Universidade Federal do Rio de Janeiro

Centro de Ciências da Saúde

Faculdade de Odontologia

INFLUÊNCIA DA GEOMETRIA DE MINI-IMPLANTES ORTODÔNTICOS NA DISTRIBUIÇÃO DE TENSÃO PARA O OSSO CORTICAL. ESTUDO PELO MÉTODO DE ELEMENTOS FINITOS

Luiz Felipe de Miranda Costa

CD, MO

Tese submetida ao corpo docente da Faculdade de Odontologia da Universidade Federal do Rio de Janeiro – UFRJ, como parte dos requisitos, para a obtenção do Título de Doutor em Odontologia (Ortodontia).

Rio de Janeiro

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1. Mini-implantes 2. Estabilidade

3. Elementos finitos 4. Teses

I. Título

II. Tese (Doutorado - UFRJ/Faculdade de Odontologia)

“A persistência é o menor caminho do êxito”

Charles Chaplin

"Não desanimes. Persiste mais um tanto.
Não cultives o pessimismo.
Centraliza-te no bem a fazer.
Esquece as sugestões do medo destrutivo.
Segue adiante, mesmo varando
a sombra dos próprios erros.
Avança ainda que seja por entre lágrimas.
Trabalha constantemente. Edifica sempre.
Não consintas que o gelo do desencanto
te entorpeça o coração.
Não te impressiones à dificuldade.
Convence-te de que a vitória espiritual
é construção para o dia a dia.
Não desistas da paciência.
Não creias em realização sem esforço.
Silêncio para a injúria.
Olvido para o mal.
Perdão às ofensas.
Recorda que os agressores são doentes.
Não permitas que os irmãos desequilibrados te
destruam o trabalho ou te apaguem a esperança.
Não menosprezes o dever que a consciência
te impõe. Se te enganaste em algum trecho
do caminho, Reajusta a própria visão e
procura o rumo certo.
Não contes vantagens nem fracassos.
Estuda buscando aprender.
Não se voltes contra ninguém.
Não dramatizes provações ou problemas.
Conserva o hábito da oração para que
se te faça luz na vida íntima.
Resguarda-te em Deus e persevera no trabalho
que Deus te confiou.
Ama sempre, fazendo pelos outros
o melhor que possas realizar.
Age auxiliando. Serve sem apego.
E assim vencerás."

Francisco Cândido Xavier

Aos meus pais **LUIZ** e **SÔNIA**, meu irmão **GABRIEL** e minha namorada **ELLEN** pelo apoio constante, garantia de paz e amor verdadeiro.

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RESUMO

COSTA, Luiz Felipe de Miranda. **Influência da geometria de mini-implantes ortodônticos na distribuição de tensão para o osso cortical. Estudo pelo método de elementos finitos.** Orientador: Dr. Lincoln Issamu Nojima. Rio de Janeiro: UFRJ/Faculdade de Odontologia, 2015. Tese Doutorado em Odontologia – (Ortodontia). 116f

O objetivo desta pesquisa foi analisar como a variação de parâmetros geométricos de mini-implantes ortodônticos (MI), bem como a variação da espessura do osso cortical, podem afetar a estabilidade primária. A análise considerou o diâmetro e o comprimento da rosca do MI, e o comprimento do perfil transmucoso, que foram associados a oito diferentes espessuras do osso cortical. Considerando sua ampla aceitação na comunidade científica, a simulação pelo método de elementos finitos (MEF) foi adotada, possibilitando modelar diferentes características, especialmente quanto a espessura da cortical óssea. A distribuição de tensão no osso cortical adjacente ao MI e o deslocamento de sua cabeça foram definidos como critérios de avaliação. Menores magnitudes da tensão no osso cortical adjacente ou menores deslocamentos da cabeça do MI indicaram maior estabilidade. Os resultados demonstraram que as variações dos parâmetros escolhidos influenciam a tensão gerada no osso cortical e o deslocamento da cabeça do MI, portanto podem afetar sua estabilidade. De maneira geral, MI com perfis transmucosos maiores podem apresentar menor estabilidade. Houve relação positiva entre aumento do diâmetro da rosca e aumento da estabilidade. Observou-se influência positiva na estabilidade mediante aumento do comprimento da rosca de 6,0 para 8,0mm; para os demais comprimentos de rosca não se detectou aumento significativo na estabilidade.

SUMMARY

COSTA, Luiz Felipe de Miranda. **Influência da geometria de mini-implantes ortodônticos na distribuição de tensão para o osso cortical. Estudo pelo método de elementos finitos.** Orientador: Dr. Lincoln Issamu Nojima. Rio de Janeiro: UFRJ/Faculdade de Odontologia, 2015. Tese Doutorado em Odontologia – Ortodontia). 116f

The objective of this research was to analyze how the variation of orthodontic mini-implant (OMI) geometric parameters, as well as the variation of cortical bone thickness (CBT) may affect primary stability. The analysis considered OMI thread diameter and length, and transgingival collar (TC) length, which were associated with eight different CBTs. Considering its broad acceptance in the scientific community, simulation using the finite element method (FEM) was adopted, allowing the modelling of different features, especially for CBTs. Tension distribution in cortical bone adjacent to the OMI and displacement of its head were defined as evaluation criteria. Lower tension magnitudes in the adjacent cortical bone or smaller displacements of OMI head indicated greater stability. The results showed that variation of the chosen parameters influenced both the tension generated in the cortical bone and the displacement of OMI head, thus they may affect stability. In General, OMI with longer TCs may present lesser stability. There was a positive relation between

increase in thread diameter and increase in stability. A positive influence in stability was observed upon increasing thread length from 6.0 to 8.0mm; for the remaining thread lengths, a significant increase in stability was not detected.

LISTA DE SÍMBOLOS, SIGLAS E ABREVIATURAS

ν	coeficiente de poisson
CBT	cortical bone thickness
Ɛ	deformação
U	deslocamento
Ø	diâmetro, diameter
δ	displacement
C3D10	elementos tetraédricos com três dimensões e dez graus de liberdade
FEM	finite element method
INP	INP implantes, São Paulo-Brazil
MEF	método de elementos finitos
μm	micrômetro
mm	milímetro
MI	mini-implante
MPS	minimum principal stress
E	módulo de elasticidade
N	Newton
OMI	orthodontic mini-implant
PT	perfil transmucoso
%	porcentagem
ν	razão de poisson

RI	região de interesse
ROI	region of interest
SD	standard-deviation
σ	tensão
TPM	tensão principal mínima
TD	thread diameter
TL	thread length
TC	transgingival collar

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1 INTRODUÇÃO

A utilização de dispositivos temporários de ancoragem, incluindo mini-implantes (MI) e mini-placas é considerada rotina na clínica ortodôntica (Nojima *et al*., 2006). Os ortodontistas, ao planejarem casos envolvendo extrações dentárias com consequente fechamento de espaço, intrusões, ou a verticalização de molares, contam com a permanência destes dispositivos durante todo o tratamento.

Estudos clínicos e experimentais têm demonstrado que mini-implantes são capazes de oferecer ancoragem estável e suficiente para possibilitar a movimentação dentária durante o tratamento (Sawa *et al*., 2001; Park, 2001; Butcher *et al*., 2005). Os mini-implantes permitem fácil inserção em diferentes posições do osso alveolar, incluindo espaços limitados como a região inter-radicular (Motoyoshi *et al*., 2009). Seu tamanho diminuído proporciona maior aplicabilidade clínica, porém traz como consequência direta a atenuação de suas propriedades mecânicas (Lim *et al*., 2008; Pithon, Figueiredo e Oliveira, 2013).

Enquanto implantes dentários possuem altas taxas de sucesso (em torno de 90-95%) segundo Adell *et al.* (1981) e Albrektsson *et al.* (1981), mini-implantes falham mais frequentemente e não alcançam taxas tão altas, apesar de serem utilizados por períodos relativamente curtos (Butcher *et al*., 2005). A realidade clínica demonstra que os ortodontistas comumente experimentam situações de perda ou falha dos mini-implantes durante o tratamento, e embora taxas de sucesso em torno de 90% tenham sido relatadas (Sawa *et al.*, 2001; Park, 2001), insucessos tem sido observados na clínica diária (Brettin *et al.*, 2008; Reynders, Ronchi e Bipat, 2009).

O fator crítico para o sucesso ou falha de implantes ou mini-implantes é a distribuição da tensão gerada no osso adjacente (Geng *et al.*, 2004; Kayabasi *et al.*, 2006), cuja magnitude depende da carga aplicada, da interface osso-implante, da geometria do MI e também da qualidade e da quantidade ósseas (Huja *et al.*, 2005; Wilmes *et al.*, 2006; Wei *et al.*, 2011). Entende-se que menor tensão concentrada diminui a possibilidade de microfraturas ou de reabsorção óssea na interface osso-implante. Portanto, buscar situações nas quais a tensão gerada atue o mais uniformemente possível em todo o osso adjacente aumenta a estabilidade do MI (Geng *et al.*, 2004).

Estudos têm sugerido que os MI compostos por ligas de Titânio, apesar de apresentarem algum grau de osseointegração (Woods *et al.*, 2009), têm sua estabilidade primária baseada no embricamento mecânico da rosca com o osso alveolar (Brettin *et al.*, 2008; Jiang *et al.*, 2009). Com base nesta interação, tem-se recomendado a aplicação de carga imediata aos MI, o que elimina a necessidade de aguardar algum período para cicatrização (Liou, Pai e Lin, 2004; Liu *et al.*, 2012). Desta forma, seu potencial de ancoragem se torna bastante influenciável pela quantidade e qualidade do osso no local em que foi inserido (Brettin *et al.*, 2008; Martinelli *et al.*, 2010). Além disso, parâmetros que afetem o osso adjacente ao MI têm recebido destaque como fatores determinantes para a estabilidade, incluindo a porção extra-óssea, tais como o comprimento do perfil transmucoso (PT) ou da cabeça do MI (Huja *et al.*, 2005;

Motoyoshi *et al.*, 2005; Jiang *et al.*, 2009; Petrey *et al.*, 2010; Liu *et al.*, 2012; Duiabis *et al.*, 2012).

Objetivando aumentar a estabilidade, mudanças em diferentes parâmetros geométricos dos MI têm sidos propostas, incluindo alterações na porção extra óssea (Petrey *et al.*, 2010; Duaibis *et al.*, 2012; Liu *et al.*, 2012) e no comprimento e diâmetro da rosca (Song, Cha e Hwang, 2007; Kim *et al.*, 2009; Lim *et al.*, 2008; Jiang *et al.*, 2009; Petrey *et al.*, 2010; Duaibis *et al.*, 2012; Singh *et al.*, 2012; Chang *et al.*, 2012; Liu *et al.*, 2012; Duaibis *et al.*, 2012; Liu *et al.*, 2012; Pithon, Figueiredo e Oliveira, 2013).

A seleção do MI adequado a cada situação clínica demanda análise criteriosa da relação entre o diâmetro da rosca e o espaço interproximal. Além disso, o comprimento do perfil transmucoso é característica fundamental que frequentemente tem sido negligenciada. Sabe-se que o espaço interproximal é geralmente limitado, o que impossibilita o uso rotineiro de mini-implantes mais largos, e que a espessura gengival é variável e interfere na definição do comprimento do PT. Entender como mini-implantes com PT de tamanhos diferentes se comportam mediante a mesma carga e a possível compensação em usar mini-implantes com roscas de maior comprimento em comparação com roscas de menor diâmetro é extremamente importante. A partir de tais questões, este trabalho objetiva analisar a influência de parâmetros físicos básicos dos mini-implantes com relação à estabilidade primária. A metodologia empregada inclui o método de elementos finitos, admitindo a hipótese de que a maior magnitude de tensão transmitida ao osso e o maior o deslocamento da cabeça do mini-implante aumentam as chances de insucesso. Além disso, planejou-se estudar como a variação da espessura do osso cortical afeta a distribuição da tensão gerada em cenário sem osseointegração, como observado na aplicação de carga imediata.

2 PROPOSIÇÃO

Objetivos gerais:

Realizar simulação computacional pelo método de elementos finitos (MEF) de sistema osso-mini-implante, caracterizado por cenário sem osseointegração, com carga imediata e considerando diferentes espessuras do tecido ósseo cortical.

Objetivos específicos:

2.1. Analisar o efeito da variação do comprimento do perfil trasmucoso na magnitude da tensão transmitida ao osso cortical adjacente e no deslocamento da cabeça do mini-implante;

2.2. Analisar o efeito da variação do diâmetro da rosca sobre a magnitude da tensão transmitida ao osso cortical adjacente e sobre o deslocamento da cabeça do mini-implante;

2.3. Analisar o efeito da variação do comprimento da rosca sobre a magnitude da tensão transmitida ao osso cortical adjacente e sobre o deslocamento da cabeça do mini-implante em cenário caracterizado por diferentes valores do módulo de elasticidade (ME) do osso medular adjacente.

3 DELINEAMENTO DA PESQUISA

Para realizar a simulação computacional pelo MEF, o sistema osso-implante com inserção monocortical foi modelado virtualmente, contendo um mini-implante inserido em bloco representativo dos tecidos ósseos cortical e medular (Figura 1).

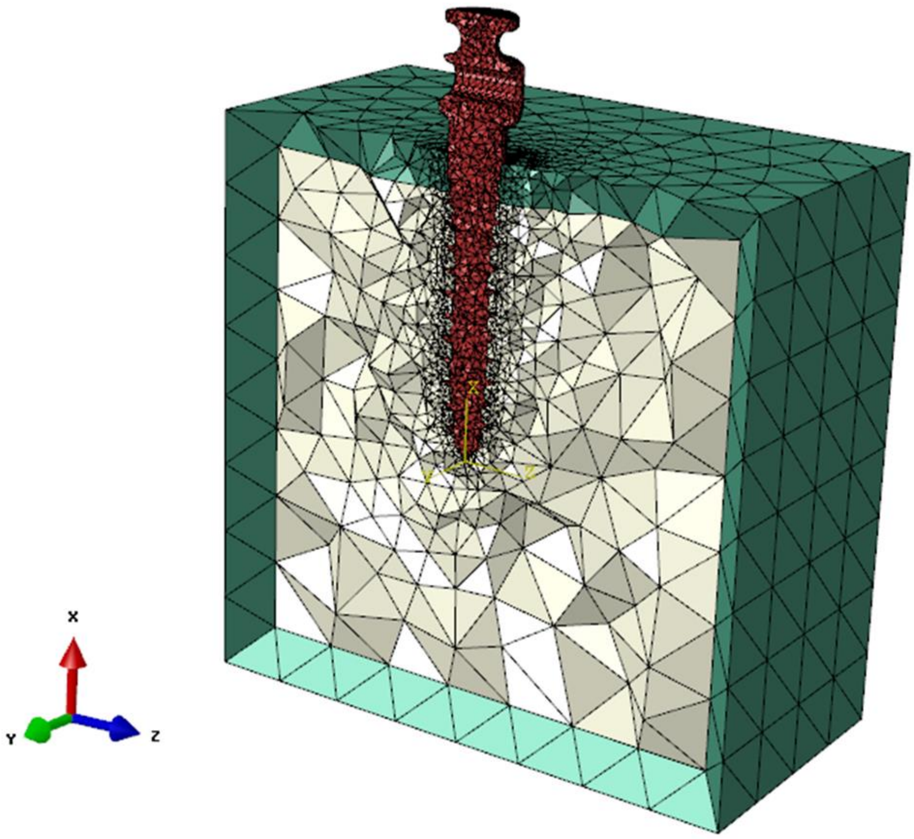

Figura 1. Vista em corte transversal do modelo de elementos finitos tridimensional do sistema osso-implante. Mini-implante (vermelho), osso cortical (verde), e osso medular (cinza).

Sistema Osso-Mini-Implante

Tecidos ósseos Cortical e Medular

O bloco representativo dos tecidos ósseos foi gerado com o *software* comercial SolidWorks (*Dassault Systems Simulia Corp, Providence, RI*), em forma de cubos interpostos com 20 mm de aresta (Figura 2). A camada externa de osso cortical, com espessura variável entre 0,25 a 2,0 mm, com intervalos de 0,25 mm, envolveu o núcleo complementar de osso medular.

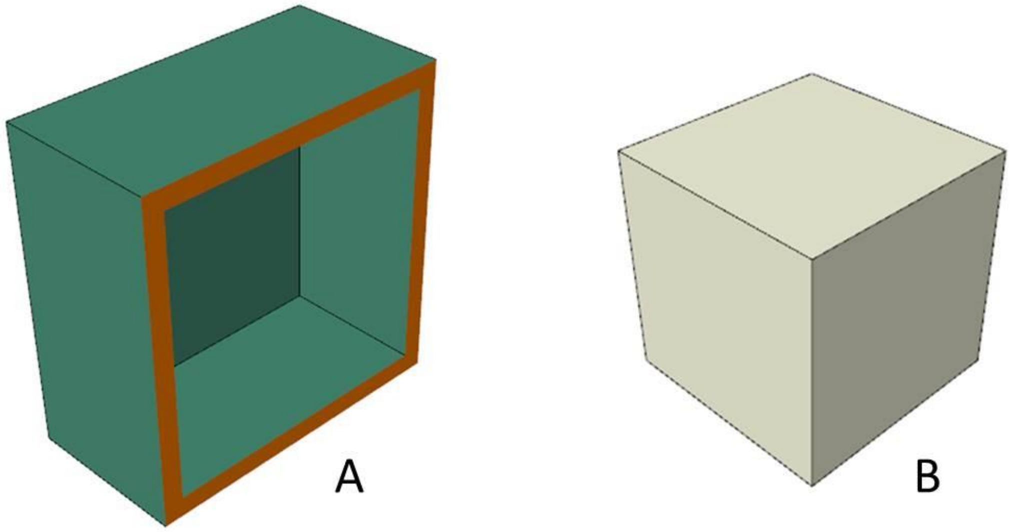

Figura 2. Ilustração mostrando os blocos cúbicos representando os tecidos ósseos cortical (A) e medular (B). Notar que o bloco representando o osso cortical é oco.

Os blocos foram unidos, conservando as propriedades mecânicas dos dois tecidos ósseos; o osso medular na camada externa, oca, de espessura variável, e o osso cortical no núcleo complementar. Tal combinação estrutural mantém os blocos unidos, porém com propriedades mecânicas diferentes.

Mini-Implantes

Mini-implantes auto-perfurantes (Figura 3), desenvolvidos pela INP (INP Implantes, São Paulo-Brasil) foram gerados com o *software* comercial SolidWorks (*Dassault Systems Simulia Corp, Providence, RI*).

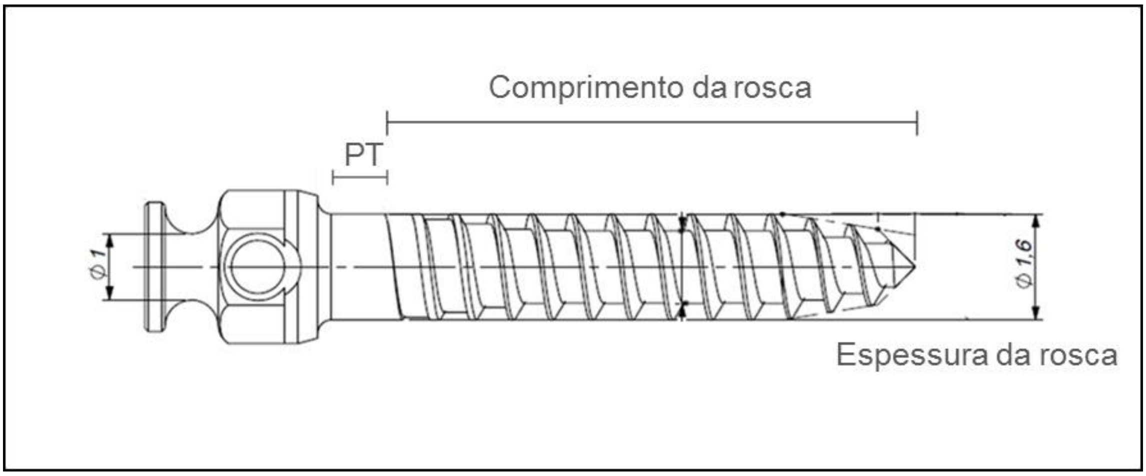

Figura 3. Desenho esquemático de mini-implante ortodôntico modelado evidenciando os parâmetros físicos analisados, comprimento e espessura da rosca e perfil transmucoso (PT).

