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JOANA ESTEPHANY GORDILLO YÉPEZ

AS TENSÕES NO MINI-IMPLANTE E NO OSSO CIRCUNDANTE COM VARIAÇÕES NO PERFIL TRANSMUCOSO E COMPOSIÇÃO DO MINI-IMPLANTE: ESTUDO PELA ANÁLISE DE ELEMENTOS FINITOS

Curitiba
2015

JOANA ESTEPHANY GORDILLO YEPEZ, CD.

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Dissertação apresentada ao Programa de Pós-Graduação em Odontologia da Pontifícia Universidade Católica do Paraná, como parte dos requisitos para obtenção do título de Mestre em Odontologia, Área de Concentração em Ortodontia

Orientador: Prof. Dr. Orlando Tanaka

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DEDICATÓRIA

A Deus...

Por que o amor dele nunca acaba, suas misericórdias jamais terminam, seus planos são os melhores e mais bonitos do que todos os seus desapontamentos. Pela ótima família que tenho e pelo anjo que me enviou, minha filha Emily

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"É muito melhor lançar-se em busca de conquistas grandiosas, mesmo expondo-se ao fracasso, do que alinhar-se com os pobres de espírito, que nem gozam muito, nem sofrem muito, porque vivem numa penumbra cinzenta, onde não conhecem nem vitória, nem derrota."

Theodore Roosevelt

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Título:

As tensões no mini-implante e no osso circundante com variações no perfil transmucoso e composição do mini-implante: Estudo pela análise de elementos finitos.

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Resumo

Introdução: O presente estudo tem como objetivo analisar pelo método de elementos finitos as tensões geradas no mini-implante (MI) e no osso circundante pela aplicação de uma força perpendicular ao MI considerando variações de comprimento do perfil transmucoso e composição do MI. **Material e Métodos:** Mini-implantes com duas variações de perfil transmucoso (1 mm e 2 mm) foram modelados em CAD e analisados pelo método de elementos finitos com propriedades mecânicas do aço-inoxidável (ASTM F138) e titânio (ASTM F136). Uma força de 3,5 N (356,90 gf) foi aplicada perpendicularmente ao MI. Foram realizados 4 ensaios mecânicos: EM1– Perfil transmucoso de 1 mm e MI de aço; EM2– Perfil transmucoso de 1 mm e MI de titânio; EM3– Perfil transmucoso de 2 mm e MI de aço; EM4– Perfil transmucoso de 2 mm e MI de titânio. **Resultados:** As distribuições de tensões em todos os ensaios mecânicos se concentraram no MI, principalmente na região em contato com a cortical óssea. A maior tensão Von Mises foi observada no EM4, seguido pelo EM3, EM2 e por último o EM1. **Conclusões:** A concentração maior de tensões ocorreu, portanto, no MI com perfil transmucoso de 2 mm e no MI de titânio, quando comparado ao MI de aço inoxidável na mesma comprimento de perfil transmucoso.

Palavras-chave: Implantes Dentários, Procedimentos de Ancoragem Ortodôntica, Análise de Elementos Finitos.

Introdução

A Ortodontia tem se desenvolvido rapidamente nos últimos anos, e atualmente existem vários dispositivos no mercado que auxiliam o ortodontista no tratamento ortodôntico, facilitando assim a terapêutica ortodôntica, diminuindo o tempo de tratamento e dispensando a colaboração do paciente.[1]

A ancoragem é o fator determinante para o sucesso ou insucesso na aplicação biomecânica. Na busca para otimizar a ancoragem os dispositivos de ancoragem esquelética temporária são denominados com diferentes termos: mini-implante (MI), micro-implantes, MI ortodônticos, miniparafusos e microparafusos[1]. O termo Dispositivo de Ancoragem Temporária (DAT) refere-se a "todas as variações de implantes que são instalados para promover ancoragem ortodôntica e são removidos após a biomecânica".[1,2] No entanto, apesar de não haver consenso do ponto de vista científico, o termo mais adequado parece ser MI. [1]

