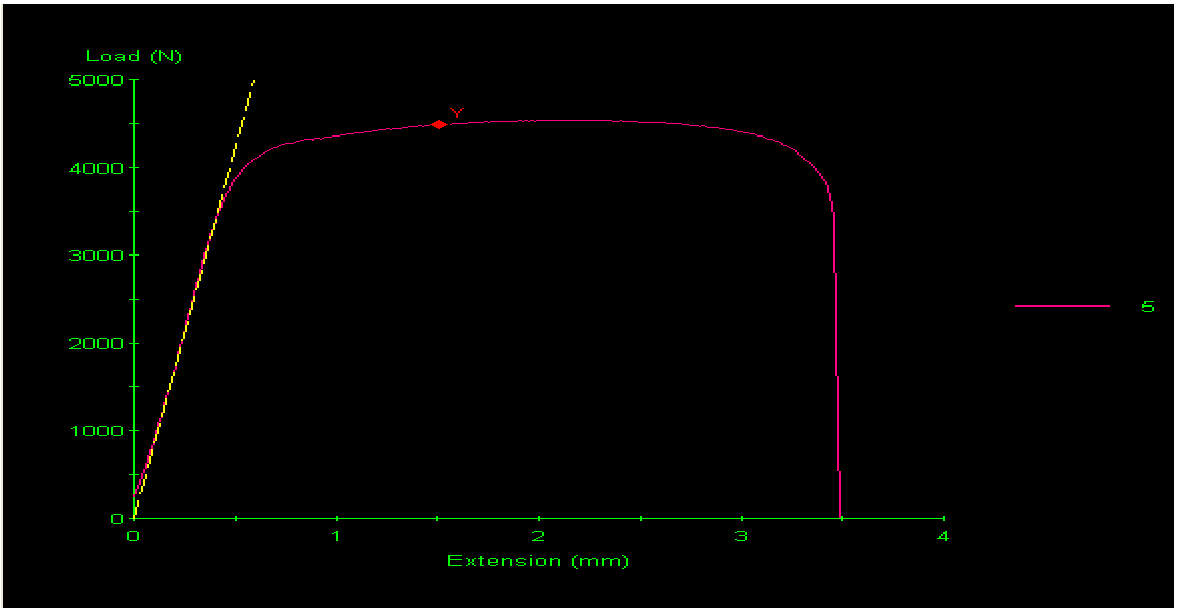

Gráfico 6. Curva Força (N) x Deformação (mm) do segundo corpo de prova fundido em Ti cp, tracionado.

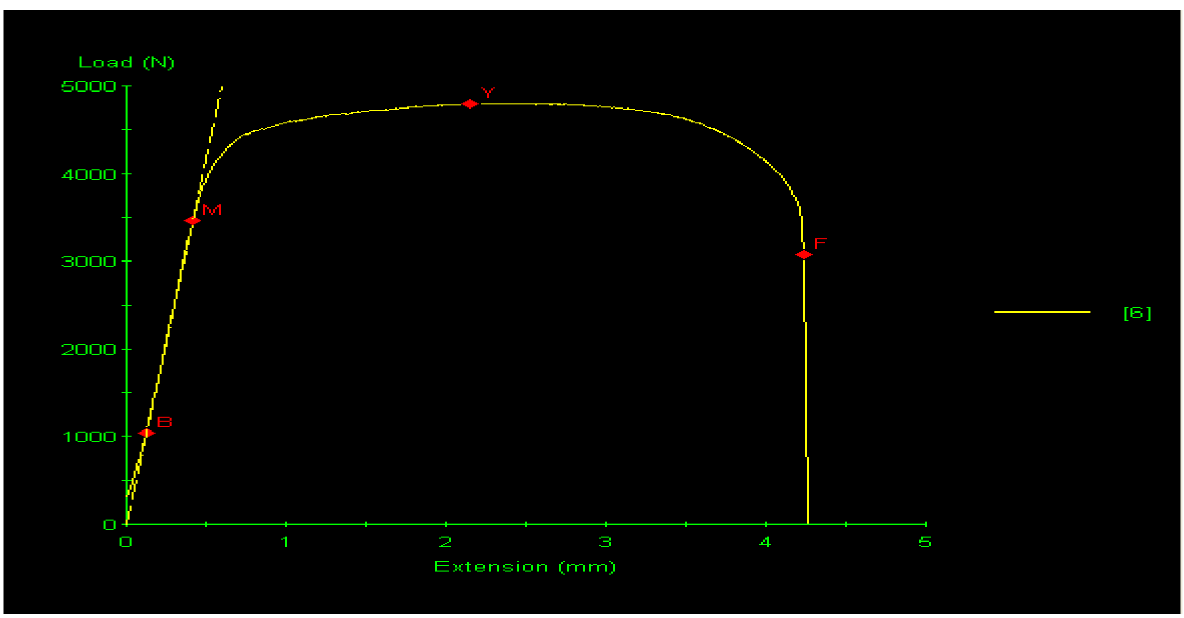

Gráfico 7. Curva Força (N) x Deformação (mm) do terceiro corpo de prova fundido em Ti cp, tracionado.