Os mini-implantes utilizados neste estudo apresentam alguns parâmetros geométricos básicos em comum, incluindo filetes com altura de 0,2 mm e passo com 0,6 mm de comprimento. No primeiro trabalho, que analisou a variação do perfil transmucoso, todos os MI possuíam rosca com 8,0 mm de comprimento e 1,6 mm de diâmetro, e o comprimento dos perfis transmucosos variou de 1 a 4 mm. No segundo trabalho, analisando a variação do diâmetro da rosca, todos os MI possuíam perfil transmucoso com 1,0 mm e rosca com 8,0 mm de comprimento, e os diâmetros analisados foram 1,2; 1,4; 1,6; 1,8 e 2,0 mm. Finalmente, no tercero trabalho, onde foi analisad a variação do comprimento da rosca, estas possuíam diâmetro de 1,6 mm e perfil transmucoso com 1.0 mm e porção extra-óssea com comprimento total de 3,5 mm. Os comprimentos de rosca analisados (porção intra-óssea) foram 6; 8; 10 e 12 mm.

Propriedades Mecânicas

O material considerado para os MI foi a liga Ti-6Al-4V e as propriedades adotadas para os modelos consideram-nos isotrópicos, homogêneos e linearmente elásticos. As propriedades mecânicas foram baseadas em dados publicados (Field *et al.*, 2009; Collings, 1984) e são apresentadas na Tabela 1.

Tabela 1. Propriedades mecânicas dos materiais do modelo geométrico.

Material	Módulo de Elasticidade (MPa)	Coeficiente de Poisson	Referências
Liga de Ti-6Al-4V	110.000	0,33	Collings,1984
Osso Cortical	14.700	0,30	Field *et al.* 2009
Osso Medular	490	0,30	Field *et al.* 2009

Módulo de Elasticidade

O **módulo de elasticidade (E)** ou **módulo de Young**, é a medida da rigidez de material sólido (Callister, 2007), propriedade intrínseca de cada material, dependente da composição química, microestrutura e descontinuidade (poros e trincas). Pode ser obtido pela razão entre a tensão exercida sobre o material e a deformação sofrida por este. A **Tensão** corresponde a força ou carga por unidade de área, e a **deformação** representa a mudança em determinada dimensão, por unidade da dimensão original. Assim, o módulo de elasticidade é dado pela fórmula:

$$E = \frac{\sigma}{\varepsilon}$$

onde:

E = módulo de elasticidade, medido em Megapascal [MPa]

σ = tensão aplicada, medida em Pascal [N/m^2]

ε = deformação elástica longitudinal (adimensional)

Coeficiente de Poisson

O **Coeficiente de Poisson (V)** é a relação entre as deformações transversais relativas e a deformação longitudinal relativa para material homogêneo e isotrópico (Morrel, 1996). É dado pela fórmula:

$$\nu = -\frac{\epsilon_x}{\epsilon_z} = -\frac{\epsilon_y}{\epsilon_z}$$

onde:

v = Coeficiente de Poisson (adimensional)

ε_x = Deformação relativa na direção X

ε_y = Deformação relativa na direção Y

ε_z = Deformação relativa na direção Z

Cenário de Análise

O cenário para análise foi montado com a interposição do mini-implante no bloco que representa o tecido ósseo, perpenducularmente à superfície. A perfuração da rosca foi gerada utilizando subtração *Booleana* (Figura 4).

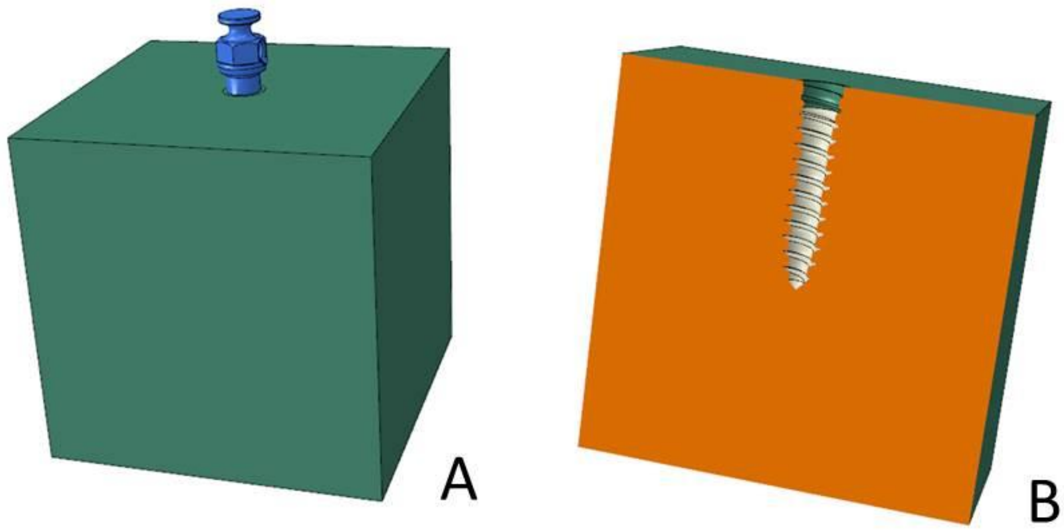

Figura 4. Cenário montado para a análise, com o mini-implante inserido perpendicularmente ao osso (A). Notar a perfuração da rosca gerada no osso utilizando subtração Booleana (B).

Critérios de Interação

A simulação buscou a situação mais próxima possível da realidade, avaliando a relação de contato entre o MI e o tecido ósseo. Na carga imediata, a rosca está inserida no osso por meio de embricamento mecânico, sem osseointegração, caracterizando estabilidade primária. O comportamento interfacial das superfícies foi representado por coeficiente de atrito estático de *Coulomb* de 0,30 (Jiang *et al.*, 2009), correspondente à textura superficial dos MI. O modelo foi construído para efeito de análise estática, sendo o processo de inserção do mini-implante (análise dinâmica) desconsiderado para o presente estudo. Portanto, não há tensão no tecido ósseo decorrente da inserção do mini-implante, ocorrendo somente após aplicação de carga. A interface osso-implante foi modelada com contato total, assumido como critério de equivalência para melhor representar a condição real entre o osso e o MI.

Condições de Contorno e Carregamento

Os modelos foram engastados na face oposta à de inserção do mini-implante, considerando a característica estrutural adotada. Um carregamento de 2 N foi aplicado na cabeça do mini-implante, com vetor paralelo à superfície do osso. Desta forma foi possível simular o mini-implante utilizado como fonte de ancoragem direta (Figura 5).

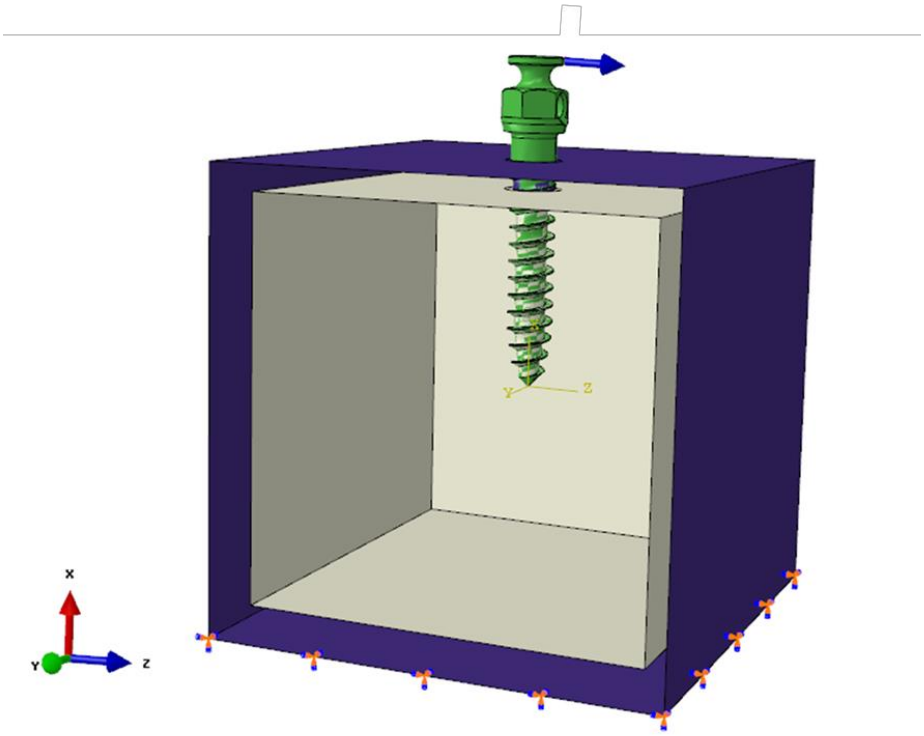

Figura 5. Engaste do bloco que representa o tecido ósseo na face oposta à de inserção do mini-implante. Notar a carga aplicada na cabeça do mini-implante (seta azul).

Descrição da Malha de Elementos Finitos

O modelo em elementos finitos foi criado utilizando elementos tetraédricos com três dimensões e dez graus de liberdade (C3D10). Objetivando apresentar resultados coerentes, mas com custo computacional aceitável, um estudo piloto de sensibilidade foi realizado para determinar a malha global mais adequada à análise. A malha foi refinada na região da rosca gerada no tecido ósseo. A dimensão global típica dos elementos para o tecido ósseo foi 2,0 mm. O refinamento na região da rosca foi estabelecido com dimensão de 0,15 mm (Figura 6). No parafuso foram utilizados elementos com dimensão de 0,2 mm. De maneira geral, os modelos completos continham cerca de 75.000 elementos e 110.000 nós, sendo 55.000 elementos e 80.000 nós no tecido ósseo, e 20.000 elementos e 30.000 nós no mini-implante.

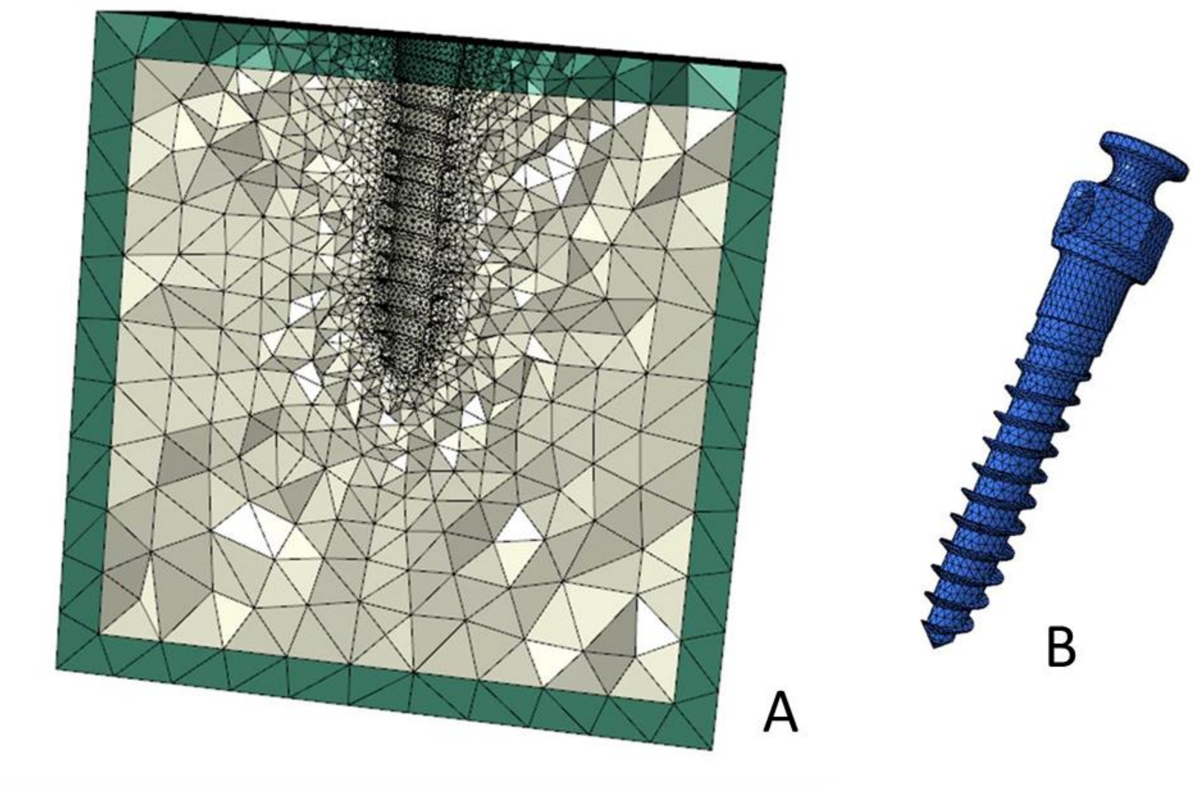

Figura 6. Desenho da malha em elementos finitos no tecido ósseo (A) e no mini-implante (B). Notar o refinamento da malha na região da rosca gerada no tecido ósseo.

Simulação

Os modelos geométricos virtuais foram convertidos em elementos finitos. Todos as malhas foram geradas e analisadas com o *software* comercial Abaqus 6.12 (*Dassault Systems Simulia Corp, Providence, RI, U.S.A.*). A tensão principal mínima (TPM) foi selecionada porque proporciona a análise de tensão possibilitando a distinção entre regiões de tração e compressão.

Valores negativos representam áreas de compressão e valores positivos representam áreas de tração.

A região externa do osso cortical, adjacente ao mini-implante, foi escolhida como região de interesse (RI) para análise de tensão. Nesta região foi definido disco circular com 1,0mm ao redor da área de contato, concordando com autores que avaliaram sua função na estabilização do MI (Motoyoshi *et al.*, 2009; Wei *et al.*, 2011; Singh *et al.*, 2012; Duaibis *et al.*, 2012) e que afirmaram que este é o local em que geralmente se concentram os picos de tensão após o carregamento (Liu *et al.*, 2012). Considerando que a tensão transferida ao osso medular é mínima (Singh *et al.*, 2012; Liu *et al.*, 2012), esta não foi analisada.

Como critério adicional à análise, também foi observado o deslocamento da cabeça do mini-implante. Este tipo de deslocamento durante o tratamento já foi relatado (Liou, Pai *e* Lin, 2004; Alves Jr. *et al.*, 2011) e correlacionado à diminuição da estabilidade (Jiang *et al.*, 2009; Brunski *et al.*, 1999), e também analisado recentemente por outros autores (Liu *et al.*, 2012; Singh *et al.*, 2012).

4 DESENVOLVIMENTO DA PESQUISA

4.1 ARTIGO 1

Costa LFM, Nojima LI, Santos CE, Cimini Júnior CA, Bortoleto E, Santiago RC. Evaluation of orthodontic mini-implant transgingival collar length on stress transmission to cortical bone: A three-dimensional finite element analysis.

A ser submetido ao **American Journal Of Orthodontics and Dentofacial Orthopedics**

4.2 ARTIGO 2

Costa LFM, Nojima LI, Santos CE, Cimini Júnior CA, Bortoleto E, Santiago RC. Effect of orthodontic mini-implant thread diameter in stress distribution in the adjacent cortical bone: a finite element analysis.

A ser submetido ao **American Journal Of Orthodontics and Dentofacial Orthopedics**

4.3 ARTIGO 3

Costa LFM, Nojima LI, Santos CE, Cimini Júnior CA, Bortoleto E, Santiago RC. Influence of orthodontic mini-implant thread length and bone mechanical properties on stress transmission to cortical bone: A three-dimensional finite element analysis.

A ser submetido ao **American Journal Of Orthodontics and Dentofacial Orthopedics**

ARTIGO 1

Evaluation of orthodontic mini-implant transgingival collar length on stress transmission to cortical bone: A three-dimensional finite element analysis

Costa LFM, Nojima LI, Santos CE, Cimini Júnior CA, Bortoleto E, Santiago RC.

ABSTRACT

Introduction: Orthodontic mini-implants (OMI) manufacturers offer many design options for thread and head geometry, as well as transgingival collar (TC) length. Studies have investigated mechanical performance considering thread geometry variation in detail, yet little is known about how TC length variation affects stability. **Methods:** Numerical simulation by means of the Finite Element Method (FEM), analyzing the minimum principal stress in a bone-implant model. Four scenarios were considered with TC lengths of 1, 2, 3 and 4 mm. To each length, 8 different cortical bone thicknesses (CBTs) were assigned, from 0.25 mm to 2.0 mm, at increments of 0.25 mm. As an additional criterion, displacement of the OMI head was observed as well. **Results:** Association between the increase in TC length and the increase in tension magnitude was identified. Keeping the 1 mm TC as a reference, as the length progressed to 2, 3 and 4 mm, an increase in tension magnitude in the adjacent bone was observed up to 29.66%, 49.17% and 78.11%, respectively. **Conclusions:** Increasing TC length also increased tension magnitude in the adjacent bone. This became more noticeable in slender cortical bones. Increasing CBT decrease the minimum principal stress in the region of interest, regardless of TC length.

INTRODUCTION

The use of temporary anchorage devices, including orthdontic mini-implants (OMI) and miniplates, is common in orthodontics.[1] For planning of cases involving tooth extraction and subsequent space closure, intrusion or molar uprighting, orthodontists require the stability of these devices during the active phase of movement. However, this is not the clinical reality, since stability loss has been observed,[2,3] even though OMI present high rates of success during treatment.[4,5]

The critical factor for success or failure of OMI is the stress distribution of tension generated in the surrounding bone,[6,7] whose magnitude depends on applied load, bone-implant interface, OMI geometry, as well as bone quality and quantity.[8-10] Less concentrated tension is understood to decrease the possibility of microdamage or bone resorption at bone-implant interface; thus seeking situations in which the generated tension acts as evenly as possible across the surrounding bone increases stability.[6]

Studies have suggested that OMI of certain alloys can show partial osseointegration,[11] although primary stability is due to mechanical interlock between the screw thread and the alveolar bone.[2,12] Based on such interaction, applying immediate load to OMI has been recommended, eliminating the need of waiting for some healing period.[13,14] This way, anchor potential becomes quite influenced by bone quantity and quality at the insertion location.[2,15] Several authors agree[8,12,16,17] and some emphasize that, besides the parameters mentioned, others are as important, among which the lengths of both transgingival collar (TC) and OMI head, which can be essential to determine stability.[14,18,19,20]

Selection of OMI suitable to each clinical situation demands careful assessment of the relation between diameter and inter-radicular space, as well as between cortical bone thickness and quality. Additionally, the TC length is a fundamental characteristic often disregarded by most orthodontists.

Knowing gingival thickness is variable and interferes in choosing TC length, it is important to understand how OMI with different transgingival collar lengths perform under the same load. In accordance with this proposition, the present study aims at employing the finite element method (FEM) to analyze how the increase in TC length influences stress distribution in the surrounding bone, in addition to evaluating stress magnitude in different scenarios considering cortical bone thickness (CBT) variation, no osseointegration and immediate load.

MATERIALS AND METHODS

Four three-dimensional geometric models simulating a monocortical insertion bone-implant system were created with OMI TC lengths of 1, 2, 3 and 4 mm, and CBT varying from 0.25 to 2.0 mm, at increments of 0.25 mm (Figure

1). OMI material was the Ti-6Al-4V Grade 5 alloy, and model properties were homogeneous, isotropic and linear elastic [21,22] (Table I).

Four self-tapping OMI (Figure 2) designed by INP (*INP implantes, São Paulo-Brazil*) were modeled using SolidWorks commercial software (*Dassault Systemes Simulia Corp., Providence, RI, USA*) with 8.0 mm thread length, 1.0 mm major diameter, 0.2 mm fillet height and 0.6 mm pitch.

Cortical and cancellous bones were modeled as interposed cubic blocks with 20 mm edge. This part merged the internal compact block with mechanical properties of cancellous bone and the hollow outer layer of variable thickness with mechanical properties of cortical bone, whose structural combination remains thorough, yet maintaining different mechanical properties.