Os MI surgiram na busca de uma ancoragem adequada para os tratamentos ortodônticos.[3,4] O procedimento de instalação e remoção é de baixa complexidade, baixo custo e confortável para o paciente.[5-11] Seu pequeno tamanho permite instalação em vários locais do osso alveolar, inclusive no espaço interradicular.[12,13] Sua composição química é de titânio grau 5 (Ti6Al-4V)[14,15] e são fabricados com a liga ASTM F136. Esta liga possui características bioativas inferiores ao titânio puro e maior resistência do que o titânio puro. A utilização deste sistema é baseada na estabilidade mecânica primária, e não na secundária, advinda da osteointegração.[16] Também existem alguns MI comerciais que são fabricados com aço inoxidável 316 grau II (ASTM F138).[14] Os dois apresentam uma resistência mecânica maior ao titânio puro comercialmente usado na fabricação dos implantes dentários osseointegráveis.[14]

Os MI possuem a forma de parafusos, e podem ser divididos em três partes distintas:

a) Cabeça - parte de acoplamento dos dispositivos ortodônticos exposta clinicamente,

b) Perfil transmucoso - área compreendida entre a porção intraóssea e a cabeça do parafuso, onde ocorre a acomodação do tecido mole peri-implantar,

c) Ponta ativa - porção intraóssea correspondente às roscas do MI[2,14].

Os MI apresentam diferentes tamanhos, formas, diâmetros e perfil transmucoso e estas variações podem comprometer o seu desempenho. A forma do MI deve prover o travamento mecânico do MI no osso que, influencia diretamente a estabilidade do mesmo. Além disso, deve permitir a distribuição de cargas de maneira que não prejudique a fisiologia óssea e, limitar o trauma apenas ao momento da inserção.[16-18]

A simulação de diversas situações clínicas por meio do método de elementos finitos (MEF) pode ser utilizado para se avaliar a distribuição de tensões nas áreas no osso circundante[19]. No entanto, nenhum estudo investigou a distribuição das tensões no MI com diferentes comprimentos de perfil transmucoso e diferentes materiais, o conhecimento da distribuição das tensões auxiliaria na obtenção de maior resistência e estabilidade dos MI.

Portanto, o objetivo deste trabalho foi analisar pelo MEF as tensões geradas pela aplicação de uma força perpendicular no MI e no osso circundante, de acordo com variações de:

a) Comprimento de perfil transmucoso (1 mm e 2 mm);

b) Composição do MI (Ti6Al-4V e aço inoxidável 316)

Material e Métodos

O MI da marca Morelli (Morelli, SP), de 1,5mm de diâmetro, 8mm de comprimento com 1 mm e 2 mm de perfil transmucoso (Morelli 37.10.102 e 37.10.202) (Figura 1) foram tridimensionalmente modelados em um aplicativo de CAD (Computer Aided Design) para criação de modelos sólidos com base nas medidas fornecidas pelo fabricante. Foram atribuídas propriedades mecânicas de dois materiais para os MI: titânio (ASTM F136) e aço inoxidável (ASTM F138) Tabela 1.