The scenario for analysis was set interposing the OMI in the cube representing the bone perpendicularly to the surface. The thread was created in the bone model by means of Boolean subtraction.

The simulation characterized a scenario as real as possible, considering the OMI-bone contact relation. For immediate load, the thread is enclosed inside the bone via mechanical interlock and there is no osseointegration, exemplifying primary stability. Surface interaction was represented by the Coulomb static friction coefficient of 0.30,[12] corresponding to the OMI superficial texture.

Models were encastered on the face opposite to the OMI and a 2N load was applied to the OMI head in a vector parallel to the bone model surface; simulating direct anchor (Figure 3).

Three-dimensional, ten degree-of-freedom tetrahedral elements were used (C3D10). A preliminary sensitivity study was conducted to determine the most appropriate global mesh, targeting consistent results at an acceptable computational cost. Following this study, the criterion to refine the mesh region surrounding the OMI was defined, creating a circular partition 1.0 mm further than the bone-implant contact area.

Meshes were generated and simulated with Abaqus 6.12 commercial software (*Dassault Systemes Simulia Corp., Providence, RI, USA*). The resulting minimum principal stress was selected because it provides an analysis with distinction of compression and traction regions, generating negative values for compression and positive values for traction. The external portion of the cortical bone adjacent to the OMI on the compression side was set as the region of interest (ROI) because this is where peak stress usually occurs after loading.[14] Understanding the stress transferred to the cancellous bone is minimal,[14,23] it was not analyzed.

As an additional criterion for mechanical performance analysis, displacement of the OMI head was observed as well. Such fact has been reported [13,24] and correlated with decrease in stability.[12,16,25]

RESULTS

In all models, regardless of TC length or CBT, compression areas were observed in the ROI and traction areas were observed on the opposite side (Figure 4).

This study considered that OMI stability increases with lower magnitude of tension transmitted to the cortical bone, as well as with smaller displacements of the OMI head.

Figure 5 shows the magnitude of tension was higher in slender cortical bone regions, lowering as CBT increased; and Figure 6 shows stress distribution comparing all four TC lengths analyzed.

The association between the increase in TC length and the increase in tension magnitude was identified in the ROI (Table II). Keeping the 1.0 mm TC as a reference, as the length progressed to 2, 3 and 4 mm, an increase in tension magnitude in the adjacent bone was observed up to 29.66%, 49.17% and 78.11%, respectively. Such increase became more evident in slender cortical bone regions, lessening as the CBT increased.

CBT increase presented direct relation with peak stress decrease in the ROI (Table II). This relation was evident in OMI with longer TC lengths (3 and 4 mm). In the OMI with 1 mm TC, it is possible to observe the tendency of tension magnitude to remain relatively similar from 1.0 mm CBT on. The 2 mm TC can be described as transitional, exhibiting intermediate characteristics (Figure 5).

Variation both in TC length and in CBT affect OMI head displacement. TC length increase favored displacement, especially for slender CBT (Figure 7), and CBT increase hindered displacement for all TC lengths (Table III).

Displacements observed for extreme CBT (0.25 and 2.0 mm) are quite different (Figure 7), especially in 3 and 4 mm TC lengths; in OMI with shorter TC lengths, such difference is less significant. In the OMI with 1 mm TC, CBT increase affects displacement, with minimum impact from 1.0 mm CBT on. For the remaining TC lengths, the same tendency occurred from 1.5 mm CBT on.

DISCUSSION

Following the criterion adopted, less tension in the cortical bone suggests increase in stability. Tension concentration is likely to cause microdamage and resorption, with subsequent loss of support, [6,9,26,27,28] as overloading is the highest risk factor for implant failure, either conventional or OMI. [29]

Studies using numerical simulations asserted that cortical bone resorption around the thread occurred in the region of high compressive stress over -50 MPa.[30,31] The results in this study showed that tension magnitude in the ROI ranged from 7.7 to 21.5 MPa (von Mises). Such values were lower than those recorded by other authors, including Motoyoshi *et al,* [32] who observed variation from -31 to -55 MPa evaluating load direction and CBT; and Li *et al,* [33] who determined the critical stress curve for bone resorption in high-density areas exceeding 25 MPa. However, the values in this study were higher than those found by Singh *et al,* [23] who identified peak stress of 6 MPa for a load parallel to the bone and 8.5 MPa for a torsional load.

The differences in results can be interpreted based on specific scenario variations in each study, including different force magnitudes, OMI dimensions, material properties, existence of osseointegration, as well as the type of tension analyzed.

Immediate load applied to OMI has been recommended to reduce treatment duration[12], being routine among orthodontists. Thus, knowing in detail the factors which influence primary stability is also important. Although the literature has already discussed the relation between CBT variation and OMI stability, greater emphasis was given to it in simulations considering osseointegration.[2,6,34] This work succeeded at effectively representing the immediate load condition to provide a more suitable analysis.

A recent study [20] states that clinicians should use OMI with the largest diameters and the shortest head lengths possible. Such statement was made after observing factors influencing tension magnitude in the bone around the OMI, evaluating diameter, lengths of the head and the thread, as well as the elastic modulus of cancellous bone. This fact is corroborated by studies reporting that larger lever arms created by the increased distance between the OMI head and the adjacent bone can reduce OMI stability.[18,19,20,35,36] Moreover, a recent study claims the outer portion of the bone is the factor that can really influence its performance [14] The results attained by the present study agree with the latter assertions.

For all models, stress distribution was represented similarly and according to physical intuition, as mentioned by Lombardo *et al.* [37] Figure 3 shows the tendency of OMI to tip in the same direction of the force, generating compression zones in the ROI and traction zones in the opposite region. This result validated simulation of the OMI-bone mechanical interlock effect, without osseointegration. Simulations considering structurally-united model components or disregarding contact between surfaces do not represent the mechanical interlock which occurs at the post-insertion stage; instead, they reproduce the secondary stability resulting from osseointegration.

Therefore, at the initial post-insertion stages, the major stability factor is the existing mechanical interaction between OMI fillets and the thread created in the bone. A frictional condition establishing this relation is essential, and it was assigned between the corresponding surfaces, as described by Jiang *et al*.[12] The Coulomb static friction coefficient of 0.30 was used, a value greater than the ones used by other authors,[20,29,37] in order to allow sliding and/or separation between the surfaces in the interlock region, in accordance with the analysis criterion adopted.

Studies have examined the bone-implant interface with similar methodologies. Among these, some simulating non-osseointegrated OMI,[12,29,37,38] and others considering osseointegration.[32,34] However, only the present study evaluated CBT variation in a scenario with no osseointegration.

Observation of mechanical performance resulting from the use of different TC lengths was made by means of comparison among magnitude and distribution of tension in the adjacent bone, as well as OMI head displacement. The minimum principal stress was chosen for evaluation because it is suitable to identify compression and traction areas, considering the effect resulting from load application.[39,40,41] Although minimum displacements do not necessarily represent failure,[24] OMI head displacement was also used as a supporting parameter in stability analysis by other authors.[14,23]

A relation between the increase in TC length and the increase in magnitude of the tension generated in the ROI was observed. Such result was expected because larger lever arms created by the increased distance between the force application point and bone surface generate greater moments and subsequent tensions in the adjacent bone. [18,19,20,39] More important than simply proving the positive relation between the increase in TC length and the increase in tension, this study pursued to substantiate the understanding of how CBT may affect this phenomenon.

Tension magnitude in the ROI exhibited the slightest difference for extreme CBT (0.25 and 2.0 mm) in OMI with TC lengths of 1, 2 and 3 mm, which significantly increased in OMI with TC length of 4 mm. OMI with TC lengths of 3 and 4 mm are not common, and would only be indicated in cases of considerable gingival thickness, as in the palate, and for suitable clinical use they should be made with larger diameters or with longer threads.

Analyzing mechanical performance based on CBT influence, it becomes noticeable that CBT increase favors mechanical interlock and reduces peak stress for all TC lengths. This fact is represented in Figure 6, particularly in OMI with longer TC lengths (3 and 4 mm).

In OMI with TC length of 1 mm, it is possible to observe the tendency of tension magnitude to remain relatively similar from 1.0 mm CBT on. Analyzing displacement, there is an actual tendency of CBT increase to influence it minimally, in particular from 1.0 mm CBT on. For the other TC lengths, this tendency occurred from 1.5 mm CBT on. A recent study [32] also using FEM and analyzing CBT influence in OMI stability identified the same tendency of tension magnitude stabilization from 1.0mm CBT on, however in a scenario considering osseointegration; displacement was not analyzed.

Even though the results reveal the importance of CBT as a critical factor for OMI stability, it should be emphasized that CBT does not exclusively explain why some OMI are more stable. The ability of the bone to endure certain tension levels without reabsorption or microdamage also results of its quality. Such consideration originates from clinical studies [16,32] comparing stability of implants and OMI inserted in the maxilla and in the mandible, investigating the success rate, which did not observe a significant difference between these bone structures, though the average CBT in the maxilla was 1.0/2.0 mm and in the mandible 2.0/3.0 mm. Furthermore, Huja *et al* [8], agree and acknowledge that bone-implant contact during OMI insertion also influences primary stability. This aspect was not included in the scenarios simulated in this study.

The expression “bone quality” has been used in the literature for a long time, but still remains vague and elusive.[42,43] This generally represents the combination of all characteristics that can affect fracture resistance, including attributes related to size, shape and material properties,[44,45] as well as density, microarchitecture, quantity and morphology of lacunae, along with configuration, distribution and alignment of collagen.[46]

Nonetheless, Martinelli *et al*[15] using computerized tomography to measure CBT also found buccal cortical bone in the upper molars region, notably slender than in the mandible, and reported that OMI stability may be directly affected by CBT.

Presently, the single parameter that can be used to define bone quality in clinical practice is cortical bone thickness. Hence, the results may be interpreted as an indication for use of shorter transgingival collars in slender cortical bone regions, which can occur in different parts of both the maxilla and the mandible.[15] Consequently, the importance of studying and designing orthodontic mini-implants appropriate for cases of slender or absent cortical bone is paramount. The authors do not recommend the usage of transgingival collars longer than 2 mm, specifically in cortical bone regions leaner than 1.5 mm.

CONCLUSIONS

A direct relation was observed between the increase in transgingival collar length and the increase in tension magnitude in the adjacent cortical bone. The increase in cortical bone thickness affected the minimum principal stress in the region of interest, causing it to decrease.

Observing mechanical performance, the authors recommend orthdontic mini-implants with transgingival collar length as short as possible, especially for cases of slender cortical bone. Upon the need for usage of mini-implants with longer collar lengths (2 and 3 mm), they should be inserted in cortical areas at least 1.5 mm thick. Mini-implants with 1 mm collar length can be used in cortical bones 1.0 mm thick. Usage of mini-implants with 4 mm transgingival collars is not recommended.

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TABLES AND FIGURES LEGENDS

Fig 1. Bone-implant model cross section view. Mini-implant (red), cortical bone (green), and cancellous bone (gray).

Fig 2. Schematic drawing of orthodontic mini-implant. The OMI with 1 mm TC length is shown.

Fig 3. Analysis scenario showing boundary conditions: the load applied to the OMI head and the encaster on the opposite face.

Fig 4. Deformation of OMI submitted to a load parallel to the bone (preview has been enhanced by 150 times for visual aid). The colors show compression zones (gray) and traction zones (blue).

Fig 5. Magnitude of the minimum principal stress in the ROI *versus* CBT, comparing all TC lengths.

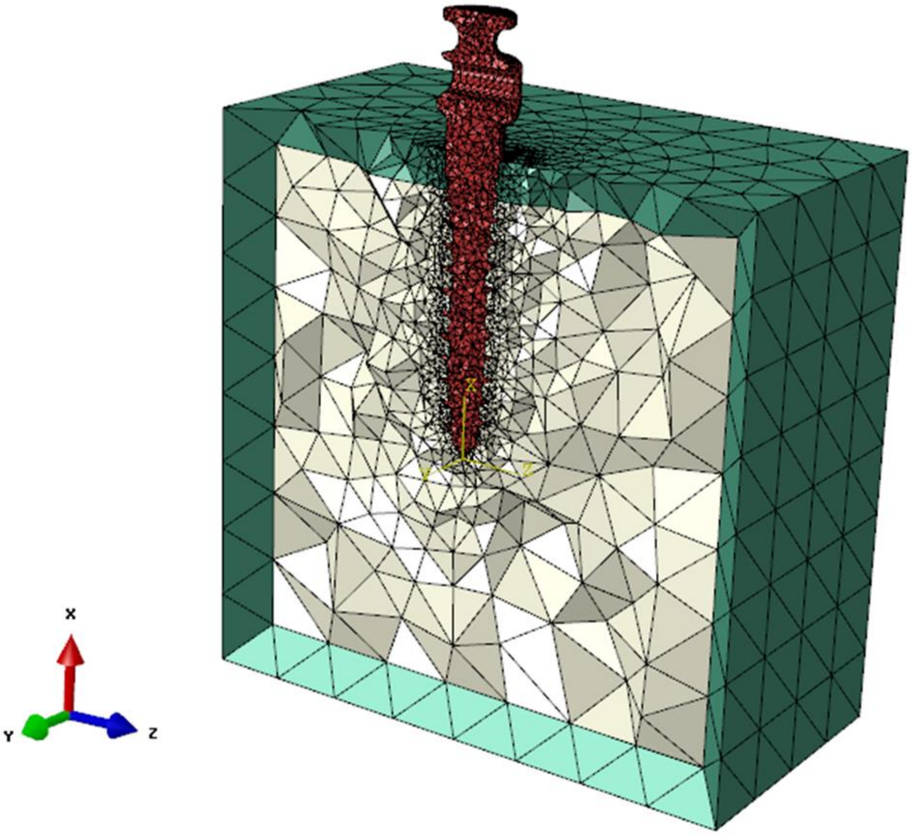
Fig 6. Distribution of minimum principal stress generated in the cortical bone with 0.25 mm thickness comparing all TC lengths: 1 mm (A), 2 mm (B), 3.0 mm (C) and 4.0 mm (D). Negative values indicate compression.

Fig 7. Displacement of OMI head in the same direction of the force *versus* CBT, comparing all TC lengths.

Table I. Mechanical properties of materials.

Table II. Magnitude of the minimum principal stress [MPa] in the ROI. The percentage increase in relation to the 1 mm TC is also represented. Negative values indicate compression.

Table III. Displacement of OMI head in the same direction of the force [μm] in the ROI. The percentage increase in relation to the 1 mm TC is also represented.



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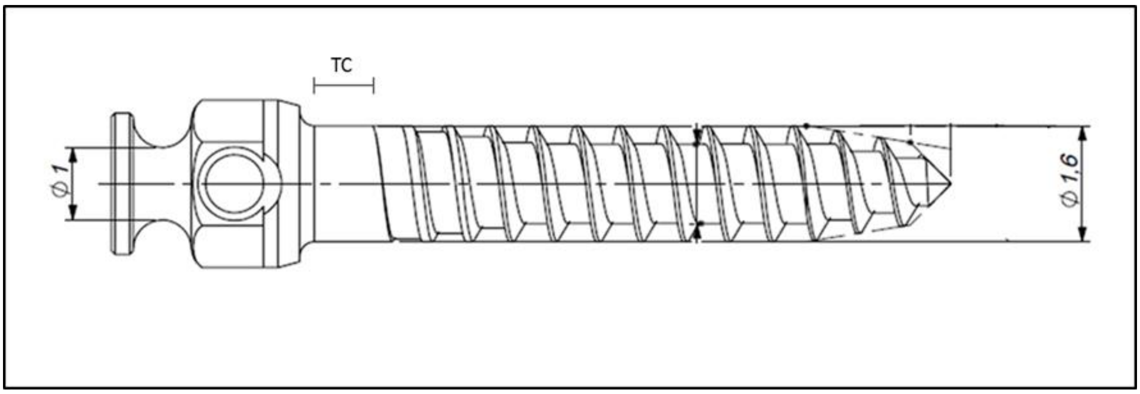

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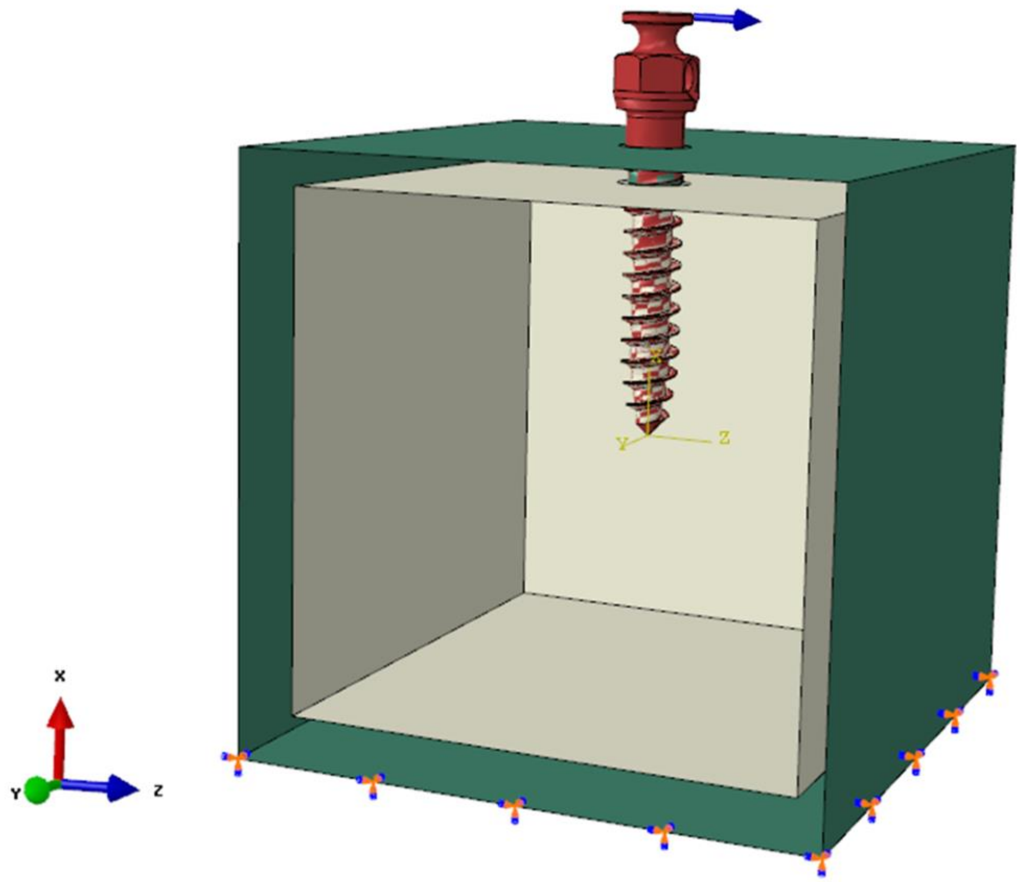
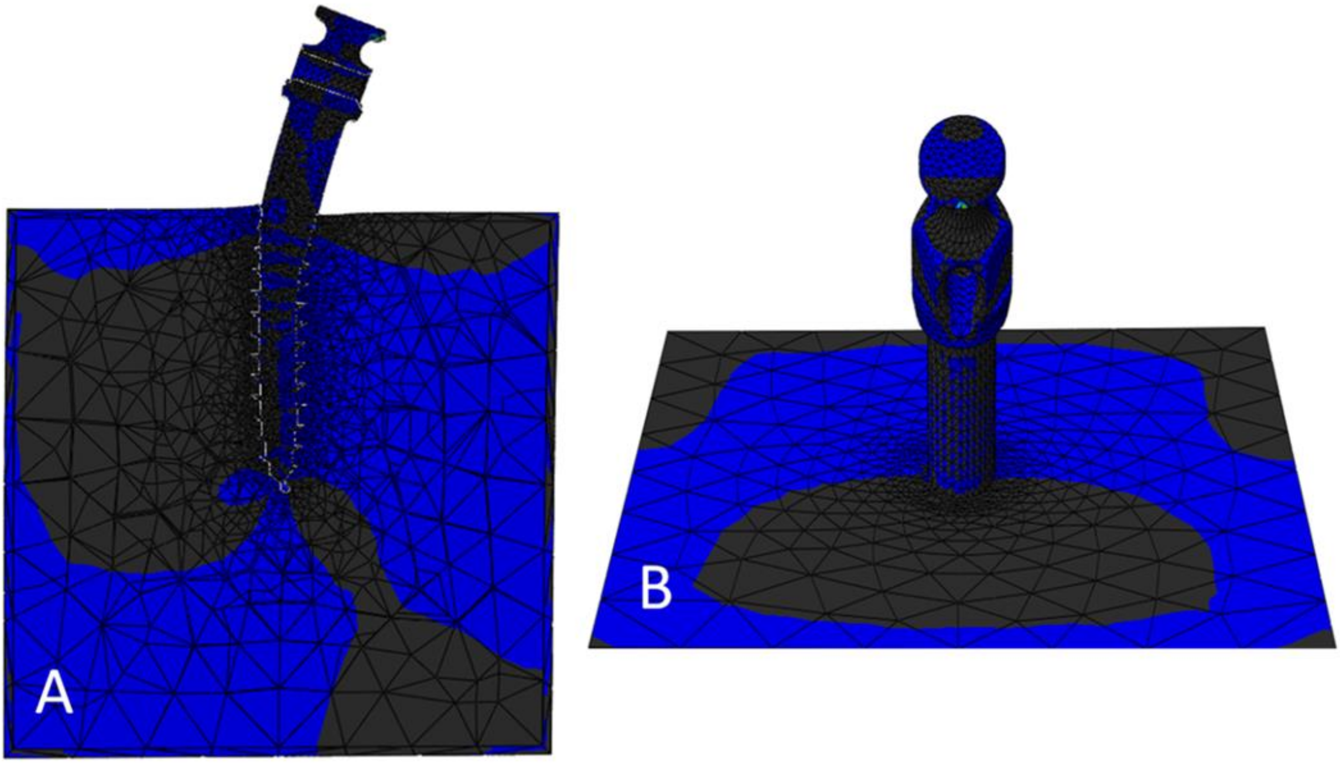


Fig 3. Analysis scenario showing boundary conditions: the load applied to the OMI head and the encaster on the opposite face.



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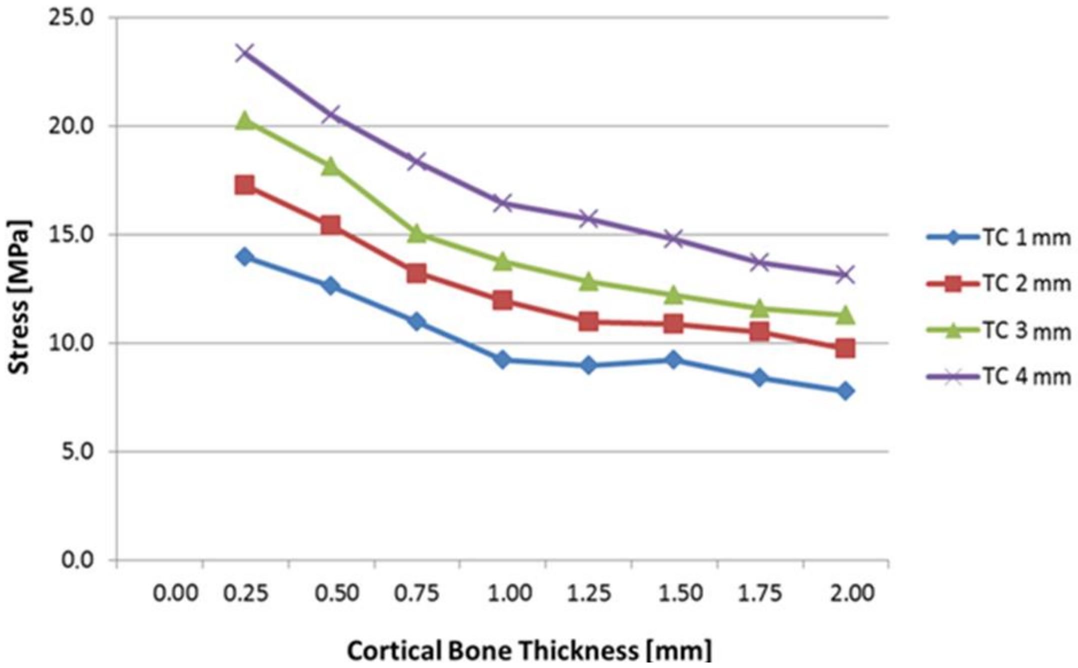

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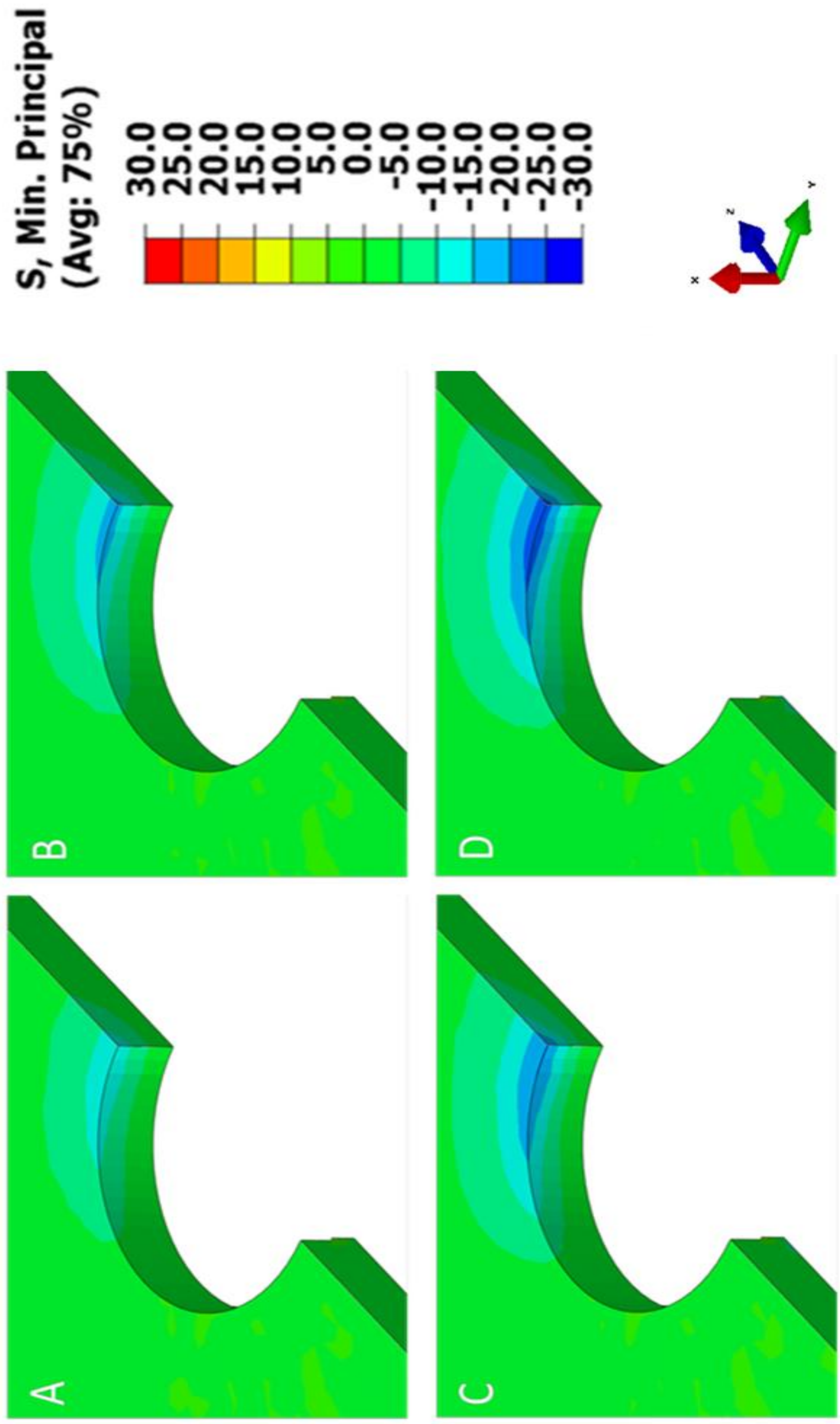

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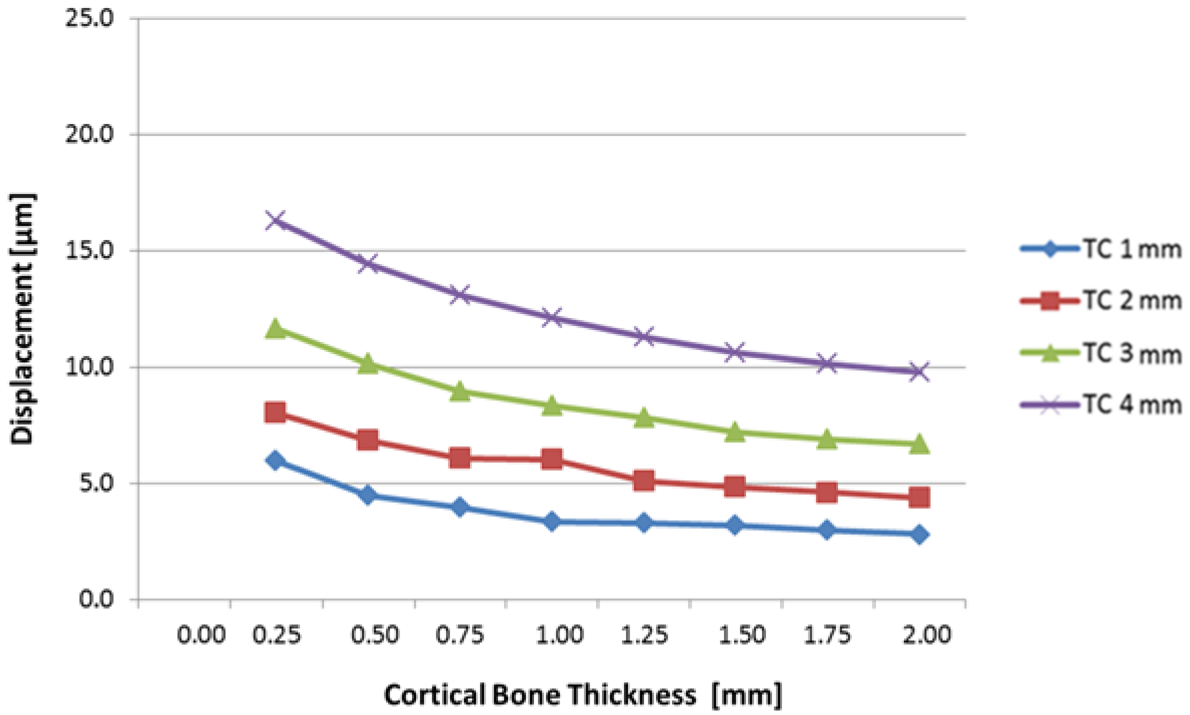

Fig 7. Displacement of OMI head in the same direction of the force *versus* CBT, comparing all TC lengths.

Table I. Mechanical properties of materials.

Material	Young modulus [MPa]	Poisson ratio	References
Ti-6Al-4V Grade 5	110,000	0.33	Collings,1984
Cortical bone	14,700	0.30	Field et al. 2009
Cancellous bone	490	0.30	Field et al. 2009

Table II. Magnitude of the minimum principal stress (MPS) in the ROI. The percentage increase in relation to the 1 mm TC length is also represented. Negative values indicate compression.

CBT	TC length [mm] and Magnitude of Minimum Principal Stress (MPS) [MPa]						
	TC 1mm	TC 2mm		TC 3mm		TC 4mm	
	MPS	MPS	%	MPS	%	MPS	%
0.25	-13.94	-17.27	23.83	-20.26	45.28	-23.33	67.27
0.50	-12.63	-15.41	22.01	-18.12	43.42	-20.53	62.51
0.75	-10.99	-13.22	20.30	-15.06	37.08	-18.36	67.03
1.00	-9.23	-11.97	29.66	-13.77	49.17	-16.44	78.11
1.25	-8.95	-11.00	22.83	-12.81	43.02	-15.72	75.48
1.50	-9.24	-10.86	14.49	-12.23	32.35	-14.81	60.22
1.75	-8.39	-10.53	25.50	-11.59	38.17	-13.71	63.47
2.00	-7.77	-9.76	25.63	-11.31	45.54	-13.16	69.41

Table III. Displacement of OMI head in the same direction of the force [μm] in the ROI. The percentage increase in relation to the 1 mm TC is also represented.

CBT	TC length [mm] and Displacement (δ) of OMI head [μm]						
	TC 1 mm	TC 2 mm		TC 3 mm		TC 4 mm	
	δ	δ	%	δ	%	δ	%
0.25	5.96	8.04	34.78	11.66	95.55	16.26	172.69
0.50	4.50	6.86	52.33	10.16	125.41	14.42	219.98
0.75	3.98	6.09	52.66	8.74	119.12	13.08	228.12
1.00	3.54	6.02	70.23	8.34	135.55	12.09	241.54
1.25	3.31	5.11	54.24	7.81	135.97	11.31	241.50
1.50	3.19	4.85	52.00	7.23	126.66	10.60	232.14
1.75	2.97	4.61	55.36	6.90	132.35	10.15	241.47
2.00	2.81	4.40	56.28	6.68	137.50	9.80	248.22

ARTIGO 2

Effect of orthodontic mini-implant thread diameter in stress distribution in the adjacent cortical bone: a finite element analysis.

Costa LFM, Nojima LI, Santos CE, Cimini Júnior CA, Bortoleto E, Santiago RC.

ABSTRACT

Introduction: Professionals use orthdontic mini-implants (OMI) with larger diameters, understanding this measure increases stability. However, the relation between diameter and stability has not been studied yet, especially when associated with cortical bone thickness (CBT) variation. **Methods:** Numerical simulation by means of the finite element method (FEM), analyzing the minimum principal stress in a bone-implant model. Different scenarios were considered with thread diameters of 1.2, 1.4, 1.6, 1.8 and 2.0 mm, assigning 8 different CBTs, from 0.25 mm to 2.0 mm, at increments of 0.25 mm. As an additional criterion, OMI head displacement was observed as well. **Results:** The OMI with 1.2 mm diameter presented mechanical performance inferior to others. The increase in OMI thread diameter, as well as in CBT, reduced stress magnitude and displacement. **Conclusions:** The increase in OMI thread diameter caused the decrease in stress magnitude and favored reduction of

displacement. A positive relation was observed between the increase in CBT and the increase in stability, especially for smaller diameters. For larger diameters, CBT bulkier than 1.0 mm can provide slight additional stability. The authors recommend using the largest possible diameter in cases with sufficient inter-radicular space. OMI with 1.2 mm diameter should be disregarded whenever possible.

INTRODUCTION

Stable anchor is an essential element for the success of orthodontic treatment. Along with miniplates, orthodontic mini-implants (OMI) have been used as skeletal anchorage and have been progressively more recommended. OMI feature advantages compared to conventional anchor, including extensive insertion possibilities in different positions of the alveolar bone, low cost and minimum patient cooperation.

Studies analyzing OMI stability are increasingly more common.[1-9] However, orthodontists remain concerned about device permanence during treatment, aware of failure possibility.[7,10,11] Among the fundamental aspects directly related to OMI failure, the possibility of micro-movement at the bone-implant interface is emphasized.[12] Different authors agree [3-5] and some of these even reinforce more aspects that directly influence micro-movement, including thread geometry,[8,9] insertion torque[13] and angle,[14] applied load,[15] root proximity,[16] cortical bone thickness (CBT) [1,6,8,9,17] and, finally, OMI dimensions, particularly length and diameter.[5,6,8,9,18-22]

Considering mechanical performance, the critical factor for success or failure of OMI is the stress distribution of tension generated in the surrounding bone,[23,24,25] whose magnitude depends on applied load, bone-implant interface, OMI geometry, and bone quality and quantity.[4,15,23,26] Less concentrated tension is understood to decrease the possibility of microdamage or bone resorption at bone-implant interface; thus seeking situations in which the generated tension acts as evenly as possible across the surrounding bone increases stability.[23]

Knowing that OMI may not osseointegrate, and that primary stability results from mechanical interlock between the thread and the alveolar bone,[5,27] it is assumed that an increased contact area, obtained by combining larger-diameter OMI and bulkier cortical bone, favors stress distribution at bone-implant interface, increasing stability. Several authors have stated that thread diameter greatly influences stability, arguing that larger-diameter OMI exhibit better success rates.[1,5,8,9,18] Yet, other authors mention that the increase in thread diameter, in addition to demanding higher insertion torque, can exert excessive pressure on the cortical bone,[13,28] permanently affecting its microstructure.[28,29,30] As there are conflicting opinions, and such situation directly relates to clinical practice, this study was designed to support the scientific community to broaden understanding about the aforementioned impact. Therefore, the finite element method (FEM) was used to evaluate the possibility that the increase in OMI thread diameter, as well as in CBT, could favor stability. A Numerical simulation was performed for static analysis of tension considering five different OMI thread diameters, each one in eight scenarios, distinguished by a specific CBT.

Similar studies have been made, although none examined both aspects simultaneously, OMI thread diameter compared for different CBTs, especially in a scenario with no osseointegration and immediate load.

MATERIALS AND METHODS

Three-dimensional geometric models simulating OMI monocortical insertion system were created with thread diameters from 1.2 to 2.0 mm, at increments of 0.2 mm, and CBT varying from 0.25 to 2.0 mm, at increments of 0.25 mm (Figure 1). Model properties were homogeneous, isotropic and linear elastic using as OMI material the Ti-6Al-4V Grade 5 alloy[31,32] (Table I).

SolidWorks commercial software (*Dassault Systemes Simulia Corp., Providence, RI, USA*) was used to model self-tapping OMI (Figure 2) designed by INP (*INP implantes, São Paulo-Brazil*) with 1.0 mm transgingival collar, 8.0 mm thread length, 0.2 mm fillet height and 0.6 mm pitch.

Models were created as interposed cubic blocks with 20 mm edge consisting of cortical and cancellous bones. The internal compact block with mechanical properties of cancellous bone and the hollow outer layer of variable thickness with mechanical properties of cortical bone were merged to form this part, combining different mechanical properties while remaining a thorough structure.

The analysis scenario was set interposing the OMI perpendicularly to the surface of the cube representing the bone. Boolean subtraction was employed to create the thread in the bone model.

The simulation scenario comprised OMI-bone contact interaction characteristics as real as possible. As there is no osseointegration for immediate load, the thread is enclosed inside the bone via mechanical interlock, exemplifying primary stability. Surface interaction was represented by the Coulomb static friction coefficient of 0.30 [5], corresponding to the OMI superficial texture.

The block face opposite to the OMI was encastered and a 2 N load was applied to the OMI head in a vector parallel to the bone model surface, simulating direct anchor (Figure 3).

The criterion to refine the mesh region surrounding the OMI was defined after a preliminary sensitivity study to determine the most appropriate global mesh, targeting consistent results at an acceptable computational cost. A circular partition 1.0 mm further than the bone-implant contact area was then created. The elements used were tetrahedral, with three dimensions and ten degrees of freedom (C3D10).

Abaqus 6.12 commercial software (*Dassault Systemes Simulia Corp., Providence, RI, USA*) was used for mesh generation and simulation. In order to provide an analysis with distinction of compression and traction regions, the resulting minimum principal stress was selected because it generates negative values for compression and positive values for traction. The external portion of the cortical bone adjacent to the OMI on the compression side was set as the region of interest (ROI) because this is where peak stress usually occurs after loading.[9] Understanding the stress transferred to the cancellous bone is minimal,[9,33] it was not analyzed.

Displacement of the OMI head was observed as an additional criterion for mechanical performance analysis. Such fact has been reported [34,35] and correlated with decrease in stability. [5,36]

RESULTS

For all models, regardless of OMI thread diameter or CBT, stress distribution was similar, exhibiting compression areas in the ROI and traction areas on the opposite side. The OMI tip demonstrated an analogous reaction, validating the mechanical interlock effect at bone-implant interface. The OMI considered more stable generated less tension in the ROI and exhibited minimum head displacement.