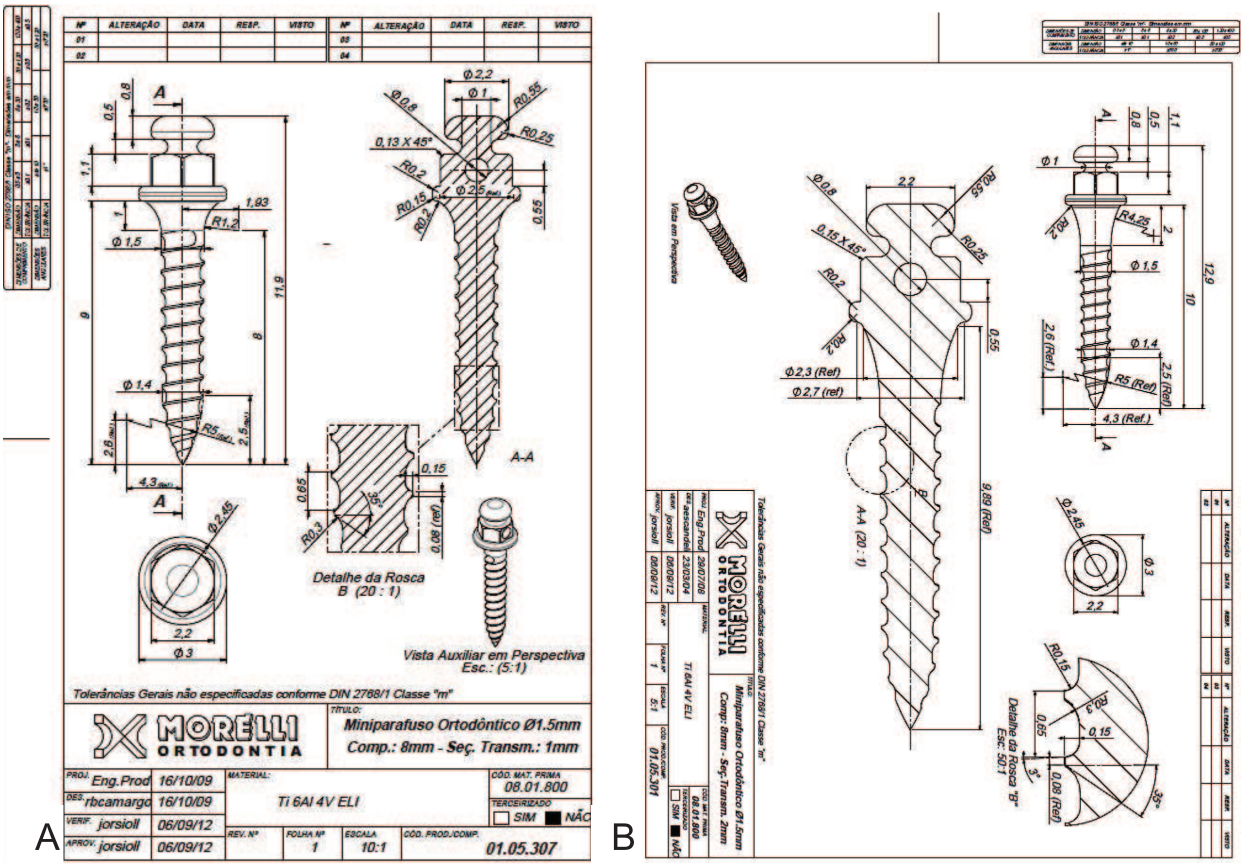

Fig 1. Desenho dos MI **A,** MI perfil transmucoso 1 mm; **B,** MI perfil transmucoso 2