Considering all CBTs and diameters evaluated, Table II presents minimum principal stress magnitude values and Table III presents OMI head displacement values. The smallest thread diameter (1.2 mm), presented stress magnitude and displacement higher than the others. A relation was observed between the increase in thread diameter and the decrease in both stress magnitude and displacement. Stress distribution in the ROI is displayed in Figure 4 for all thread diameters in 0.25 mm CBT, in which the distinctive regions are better featured.

As CBT became bulkier, the generated tension showed additional reduction. Larger-diameter OMI (1.8 and 2.0 mm) showed less significant stress magnitude reduction, especially from 0.75 mm CBT on (Figure 5). OMI head displacement also showed the same tendency, and the 1.4 and 1.6 mm diameters exhibited similar values (Figure 6).

A relation between the increase in CBT and the increase in OMI stability was determined. A significant decrease in stress magnitude, as well as in displacement, was not observed from 1.25 mm CBT on. In accordance with the evaluation criteria, the smallest thread diameter (1.2 mm) performed inferior to others.

DISCUSSION

The literature is vast in clinical and *in vitro* studies [4,6,11,12,14] analyzing different OMI geometric aspects and its stability, though current scientific knowledge still has not provided results leading to definitive conclusions.

Selection of OMI suitable to each clinical situation demands careful assessment of the relation among different aspects, being the inter-radicular space one of the most important. It is common practice among orthodontists to use OMI with the largest possible diameter, being careful to maintain a safe distance from adjacent roots. Knowing that OMI with different diameters can exhibit similar mechanical performance is relevant to demonstrate the possibility of using smaller diameters, favoring stability and minimizing the risk of root impairment.

Stress magnitude analysis supports previous studies [27,37] asserting that less tension in the cortical bone is related to greater stability, and that overloading may cause OMI failure. Among the authors who studied cortical bone failure, Motoyoshi *et al* [38] observed variation from -31 to -55 MPa evaluating load direction and CBT; Kaplan *et al* [39], found maximum values of 50 MPa for cancellous bone and 170 MPa for cortical bone; and Li *et al* [40] determined the critical stress curve for bone resorption in high-density areas exceeding 25 MPa. The results in this study showed that tension magnitude in the ROI ranged from 7.52 to 25.38 MPa (von Mises). Such values were lower than those recorded by other authors.

Maximum OMI head displacement as a supporting stability parameter was also used by other authors.[9,33] The results observed in this study confirm its validity.

Among orthodontists it is common routine to recommend immediate load application to reduce treatment duration.[5] Comprehensive knowledge of factors which influence primary stability is important as well. Studies have examined the bone-implant interface with similar methodologies. Among these, some simulating non-osseointegrated OMI [5,22,37,41] and others considering osseointegration.[16,38] This study evaluated CBT variation, which is usual in different regions of both the maxilla and the mandible,[42] considering no osseointegration.

Stress distribution was represented similarly and in accordance with physical intuition in all models, as previously discussed.[41] The OMI tendency to tip in the same direction of the force was characterized in the simulation, generating compression zones in the ROI and traction zones in the opposite region. The OMI-bone mechanical interlock effect without osseointegration was validated by this simulation result. Structurally-united model components or scenarios disregarding surface contact do not represent the mechanical interlock which occurs at the post-insertion stage; instead, they reproduce the secondary stability resulting from osseointegration.

The major stability factor is the existing mechanical interaction between OMI fillets and the thread created in the bone at the initial post-insertion stages. Consequently, it is essential to establish a frictional condition for this relation between the corresponding surfaces, as described by Jiang *et al* [5]. The analysis criterion required sliding and/or separation between the surfaces in the interlock region, so the Coulomb static friction coefficient of 0.30 was used; such value is greater than the ones used by other authors.[8,37,41]

An association was observed between the increase in OMI diameter and the increase in stability, which was expected since a larger contact area at the bone-implant interface extends stress distribution in the adjacent bone.[1,6,8,12] More important than simply proving the positive relation between the increase in OMI diameter and the increase in stability, this study pursued to substantiate the understanding of how CBT may affect this phenomenon.

Thus, in cases with adequate space, the authors indicate OMI with the largest possible diameter. There are authors who agree,[8,9] recommending OMI with the largest diameters and the shortest extra-bone portions possible, seeking greater stability without affecting bone structure, as well as decreasing magnitude of the tension generated in the bone, resulting from the lever arm effect caused by the extra-bone portion of the OMI. The smallest thread diameter (1.2 mm) presented stress magnitude and displacement higher than the others and, consequently, the worst mechanical performance. Pithon, Figueiredo and Oliveira [43] validate this statement, expressing that diameter reduction not only decreases mechanical resistance, but it also affects primary stability.

Nonetheless, and opposing recent assertions[9], authors have declared the increase in diameter can permanently affect cortical bone microstructure during insertion.[28,29,30] Such claim can be assessed by means of simulating OMI penetration into the bone. This study has not performed dynamic analysis, so its findings can be interpreted in clinical practice assuming the OMI did not cause considerable impairment during insertion, or the impairment caused by all OMI was equivalent.[9] Recent studies [22,33] have adopted a similar methodology, not performing dynamic analysis, validating results with the allegation that the bone tissue has viscoelastic properties, being adaptable and dissipating tension during OMI insertion.[44]

The same tendency of tension magnitude stabilization starting at 1.0 mm CBT was identified in a recent FEM study [17] analyzing CBT influence in OMI stability considering osseointegration; displacement was not analyzed.

Analyzing the stress magnitude in the ROI, it was observed that as CBT became bulkier the stress magnitude decreased. This tendency is more noticeable in the OMI with the smallest thread diameter, which presented the greatest peak stress magnitude difference for extreme CBTs (0.25 and 2.0 mm). For the other diameters, there is a tendency of stress magnitude to remain relatively similar from 1.25 mm CBT on. Additionally, a noticeable tendency was observed that the increase in CBT causes less significant displacement reduction, particularly from 1.5 mm CBT on.

CONCLUSIONS

The increase in OMI thread diameter caused the decrease in minimum principal stress magnitude in the adjacent bone and favored reduction of OMI head displacement in all scenarios evaluated.

A positive relation was observed between the increase in CBT and the increase in stability, especially for OMI with smaller diameters. For OMI with larger diameters, CBT bulkier than 1.0 mm can provide slight additional stability.

The authors recommend using OMI with the largest possible diameter in cases with sufficient inter-radicular space. OMI with 1.2 mm diameter performed inferior to others and should be disregarded whenever possible.

Understanding of the relation between mini-implant diameter and cortical bone thickness is highly beneficial to achieve a successful stability during treatment.

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TABLES AND FIGURES LEGENDS

Fig. 1 Bone-implant model cross section view. Mini-implant (green), cortical bone (purple), and cancellous bone (yellow).

Fig. 2 Schematic drawing of orthodontic mini-implant. The OMI with 1.6 mm diameter is shown.

Fig. 3 Analysis scenario showing boundary conditions: the load applied to the OMI head and the encaster on the opposite face.

Fig. 4 Distribution of minimum principal stress generated in the cortical bone with 0.25mm thickness comparing all thread diameters: 1.2 mm (A), 1.4 mm (B), 1.6 mm (C), 1.8 mm (D), and 2.0 mm (E). Negative values indicate compression.

Fig. 5 Magnitude of the minimum principal stress in the ROI versus CBT, comparing all thread diameters.

Fig. 6 Displacement of OMI head in the same direction of the force versus CBT, comparing all thread diameters.

Table I. Mechanical properties of materials.

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Table III. Displacement of OMI head in the same direction of the force [μm] in the ROI. The percentage increase in relation to the previous thread diameter is also presented.

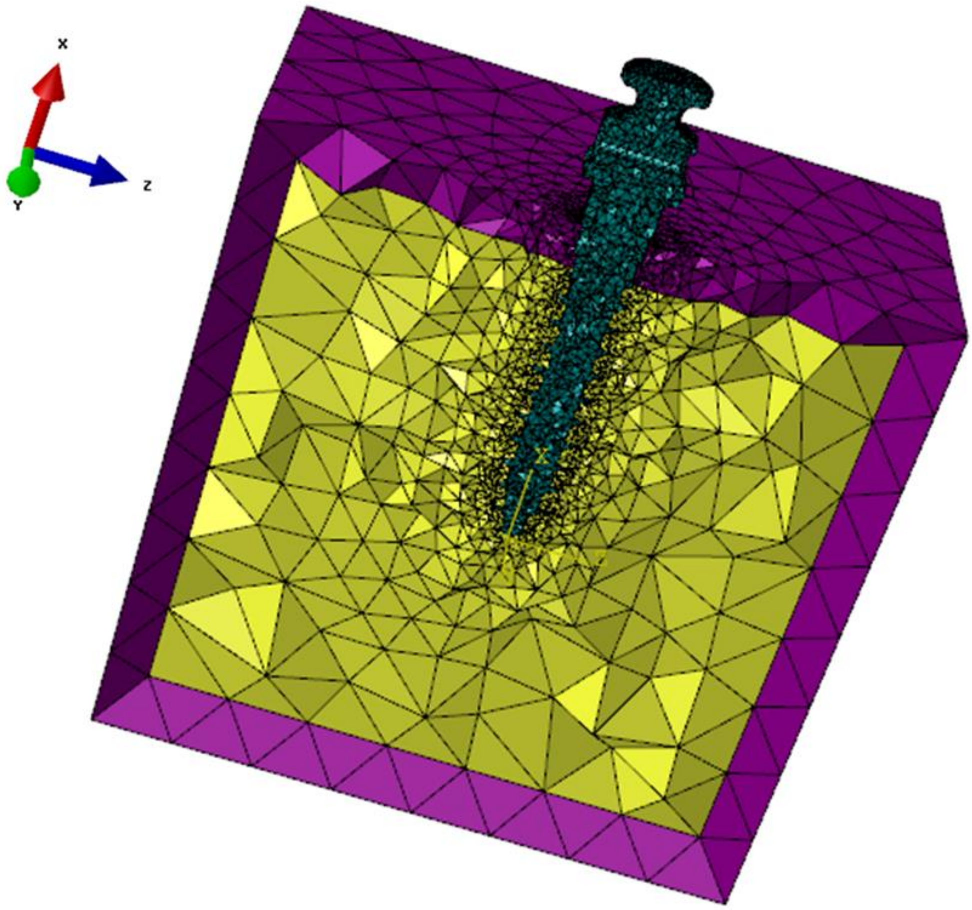

Fig 1. Bone-implant model cross section view. Mini-implant (green), cortical bone (purple), and cancellous bone (yellow).

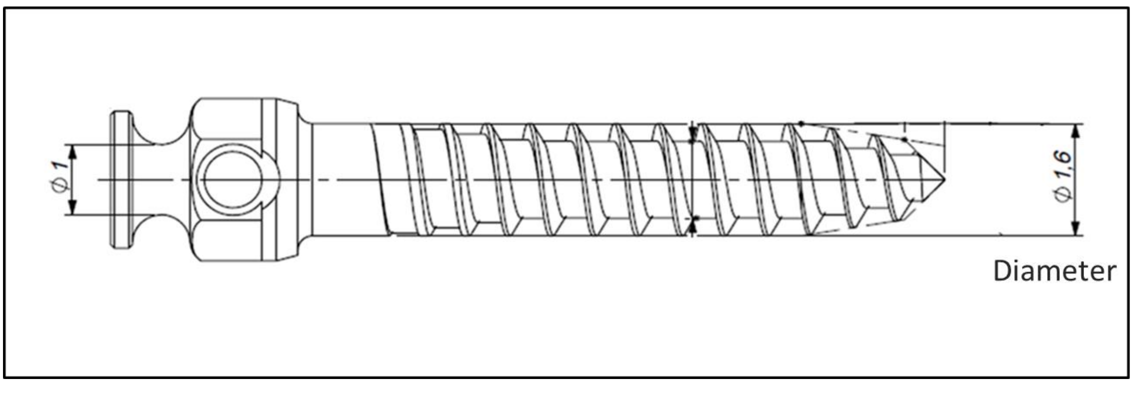

Fig 2. Schematic drawing of orthodontic mini-implant. The OMI with 1.6 mm diameter is shown.

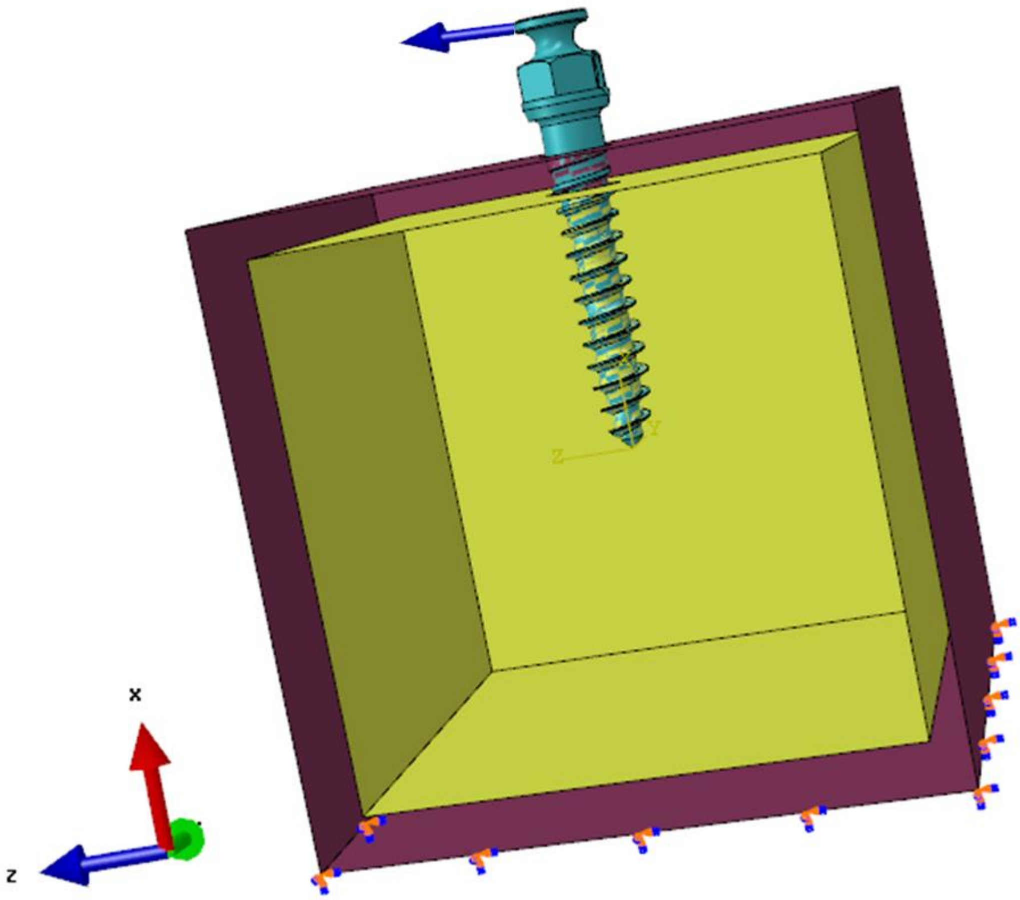

Fig 3. Analysis scenario showing boundary conditions: the load applied to the OMI head and the encaster on the opposite face.

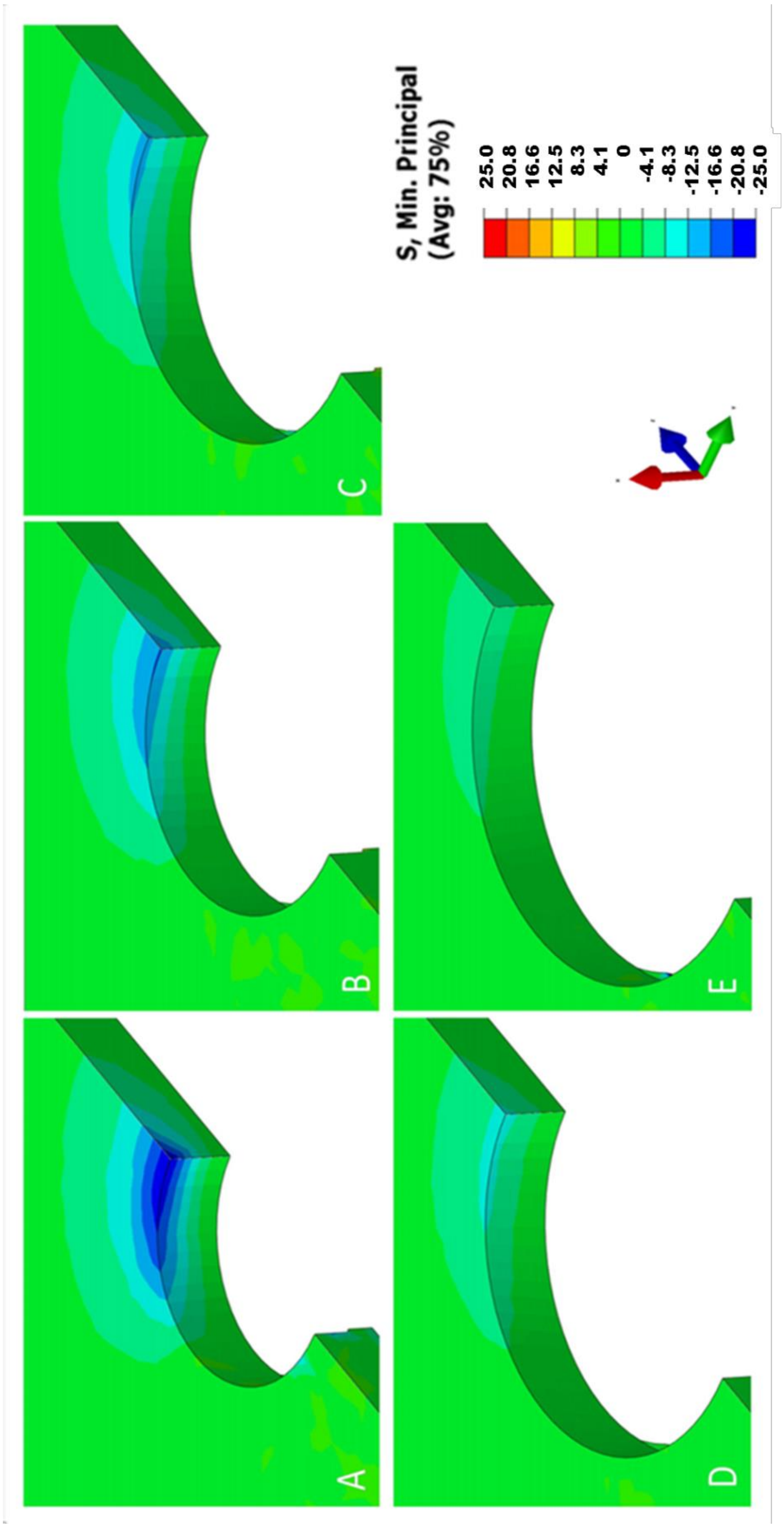

Fig 4. Distribution of minimum principal stress generated in the cortical bone with 0.25 mm thickness comparing all thread diameters: 1.2 mm (A), 1.4 mm (B), 1.6 mm (C), 1.8 mm (D), and 2.0 mm (E). Negative values indicate compression.

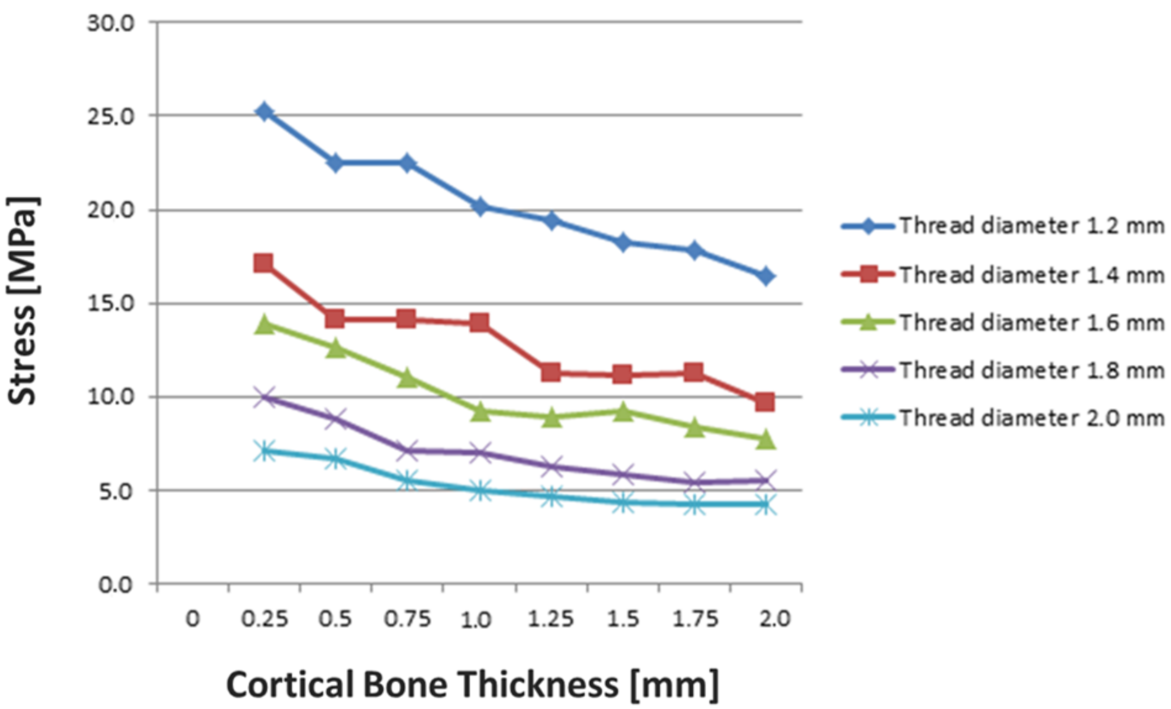

Fig 5. Magnitude of the minimum principal stress in the ROI versus CBT, comparing all thread diameters.

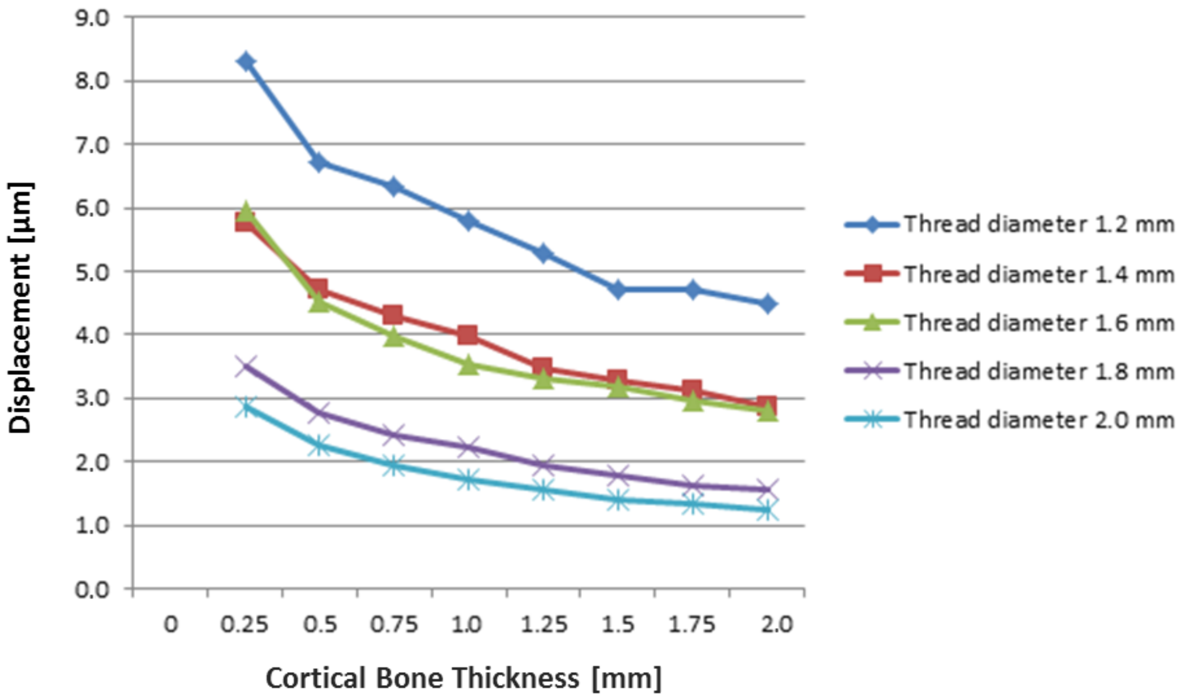

Fig 6. Displacement of OMI head in the same direction of the force versus CBT, comparing all thread diameters.

Table I. Mechanical properties of materials.

Material	Young modulus [MPa]	Poisson ratio	References
Ti-6Al-4V Grade 5	110,000	0.33	Collings,1984
Cortical bone	14,700	0.30	Field et al. 2009
Cancellous bone	490	0.30	Field et al. 2009

Table II. Magnitude of the minimum principal stress [MPa] in the ROI. The percentage increase in relation to the previous thread diameter is also presented. Negative values indicate compression.

CBT	Thread diameter (TD) [mm] and Magnitude of Minimum Principal Stress (MPS) [MPa]								
	TD 1.2 mm	TD 1.4 mm		TD 1.6 mm		TD 1.8 mm		TD 2.0 mm	
	MPS	MPS	%	MPS	%	MPS	MPS	%	MPS
0.25	-25.23	-17.10	32.21	-13.94	18.47	-10.02	28.11	-7.15	28.69
0.50	-22.51	-14.15	37.16	-12.63	10.71	-8.77	30.56	-6.64	24.28
0.75	-22.48	-14.15	37.03	-10.99	22.36	-7.15	34.93	-5.53	22.57
1.00	-20.20	-13.87	31.34	-9.23	33.43	-7.03	23.87	-4.96	29.33
1.25	-19.43	-11.29	41.86	-8.95	20.70	-6.24	30.25	-4.68	25.04
1.50	-18.30	-11.15	39.08	-9.24	17.09	-5.89	36.28	-4.40	25.28
1.75	-17.83	-11.25	36.86	-8.39	25.47	-5.38	35.81	-4.25	21.04
2.00	-16.41	-9.70	40.87	-7.77	19.91	-5.49	29.31	-4.25	22.57

Table III. Displacement of OMI head in the same direction of the force [μm] in the ROI. The percentage increase in relation to the previous thread diameter is also presented.

CBT	Thread diameter (TD) [mm] and Displacement (δ) of OMI head [μm]								
	TD 1.2 mm	TD 1.4 mm		TD 1.6 mm		TD 1.8 mm		TD 2.0 mm	
	δ	δ	%	δ	%	δ	%	δ	%
0.25	8.30	5.75	30.70	5.96	3.68	3.50	41.18	2.86	18.27
0.50	6.73	4.70	30.10	4.50	4.15	2.78	38.28	2.26	18.47
0.75	6.32	4.29	32.08	3.98	7.15	2.41	39.58	1.94	19.50
1.00	5.78	3.98	31.11	3.54	11.07	2.21	37.33	1.71	22.58
1.25	5.29	3.48	34.28	3.31	4.85	1.94	41.29	1.55	20.00
1.50	4.72	3.27	30.68	3.19	2.50	1.78	44.08	1.41	20.84
1.75	4.72	3.12	33.73	2.97	4.96	1.62	45.27	1.32	18.44
2.00	4.49	2.85	36.39	2.81	1.47	1.57	44.18	1.25	19.91

ARTIGO 3

Influence of orthodontic mini-implant thread length and bone mechanical properties on stress transmission to cortical bone: A three-dimensional finite element analysis

Costa LFM, Nojima LI, Santos CE, Cimini Júnior CA, Bortoleto E, Santiago RC.

ABSTRACT

Introduction: Orthodontic mini-implant (OMI) thread length influence in its mechanical performance has been studied and is intensely debated in the literature. Clinical understanding is that OMI with longer thread length are more stable. **Methods:** Numerical simulation by means of the finite element method (FEM), analyzing the minimum principal stress in a bone-implant model. Different scenarios were considered with thread lengths of 6; 8; 10 and 12 mm, assigning 8 different CBTs, from 0.25 mm to 2.0 mm, at increments of 0.25 mm, associated with cancellous bone Young moduli of 490 MPa and 1300 MPa. As an additional criterion, OMI head displacement was observed as well. **Results:** The OMI with the shortest thread length (6 mm) exhibited the highest stress magnitude for both cancellous bone Young moduli. The remaining threat lengths (8, 10 and 12 mm) exhibited similar magnitudes. **Conclusions:** The authors identified a relation between the increase in thread length and the increase in OMI stability. Short-length OMI (6 mm) are indicated only for bulkier

CBTs (2.0 mm). Thread lengths exceeding 8 mm may cause aggravated impairment during insertion and may not provide additional stability.

INTRODUCTION

Primary stability is critical to maintain orthodontic mini-implants (OMI) in position, since failure often occurs during the post-insertion period.[1,2] Dental implants show success rates of 90-95%; [3,4] nonetheless, OMI do not perform similarly, even though they are used for relatively short periods.[5] The smaller OMI dimensions provide greater clinical applicability, though limited mechanical performance. Improvements have been proposed by changing different geometrical parameters, including modifications in thread shape and length.[1,2,6-11]

Stress distribution of tension generated in the surrounding bone is the critical factor for OMI performance,[12,13] whose magnitude depends on applied load, bone-implant interface, OMI geometry, and bone quality and quantity.[14-16] The possibility of microdamage or bone resorption at bone-implant interface is understood to decrease upon less concentrated tension; desirable scenarios are those in which the generated tension acts as evenly as possible across the surrounding bone to increase stability.[12]

The influence of thread length on mechanical performance has been underinvestigated and intensely debated. Physical intuition indicates that its increase may favor stability, especially in higher quality bone regions, such as those featuring bulkier cortical bone and/or denser cancellous bone. Considering such factors, it is common practice among orthodontists to use longer OMI in inter-radicular spaces which restrict the diameter. Still, there are studies indicating the increase in length does not directly affect stability.[15,17-20]

Assuming that bulkier cortical bones and denser cancellous bones contribute to primary stability, and that failure most frequently occurs at the bone-implant interface, this study aims at investigating whether the increase in OMI thread length can be a relevant factor to increase stability. Additionally, factors associated with alveolar bone quality were assessed: cortical bone thickness (CBT) variation and cancellous bone Young modulus. The analysis scenarios consider immediate load and no osseointegration in the bone-implant interface.

MATERIALS AND METHODS

Virtual geometric models were created to simulate a three-dimensional bone-implant monocortical insertion system with OMI thread lengths of 6, 8, 10 and 12 mm, CBT varying from 0.25 to 2.0 mm, at increments of 0.25 mm, and cancellous bone with Young modulus of 490 MPa and 1,300 MPa (Figure 1). Ti-6Al-4V Grade 5 alloy was defined as OMI material; all model properties set as homogeneous, isotropic and linear elastic [21,22] (Table I).

The aspects chosen to characterize alveolar bone quality were CBT and cancellous bone Young modulus. These aspects were associated to comprise the analysis scenarios; hence, the bulkiest cortical bone and the cancellous bone with the highest Young modulus configure a bone with better quality.

Self-tapping OMI were modeled using SolidWorks commercial software (*Dassault Systemes Simulia Corp., Providence, RI, USA*) designed by INP (*INP* *implantes, São Paulo-Brazil*) with 3.5mm extra-bone portion (1.0 mm transgingival collar), 1.0 mm major diameter, 1.6 mm thread diameter, 0.2 mm fillet height and 0.6 mm pitch (Figure 2).

Interposed cubic blocks with 20 mm edge were used to model the cortical and cancellous bones. To maintain a thorough structure, the internal compact block representing the cancellous bone was merged to the outer layer of variable thickness representing the cortical bone, combining and maintaining their respective and distinct mechanical properties.

The OMI was interposed perpendicularly to the surface of the cube representing the bone for analysis. The thread was subtracted from the bone model using a Boolean combination.

A realistic simulation scenario was characterized in terms of OMI-bone contact relation. The thread is enclosed inside the bone via mechanical interlock for immediate load, without osseointegration, exemplifying primary stability. Surface interaction was represented by the Coulomb static friction coefficient of 0.30[23], corresponding to the OMI superficial texture.

Encaster restriction was applied to the model on opposite surface to the OMI and a 2 N load was applied to the OMI head in a vector parallel to the bone model surface; simulating direct anchor (Figure 3).

A preliminary sensitivity study was conducted to determine the most appropriate global mesh, targeting consistent results at an acceptable computational cost. Following this procedure, the criterion to refine the mesh region surrounding the OMI was defined, creating a circular partition 1.0 mm further than the bone-implant contact area. Three-dimensional, ten degree-of-freedom tetrahedral elements were used (C3D10).

Mesh generation and simulation were performed with Abaqus 6.12 commercial software (*Dassault Systemes Simulia Corp., Providence, RI, USA*). To analyze different compression and traction regions, the resulting minimum principal stress was selected because it generates negative values for compression and positive values for traction. The external portion of the cortical bone adjacent to the OMI on the compression side was set as the region of interest (ROI) because this is where peak stress usually occurs after loading.[10] Understanding the stress transferred to the cancellous bone is minimal, [9,10] it was not analyzed.

Mechanical performance was also analyzed by observing OMI head displacement as an additional criterion. Such fact has been reported [24,25] and correlated with decrease in stability.[17,23,26]

RESULTS

In this study, the OMI considered more stable generated less tension in the ROI and exhibited minimum head displacement.

Magnitude of minimum principal stress

Values of minimum principal stress in the ROI are listed in Table 2 for all CBTs, considering both cancellous bone Young moduli and all thread lengths evaluated. The OMI with the shorter thread (6 mm) presented stress magnitude higher than the others. This occurred regardless of increase in cancellous bone Young modulus, as shown in Figure 3. The remaining thread lengths (8, 10 and 12 mm) exhibited similar magnitudes.

CBT *versus* magnitude of minimum principal stress

Figure 3 compares all thread lengths considering both cancellous bone Young moduli, and indicates that increasing CBT reduces peak stress in the ROI. For all OMI this tendency was observed, particularly between 0.25 and 1.0 mm CBT, with relatively similar magnitudes from 1.0 mm CBT on. As for the bulkiest CBT evaluated (2.0 mm), its influence is substantial for all thread lengths, especially 6.0 mm, which shows magnitude similar to the others.

Stress distribution in the ROI is displayed in Figure 4 for all thread lengths in 0.25 mm CBT, in which the distinctive regions are better featured.

Figure 5 compares each thread length considering both cancellous bone Young moduli. Physical intuition suggests the cancellous bone with higher Young modulus has increased contention capacity, favoring OMI stability. This fact was noted and is relevant for slender CBTs, reducing peak stress magnitude difference as CBT becomes bulkier. This additional contention was not significant from 1.0 mm CBT on, especially for 10 and 12 mm thread lengths.

CBT *versus* OMI head displacement

Displacements followed a tendency similar to that observed for stress, *i.e.*, decrease as CBT became bulkier (Table III). Thread lengths of 6 and 8 mm exhibited similar values, likewise for the lengths of 10 and 12 mm (Figure 6).

DISCUSSION

OMI stability is intensely debated in the literature. Different authors claim the increase in thread length does not favor stability,[15,18-20] considering both bone morphology and OMI geometry as relevant factors; others highlight the extra-bone portion is more significant for mechanical performance than the intra-bone portion.[15,18-20] According to the latter authors, interpreting the OMI as a cylinder subject to pure bending, geometric analysis indicates the diameter has greater influence in stress generation than the length.

However, there is a significant amount of studies emphasizing the increase in thread length may be related to the increase in stability [1,2,7,11,15,17-20,23,28,29] and Liu *et al* [10] support such analysis stating the extra-bone portion should be considered, because it acts as a lever arm upon load. The intra-bone portion should be considered in case it is sufficient to reduce the action of the lever arm.

This study observed the shortest thread length exhibited the highest stress magnitude in the ROI and, according to the assessment criteria, the worst performance. Agreeing with Liu *et al* [10], since all OMI evaluated have the same extra-bone length, one possible explanation for this performance could be the inability of the intra-bone portion (6 mm) to contain the action of the lever arm caused by the extra-bone portion (3.5 mm). Yet, it should be noted that stress magnitude decreased as CBT became bulkier, remaining similar to the values for the other thread lengths in 2.0 mm CBT. The mechanical interlock at the bone-implant interface may have produced such contention, thus using shorter-length OMI should be restricted to bulkier CBTs, from 2.0 mm on. For slender CBTs, orthodontists should use OMI with longer thread length to minimize the lever arm, especially in cases with longer extra-bone portions, such as in OMI with 2.0 or 3.0 mm transgingival collars or bracket heads.

For the other thread lengths (8, 10 and 12 mm), a relation between the increase in length and the decrease in stress magnitude in the ROI was not observed, possibly because the 8 mm intra-bone portion was satisfactory to reduce the action of the lever arm, and the additional increase was less effective. From these observations, it can be concluded that using OMI with thread length exceeding 8 mm may not provide additional stability in case the OMI head is short. Increasing thread length is recommended for longer extra-bone portions (transgingival collar exceeding 1.0 mm or bracket head) or when seeking bicortical anchor which, as stated by Brettin *et al* [30] is preferable to monocortical anchor, providing lesser tension magnitude in the cortical bone and greater stability.

For a long time, the expression “bone quality” has been used in the literature, yet it still remains vague and elusive, [31,32] commonly indicating the association of all characteristics that can affect fracture resistance, including attributes related to size, shape and material properties,[33,34] as well as density, microarchitecture, quantity and morphology of lacunae, along with configuration, distribution and alignment of collagen.[35]

In this research, the parameters defining bone quality were selected to sustain the numerical simulation, being CBT and cancellous bone Young modulus. CBT is widely adopted and has been directly related to OMI stability.[1,10,16,36-39] The parameters chosen comply with the classification proposed by Lekholm & Zarb [40], the most accepted in implantology, which is also based on the amount of cortical and cancellous bones.

It is understood that the cancellous bone of better quality presents a greater contention capacity, favoring OMI stability and decreasing stress magnitude in the ROI. This fact was observed, especially for slender CBTs. From 1.0 mm CBT on, stress magnitude remained similar, considering both cancellous bone Young modulus, substantiating the respective functions of cancellous and cortical bones in stabilization. As CBT became bulkier, it also increased retention capacity, and the contention provided by the cancellous bone became less significant, in agreement with Liu *et al.* [10]

Cancellous bone contribution to osseointegrated implant stability has been investigated, although there are few studies regarding OMI. Marquezan et al [41] assessing primary stability of OMI inserted into blocks of different bone mineral densities, with and without cortical bone, investigated if certain cancellous bone properties could influence primary stability and stated that, in fact, the cancellous bone is relevant. The results of the present study validate such claim, especially in cases with CBT leaner than 1.0 mm. For bulkier CBTs, cancellous bone contention is less effective.

To reduce treatment duration orthodontists regularly apply immediate load to OMI as a recommended routine. [23] Likewise, detailed comprehension of factors which influence primary stability is important. Even though the relation between CBT variation and OMI stability has already been discussed in the literature, greater emphasis was given to osseointegration scenarios.[12,30,39,42]

Analyzing the stress magnitude in the ROI, it was observed that as CBT became bulkier the stress magnitude decreased. This tendency is more noticeable in the OMI with the shortest thread length, which presented the greatest peak stress magnitude difference for extreme CBT (0.25 and 2.0mm). For the other thread lengths, there is a tendency of stress magnitude to remain relatively similar from 1.0mm CBT on. Stress magnitude and distribution in the ROI result from the lever arm effect caused by the extra-bone portion of the OMI.

CBT influence in OMI stability was also analyzed in a recent study[39] using FEM; the same tendency of stress magnitude to stabilize was identified from 1.0 mm CBT on, however in a scenario considering osseointegration.

Analyzing OMI head displacement, it was observed that as CBT became bulkier the displacement decreased. Thread lengths of 6 and 8 mm exhibited similar displacement values, likewise for the lengths of 10 and 12 mm. OMI head displacement results from the resistance to deformation of its constituent material; considering the applied load is equivalent to 0.002% of the OMI Young modulus, no significant difference can be perceived in the simulation.

CONCLUSIONS

In accordance with the analysis criteria adopted, the OMI with the shortest thread length (6 mm) exhibited the worst mechanical performance. The remaining thread lengths performed similarly.

Increase in CBT favored peak stress reduction in the ROI for all thread lengths analyzed. Increase in cancellous bone Young modulus may be associated with greater OMI stability, supplementing the cortical bone. This statement is valid, especially in cases with CBT leaner than 1.0 mm.

The authors identified a relation between the increase in thread length and the increase in OMI stability. Such relation is based on the capacity of the intra-bone portion, represented by the thread, to compensate the lever arm caused by the extra-bone portion, represented by the transgingival collar and head.

Clinically, short-length OMI (6 mm) are indicated only for bulkier CBTs (2.0 mm). For slender CBTs, orthodontists should use OMI with longer thread length, especially in cases with longer extra-bone portions, such as in OMI with 2.0 or 3.0 mm transgingival collars or bracket heads. Mini-implants with thread length exceeding 8 mm may cause aggravated impairment during insertion and may not provide additional stability.

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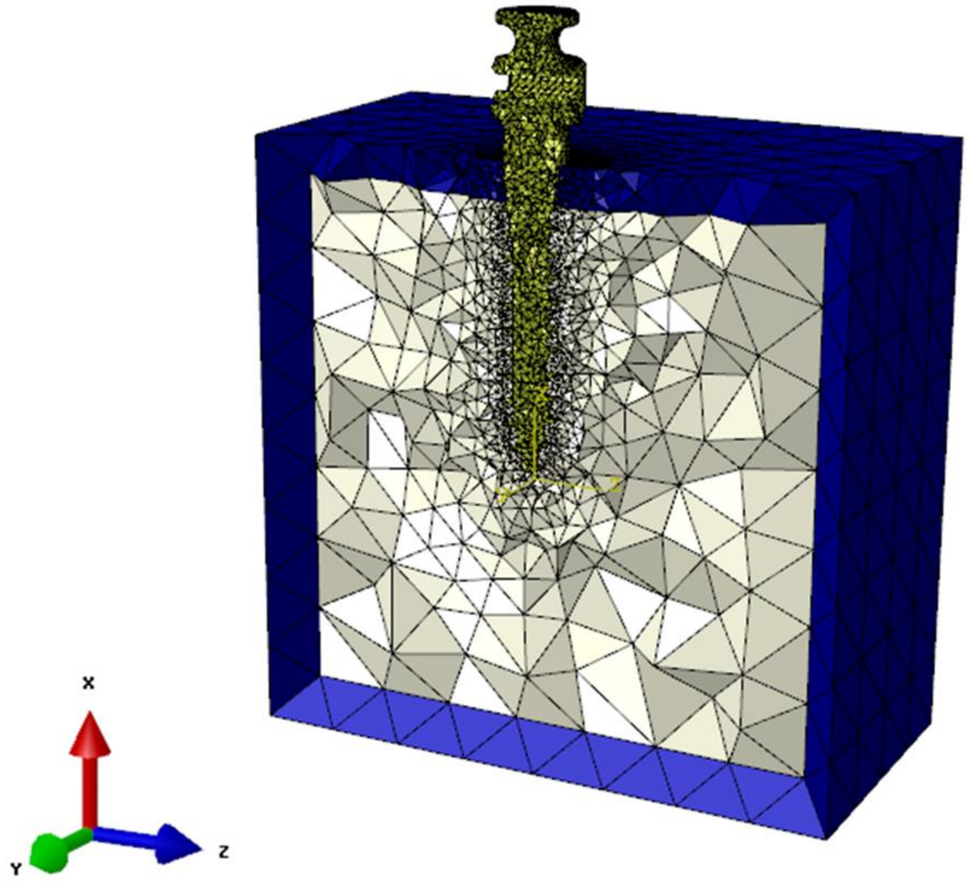

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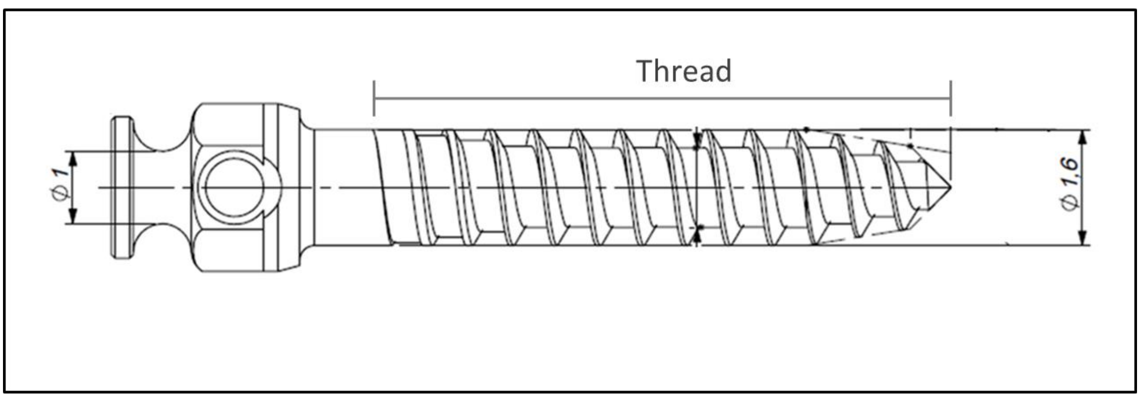

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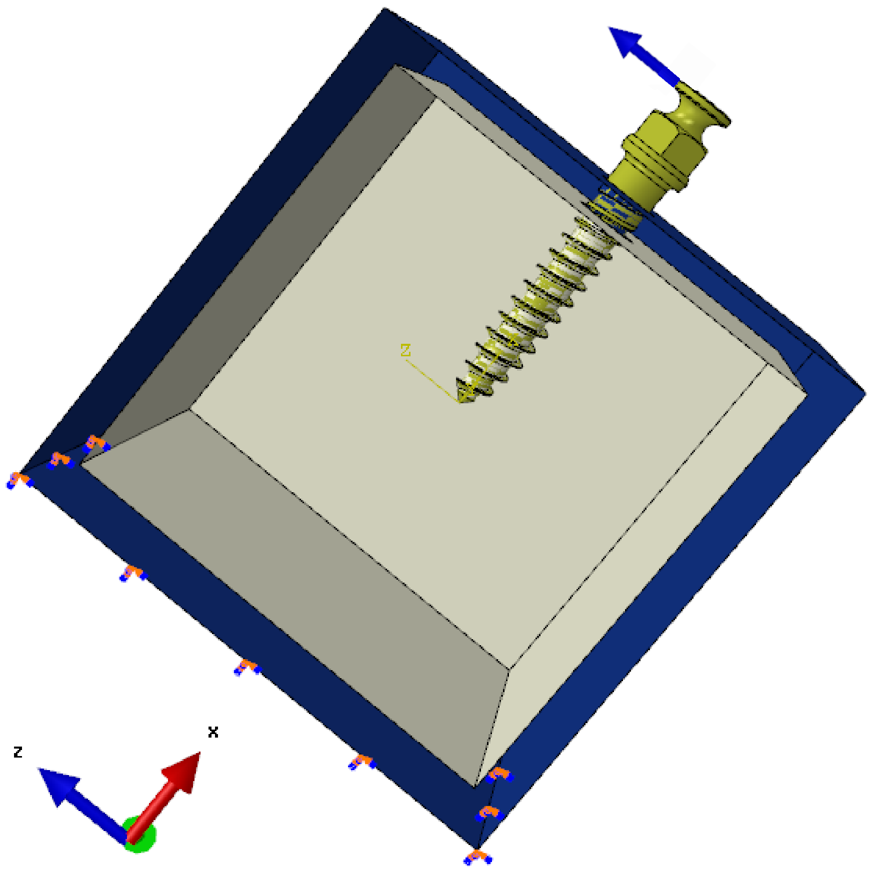
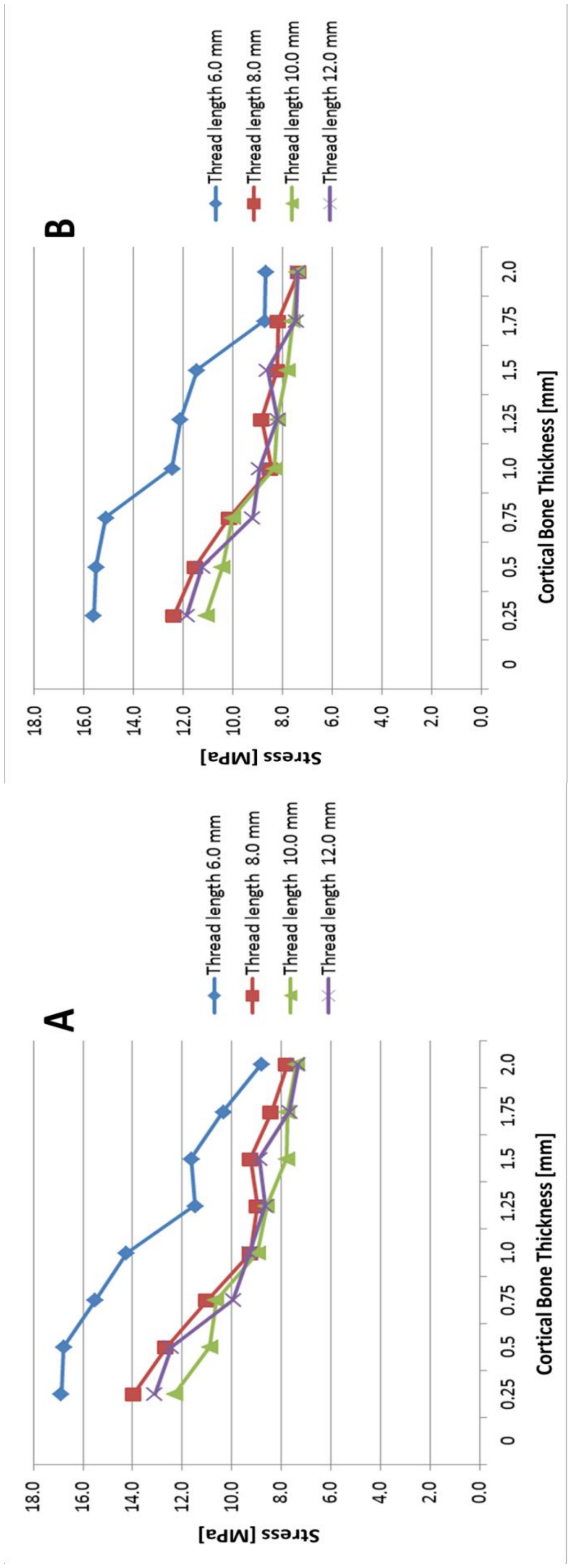


Fig 3. Analysis scenario showing boundary conditions: the load applied to the OMI head and the encaster on the opposite face.



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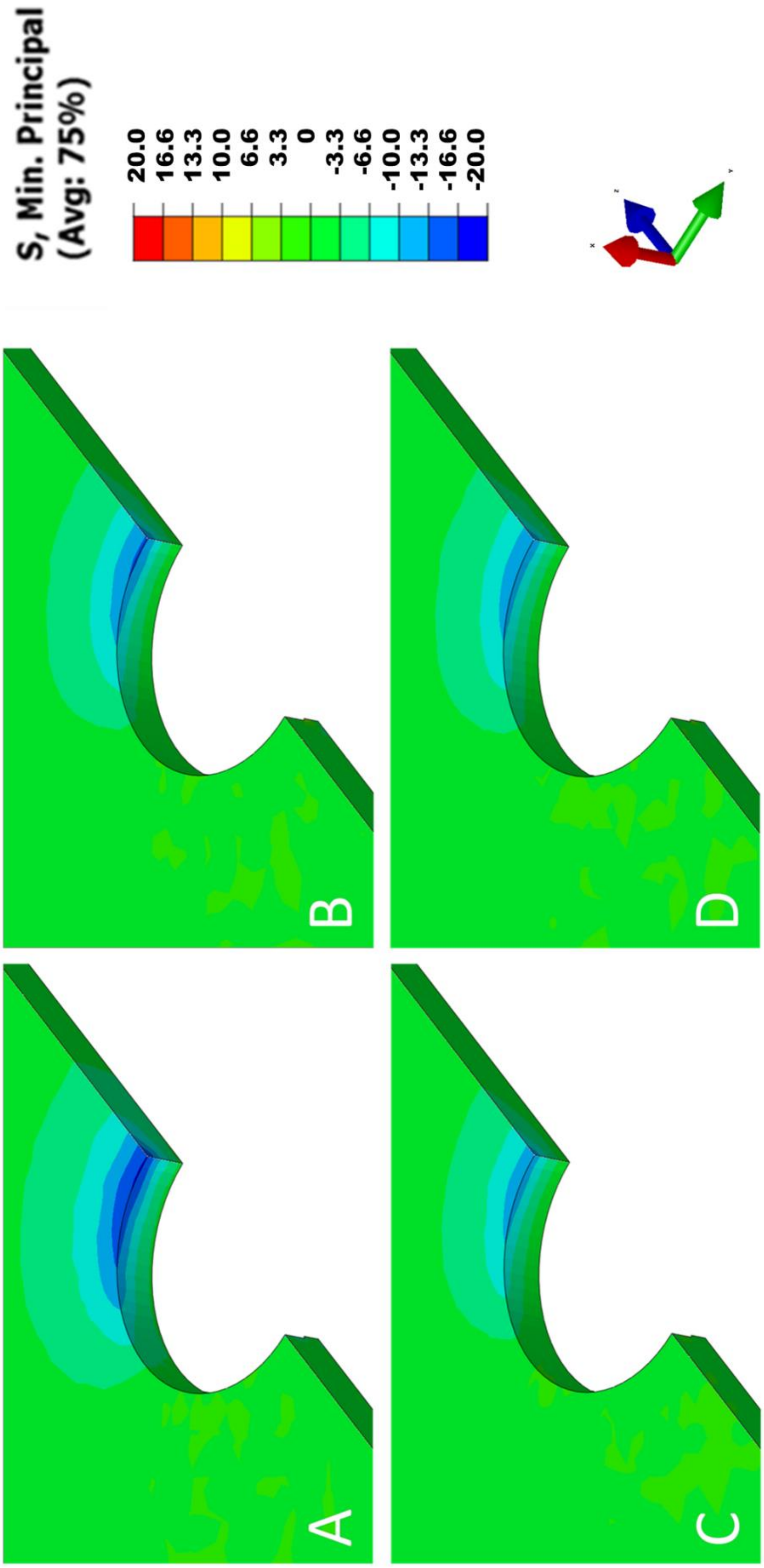

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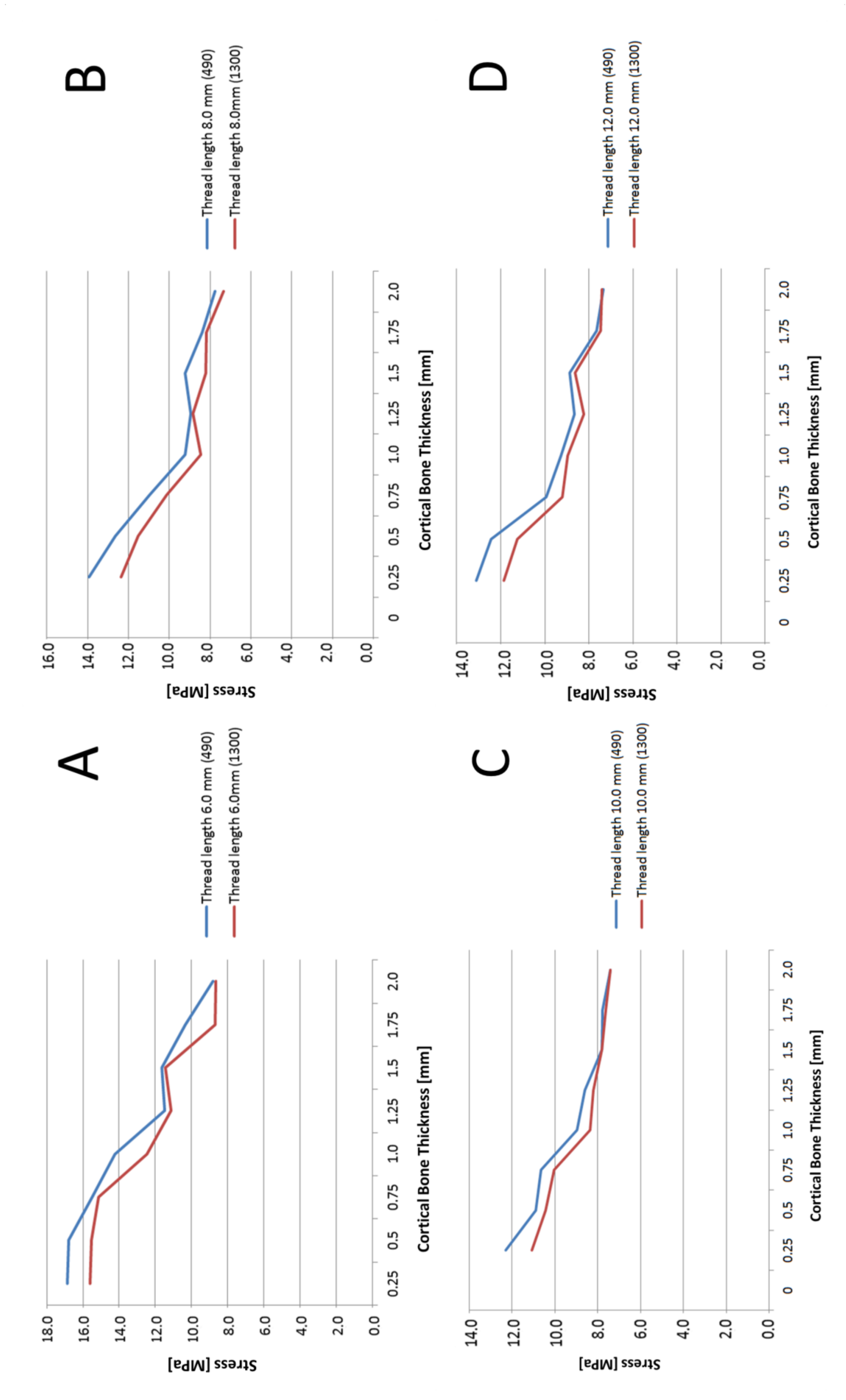

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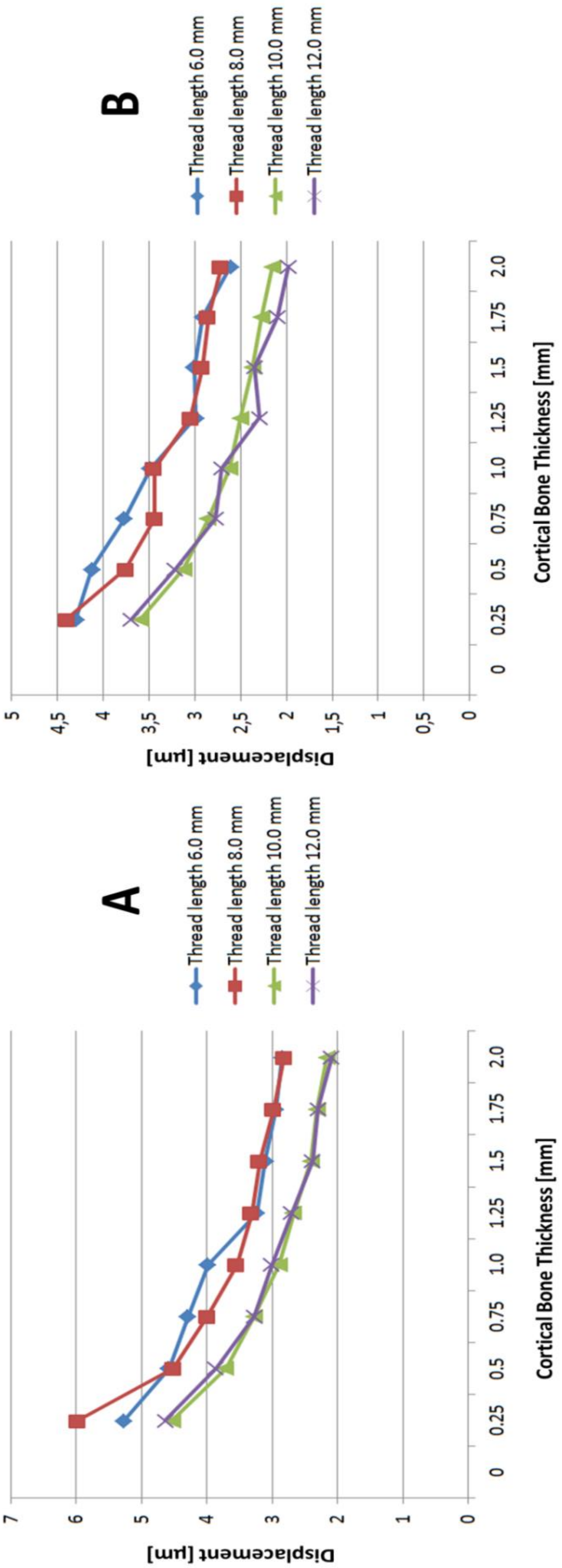

Fig 7. Maximum OMI head displacement in the same direction of force for all thread lengths and different CBTs. The cancellous bone Young modulus is 490 MPa (A) and 1300 MPa (B).

Table I. Mechanical properties of materials.

Material	Youngs modulus [MPa]	Poissons ratio	References
Ti-6Al-4V Grade 5	110,000	0.33	Collings, 1984
Cortical Bone	14,700	0.30	Field et al. 2009
Cancellous Bone (490 MPa)	490	0.30	Field et al. 2009
Cancellous Bone (1300 MPa)	1,300	0.30	Gracco et al. 2007

Table II. Magnitude of the minimum principal stress [MPa] in the ROI considering all aspects evaluated.

CBT	OMI thread lengths (TL) [mm] and Magnitude of Minimum Principal Stress (MPS) [MPa]							
	TL 6 mm		TL 8 mm		TL 10 mm		TL 12 mm	
	490 Mpa	1300 Mpa	490 Mpa	1300 Mpa	490 Mpa	1300 Mpa	490 Mpa	1300 Mpa
0.25	-16.87	-15.62	-13.94	-12.38	-12.30	-11.07	-13.09	-11.87
0.50	-16.76	-15.52	-12.63	-11.52	-10.88	-10.44	-12.44	-11.24
0.75	-15.50	-15.13	-10.99	-10.13	-10.63	-10.02	-9.94	-9.21
1.00	-14.25	-12.45	-9.23	-8.47	-8.96	-8.34	-9.27	-8.94
1.25	- 12.12	-11.12	-8.95	-8.84	-8.60	-8.20	-8.65	-8.22
1.50	-11.62	-11.46	-9.24	-8.22	-7.79	-7.81	-8.87	-8.63
1.75	-10.33	-8.71	-8.39	-8.18	-7.77	-7.60	-7.66	-7.47
2.00	-8.80	-8.67	-7.77	-7.35	-7.40	-7.41	-7.33	-7.39

Table III. Displacement of OMI head in the same direction of the force [μm] in the ROI considering all aspects evaluated.

CBT	OMI thread lengths (TL) [mm] and Displacement of OMI head [μm]							
	TL 6 mm		TL 8 mm		TL 10 mm		TL 12 mm	
	490 Mpa	1300 Mpa	490 Mpa	1300 Mpa	490 Mpa	1300 Mpa	490 Mpa	1300 Mpa
0.25	5.26	4.29	5.96	4.39	4.52	3.59	4.62	3.69
0.50	4.57	4.12	4.50	3.75	3.70	3.12	3.86	3.21
0.75	4.28	3.77	3.98	3.44	3.27	2.85	3.27	2.77
1.00	3.98	3.48	3.54	3.44	2.88	2.62	3.00	2.70
1.25	3.23	2.98	3.31	3.04	2.66	2.50	2.70	2.29
1.50	3.10	3.01	3.19	2.92	2.40	2.37	2.38	2.34
1.75	2.94	2.91	2.97	2.85	2.31	2.27	2.29	2.09
2.00	2.84	2.60	2.81	2.71	2.15	2.15	2.08	1.98

5 DISCUSSÃO

A literatura é ampla em estudos clínicos ou *in vitro* analisando a relação entre geometria e estabilidade de mini-implantes (Motoyoshi et al., 2009; Field et al., 2009; Jiang et al., 2009; Petrey et al., 2010; Duiabis et al., 2012); contudo, os resultados frequentemente são inconclusivos e conflitantes. Desta forma, com o conhecimento científico atual, prever a estabilidade para um MI a partir de sua geometria é tarefa difícil, especialmente quando se entende que quantidade e qualidade ósseas interferem significativamente na estabilidade.

A seleção do MI mais adequado a cada situação clínica passa por análise criteriosa de diferentes fatores, sendo o espaço interproximal um dos mais importantes. Ortodontistas geralmente selecionam mini-implantes com o maior diâmetro possível, desde que se mantenha distância segura das raízes dentárias adjacentes. Além disso, é senso comum que nos casos em que MI com maiores diâmetros não podem ser inseridos em decorrência de limitação mésio-distal, geralmente se adota o critério de utilizar maiores comprimentos de rosca. Entender como a variação do diâmetro e do comprimento da rosca do MI afeta a estabilidade é, portanto, fundamental para a prática clínica mais embasada cientificamente. Mesmo assim, embora a grande maioria dos ortodontistas tenha como objeto de análise a parte ativa da rosca, sabe-se que o comprimento do perfil transmucoso (PT) é outra característica geométrica básica relevante, porém que não vem recebendo a devida atenção. Uma vez que a espessura gengival é variável e interfere na escolha do comprimento do PT, entender como mini-implantes com diferentes comprimentos de PT se comportam mediante a mesma carga também é de singular importância.

Entendendo que parâmetros geométricos básicos como diâmetro e comprimento, seja da rosca ou da cabeça de MI, podem interferir na estabilidade, este estudo objetivou analisá-los detalhadamente. Além disso, uma vez que a grande maioria dos ortodontistas aplica carga precoce aos MI acreditando que a estabilidade primária se deve ao embricamento mecânico entre a rosca e o osso alveolar; os autores também estudaram a estabilidade considerando tais parâmetros geométricos em diferentes espessuras do osso cortical.

A quantidade e a qualidade do osso no local de inserção são fatores amplamente atribuídos e relacionados à estabilidade de MI (You *et al.*,1994; Butcher *et al.*, 2005; Brettin *et al.*, 2008; Jiang *et al.*, 2009; Petrey *et al.*, 2010). Duaibis *et al.* (2012) concordam e também destacaram que, além da quantidade, a qualidade do osso – devida à espessura da cortical ou ao módulo de elasticidade do osso medular – é fator igualmente importante. Além disso, destacaram que os comprimentos da rosca e da cabeça do MI têm relação direta com o aumento da estabilidade e concluíram que o comprimento da cabeça do MI pode afetar diretamente a tensão no osso cortical adjacente, também interferindo na estabilidade. Segundo os autores citados, os clínicos deveriam utilizar MI com os maiores diâmetros e os menores comprimentos de cabeça possíveis. Esta afirmação é corroborada por alguns autores afirmando que maiores braços de alavanca são criados pelo aumento da distância entre a cabeça do MI e a margem óssea adjacente, e que isso pode diminuir a estabilidade do MI (Petrey *et al*., 2010; Duaibis *et al*., 2012; You *et al*., 1994; Butcher *et al*., 2005), incluindo Liu *et al*. (2012) que declaram que a porção externa ao osso é o fator que realmente pode afetar o desempenho mecânico.

Este estudo apresentou conclusões semelhantes, porém corroboradas por análise mais criteriosa da influência da espessura do osso cortical, situação comumente observada na prática clínica.

Em todos os cenários avaliados, objetivou-se simular a distribuição e a magnitude da tensão máxima gerada após a aplicação de força em sistema osso-MI caracterizado por ausência de osseointegração e por aplicação de carga imediata. Desta forma, analisou-se a relação superficial entre o osso e o MI considerando o contato puramente mecânico.

O primeiro estudo analisou a variação do PT, correlacionando diferentes comprimentos de PT a diferentes espessuras do osso cortical. Como a literatura é extremamente escassa de trabalhos acerca do assunto, os autores avaliaram PT de comprimentos 1, 2, 3, e 4 mm.

Evidenciou-se a relação entre o aumento do comprimento do PT e o aumento da magnitude da tensão na região de interesse (RI). Este resultado pode ser explicado pelo aumento da distância do ponto de aplicação de força em relação à superfície óssea, que aumenta o braço de alavanca e, consequentemente, a tensão no osso adjacente (Petrey *et al*., 2010; Duaibis *et al*., 2012; You *et al*., 1994; Butcher *et al*., 2005). Vale ressaltar que se buscou mais que apenas comprovar a relação entre aumento do comprimento do PT e aumento da magnitude da tensão, observando como a espessura do osso cortical afeta tal aumento.

Portanto, à medida que a espessura do osso cortical aumentou, também assim ocorreu com o embricamento mecânico do sistema osso-MI, possibilitando melhor distribuição de tensão e diminuindo a tensão máxima para todos os PT avaliados. Notou-se que a magnitude de tensão e o deslocamento observado para a cabeça do MI em corticais com espessuras extremas (0,25 e 2,0 mm) foram bastante diferentes, especialmente para os comprimentos de 3 e 4 mm; para os menores comprimentos, tal diferença foi menos significativa. Desta forma foi possível concluir que, para MI com PT de maior comprimento deve-se sempre buscar locais de inserção com osso cortical mais espesso. Já para os MI com PT de menor comprimento, especialmente 1 mm, não foi observada diferença significativa na tensão máxima e no deslocamento máximo a partir de corticais com 1,0 mm de espessura. Logo, pode-se concluir que ao instalar MI com PT de 1 mm de comprimento não há necessidade de selecionar locais com cortical mais espessa que 1,0 mm.

A análise do MI com PT de 4 mm foi realizada objetivando melhor entender o efeito de braços de alavanca longos no osso cortical adjacente e, desta forma, analisar a capacidade de contenção, especialmente de corticais mais espessas. Entende-se que MI com PT de maior comprimento apresente uso clínico bastante restrito, sendo indicado possivelmente apenas para regiões com espessura gengival muito aumentada, como ocorre no palato. Para tal adequação de uso, o MI deveria ser mais espesso ou possuir parte ativa de maior comprimento. Este tipo de PT foi analisado com o objetivo de estudar melhor a estabilidade primária com relação ao osso cortical e evidenciar que, mesmo sob circunstâncias que aumentam significativamente a tensão no osso adjacente, como no caso de braço de alavanca longo, existe possibilidade de aplicação satisfatória. Até então isso não havia sido comprovado.

No segundo estudo, evidenciou-se a relação entre o aumento do diâmetro da rosca e o aumento da estabilidade. Tal resultado pode ser explicado pela existência de maior área de contato com o osso cortical e, consequentemente, diminuição da tensão no osso adjacente (You *et al.*,1994; Petrey *et al.*, 2010; Duaibis *et al.*, 2012; Butcher *et al.*, 2005). Desta forma, nos casos em que houver espaço adequado, os autores sugerem selecionar o MI com o maior diâmetro possível. Duaibis *et al.*, 2012 concordam, afirmando que os clínicos deveriam utilizar MI com os maiores diâmetros e os menores comprimentos de cabeça possíveis.

O MI com rosca de 1,2 mm de diâmetro apresentou magnitude de tensão e deslocamento bem superiores aos demais, e a este MI foi atribuído o pior desempenho mecânico. Os autores sugerem que MI delgados como este não deveriam ser utilizados. Suporte a esta recomendação foi oferecido por Pithon, Figueiredo e Oliveira (2013), que afirmaram que a redução do diâmetro pode diminuir a resistência mecânica, além de afetar a estabilidade primária, particularmente em situações nas quais o osso se caracteriza por maior densidade e espessura da cortical; além disso, Liu *et al.* (2012) recentemente utilizaram microscopia por fluorescência para analisar a ocorrência de trincas no osso cortical adjacente a MI com diferentes diâmetros e concluíram que o aumento do diâmetro da rosca não afeta a quantidade de trincas no osso adjacente.

Mais que simplesmente comprovar a relação entre o aumento do diâmetro da rosca e o aumento da estabilidade, o mais relevante nos presentes resultados foi apresentar indícios de desempenho mecânico semelhante para MI com diâmetros diferentes associados a corticais com a mesma espessura. Embora a análise de tensão tenha evidenciado a relação entre o aumento do diâmetro e o aumento da estabilidade, a análise de deslocamento da cabeça do MI evidenciou comportamento análogo entre os MI com diâmetros de 1,4 e 1,6 mm. A figura 4 do artigo 2 (página 66) mostra que os valores de deslocamento são bastante próximos, especialmente a partir da cortical com espessura de 1,25 mm. Os índices percentuais observados na tabela 3 do artigo 2 (página 69) reforçam esta observação, com diferença máxima de 11,07%. Mais estudos são necessários para melhor entender esta relação.

A despeito disso, contrapondo Liu *et al.* (2012), alguns autores têm afirmado que o aumento do diâmetro pode estar relacionado ao aumento de microfraturas no osso cortical durante a inserção (Martin, 2003; Huja *et al.*, 1999; Lee e Baek, 2010). Isso indicaria possível relação entre MI mais espessos e perda de estabilidade. Algum esclarecimento a esta questão poderia ser oferecido caso também tivesse sido realizada simulação dinâmica da penetração do MI no osso, que não foi incluída no presente estudo. Deste modo, as conclusões obtidas podem ser extrapoladas para a prática clínica somente nas situações em que não exista dano significativo ao osso adjacente durante a inserção. Singh *et al.* (2012) e Gracco *et al.* (2009) seguiram metodologia semelhante e também não realizaram simulação dinâmica em seus estudos com elementos finitos, defendendo seus resultados com base no argumento de Mano (2005), que afirmou que o tecido ósseo possui propriedades viscoelásticas e que, portanto, é capaz de se adaptar, dissipando tensões durante a inserção do MI.

A variação do comprimento da rosca foi analisada no terceiro estudo. A literatura é ampla em estudos destacando que o aumento do comprimento da rosca pode estar relacionado ao aumento da estabilidade (Lim *et al.*, 2008; Mortensen *et al.*, 2009; Gracco *et al.*, 2009, Petrey *et al.*, 2010; Jiang *et al.*, 2009; Pithon, Figueiredo e Oliveira, 2013; Chang *et al.*, 2012; Miyawaki *et al.*, 2003; Wilmes *et al.*, 2006; Kuroda *et al.*, 2007; Lim *et al.*, 2009; Chen *et al.*, 2007) e existe consenso entre os ortodontistas de que nos casos em que não seja possível usar MI mais espessos, deve-se adotar o critério de usar MI com roscas de maior comprimento.

Neste estudo observou-se que o MI com rosca de menor comprimento (6 mm) apresentou o pior desempenho mecânico, transferindo mais tensão ao osso adjacente e apresentando maior deslocamento da cabeça. Valores semelhantes encontrados para os demais comprimentos também comprovam a afirmação de Liu *et al.* (2012), e indicam que, a partir de determinado comprimento de rosca, que seja capaz de atenuar a ação do braço de alavanca, o aumento adicional deste comprimento não proporciona maior estabilidade. Neste caso, a utilização de MI com roscas de comprimento maior que 8 mm não é indicada, especialmente em situações nas quais a cortical é mais espessa, pois potencialmente geraria maior desgaste durante a inserção e não ofereceria estabilidade adicional.

Sob o ponto de vista puramente mecânico, a intuição física sugere que o osso medular de melhor qualidade favorece sua capacidade de contenção, auxiliando mais efetivamente o osso cortical a estabilizar o MI, especialmente os mais longos. Portanto, espera-se que a tensão na RI ou o deslocamento da cabeça do MI sejam diminuídos. Este fato foi observado, especialmente nos cenários caracterizadas por corticais mais delgadas, e pode ser validado pela figura 6 do artigo 3 (página 93). Nota-se a curva mais acentuada para as corticais com espessuras entre 0,25 e 1,0 mm quando o módulo de elasticidade foi menor. A partir da espessura de 1,0 mm, a os valores tenderam a se manter relativamente similares, comprovando a função coadjuvante do osso medular (mesmo com o aumento do módulo de elasticidade) e a função principal do osso cortical como para a estabilização. Assim sendo, entende-se que à medida que a espessura da cortical aumenta, também aumenta sua capacidade retentiva, e a contenção adicional oferecida pela medular se torna menos significativa.

Em todos os ensaios realizados, a tensão transferida ao osso e o deslocamento da cabeça do MI foram adotados como indicadores de estabilidade. A análise da tensão principal mínima corrobora as afirmações de Brettin *et al.* (2008), destacando que a diminuição de tensão no osso cortical está relacionada a maior estabilidade; e Yu *et al.* (2012), afirmando que a sobrecarga no osso adjacente pode resultar em falha do MI, caso a tensão supere o limite de resistência do osso. A utilização do deslocamento máximo da cabeça do MI como indicador secundário de estabilidade representa a função do osso cortical como apoio ao MI e já foi estudada por outros autores (Liu *et al.*, 2012; Singh *et al.*, 2012). Esta consideração é válida, uma vez que resultados semelhantes foram observados para a tensão principal mínima. Por conseguinte, os autores acreditam que MI que transfiram menor tensão ao osso adjacente ou que se movimentem menos sejam mais estáveis.

6 CONCLUSÃO

6.1 em relação ao comprimento do perfil transmucoso:

6.1.1. Observou-se relação direta entre aumento do tamanho do perfil transmucoso e diminuição da estabilidade.

6.2 em relação ao diâmetro da parte ativa da rosca:

6.2.1. Observou-se relação direta entre aumento do diâmetro da rosca e aumento da estabilidade.

6.2.2. O aumento da espessura da cortical também aumentou a estabilidade, especialmente para MI com diâmetros menores. Para MI com diâmetros maiores, corticais mais espessas que 1 mm podem oferecer pouca estabilidade adicional.

6.3 em relação ao comprimento da parte ativa da rosca:

6.3.1 Os autores identificaram relação entre aumento do comprimento da rosca e aumento da estabilidade do MI. Esta relação está fundamentada na capacidade da porção intraóssea do MI, representada pela rosca, equilibrar o braço de alavanca exercido pela porção extra-óssea, representada pelo perfil transmucoso e pela cabeça.

6.3.2. O MI com rosca de menor comprimento (6 mm) apresentou o pior desempenho mecânico. Os demais comprimentos apresentaram desempenho mecânico semelhante.

6.3.3. Mini-implantes com rosca de comprimento maior que 8 mm podem não oferecer estabilidade adicional caso a ancoragem seja monocortical.

7 RECOMENDAÇÕES

Diante dos resultados obtidos no presente trabalho, os autores sugerem a seleção de mini-implantes:

- Com perfil transmucoso de menor comprimento possível, especialmente nos casos com corticais mais delgadas. Mediante a necessidade de utilizar mini-implantes com perfis maiores (2 e 3 mm), deve-se tentar inseri-los em locais com cortical de no mínimo 1,5 mm de espessura. Mini-implantes com perfis de 1 mm podem ser utilizados em corticais com espessura a partir de 1,0 mm. Não se indicam os mini-implantes com perfis transmucosos de 4 mm.

- Com o maior diâmetro possível desde que haja espaço inter-radicular adequado. Os mini-implantes com diâmetro de 1,2 mm devem ser evitados sempre que possível.

- Com comprimento de rosca de 8 mm. Mini-implantes menores (6 mm) devem ser usados somente em regiões com cortical espessa (2,0 mm).

De maneira geral, os autores indicam a seleção de mini-implante com perfil transmucoso de 1 mm, rosca com diâmetro de 1.6 mm e comprimento de 8 mm.

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