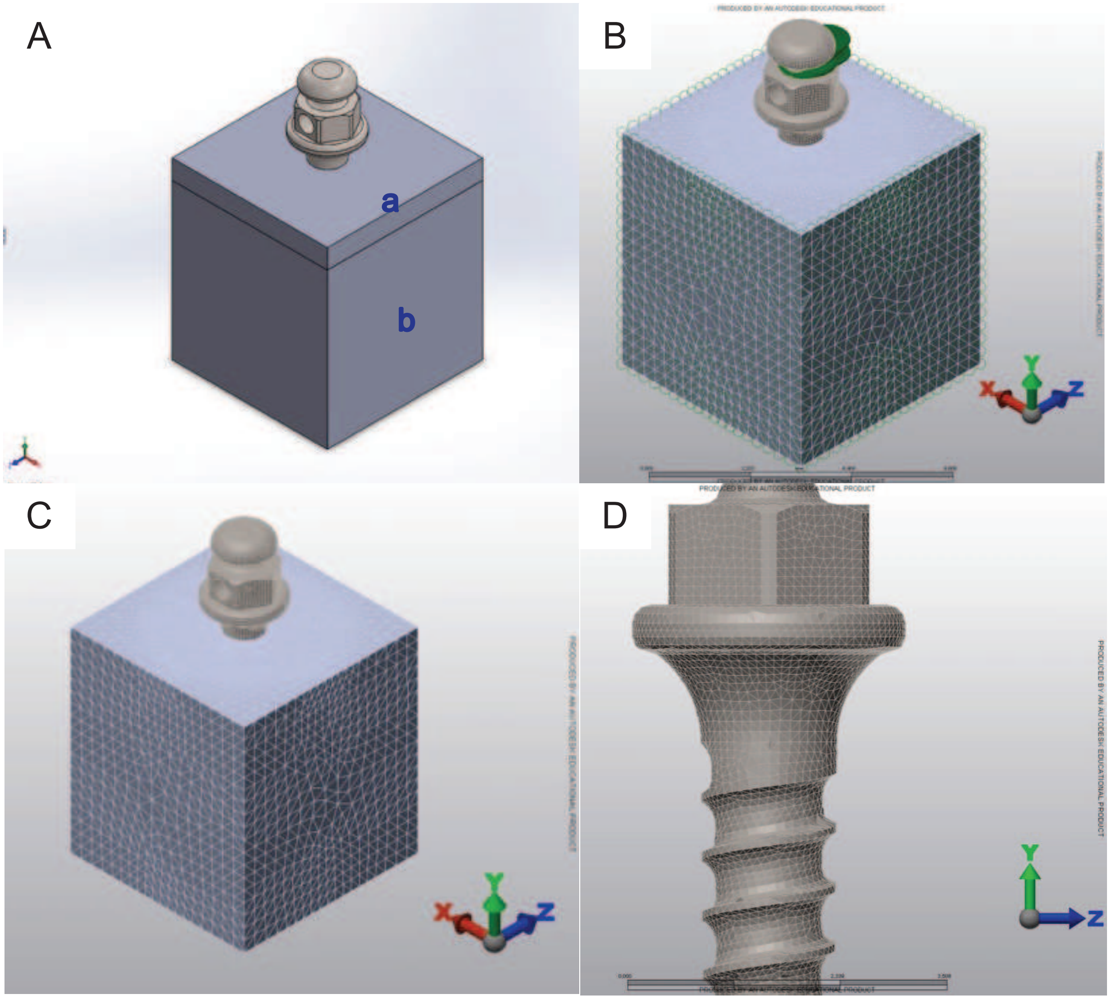

Fig 2. **A,** Modelagem tridimensional utilizada para os EM: a) osso cortical de 1mm, b) osso trabecular; **B**, Condições de contorno vista tridimensional e aplicação da carga na cabeça do MI; **C,** Malha tetraédrica tridimensional; **D,** detalhe da malha do MI

Tabela I. Propriedades dos materiais usados para a analise de elementos finitos

Material	*Módulo de Young (MPa)*	*Coeficiente de Poisson*
Titânio	110.000[a]	0.33[a]
Aço Inoxidável	205.000[b]	0.29[b]
Osso Cortical	13.800[c]	0.26[c]
Osso Trabecular	345[c]	0.31[c]

[a]:Suzuki et al[20]; [b]Kojima e Fukui[21]; [c]Jones et al[22]

Todos os modelos computacionais do MI foram inseridos em blocos de formato cuboide, com 9 mm de altura e 8 mm de largura e profundidade, que foram modelados tridimensionalmente no mesmo programa. Os blocos foram constituídos de duas camadas, as quais foram atribuídas propriedades de comportamento correspondentes ao osso cortical e trabecular (Figura 2A). A camada correspondente ao osso cortical foi fixada em 1 mm. A interface entre o osso e o MI era de contato perfeito. O modelo sólido completo é transferido para o programa de elementos finitos Autodesk Simulation Multiphysics® 2013.

Sobre a geometria do modelo, foi gerada uma malha de elementos finitos composta de elementos tetraédricos lineares de alta ordem (primeira ordem). Após uma analise de convergência do campo de tensões definiu-se a malha para realização das analises de MEF. Os comprimentos das arestas dos elementos resultantes desta análises variam de 0,375 mm a 0,500 mm (Figura 2C e D). O bloco apresentou maior refino de malha na região do osso circundante ao MI em um raio aproximado de 3 mm (Figura 3A e B). Assim, a malha final do EM1 e EM2 foi composto de 416.889 elementos e 81.421 nós, enquanto o modelo final do EM3 e EM4 foi composto de 577.429 elementos e 109.645 nós. Nas faces laterais do cuboide, como condições de contorno, foram restringidas as translações nas direções x, y, z, representadas por círculos verdes (Figura 2C). Estas condições fazem-se necessárias para simular o restante dos ossos maxilares. O carregamento de 3,5 N foi aplicado na cabeça do MI na direção do eixo z, representados por linhas verdes na cabeça do MI (Figura 2B). Esta magnitude representa uma força real que é aplicada clinicamente. [23]

Considerou-se todos os materiais (osso cortical, osso trabecular, MI de titânio e MI de aço inoxidável) como tendo comportamento homogêneo, isotrópico e linearmente elásticos; com módulo de Young e coeficiente de Poisson específicos (Tabela 1). Além disso, o estudo do campo de tensões e deformações da análise de elementos finitos ocorreu dentro do regime elástico de tensões.

Os ensaios mecânicos foram divididos com as seguintes combinações de comprimento do perfil transmucoso e material do MI:

1) EM1- MI com perfil transmucoso de 1 mm de aço inoxidável.

2) EM2- MI com perfil transmucoso de 1 mm de titânio.

3) EM3- MI com perfil transmucoso de 2 mm de aço inoxidável.

4) EM4- MI com perfil transmucoso de 2 mm de titânio.

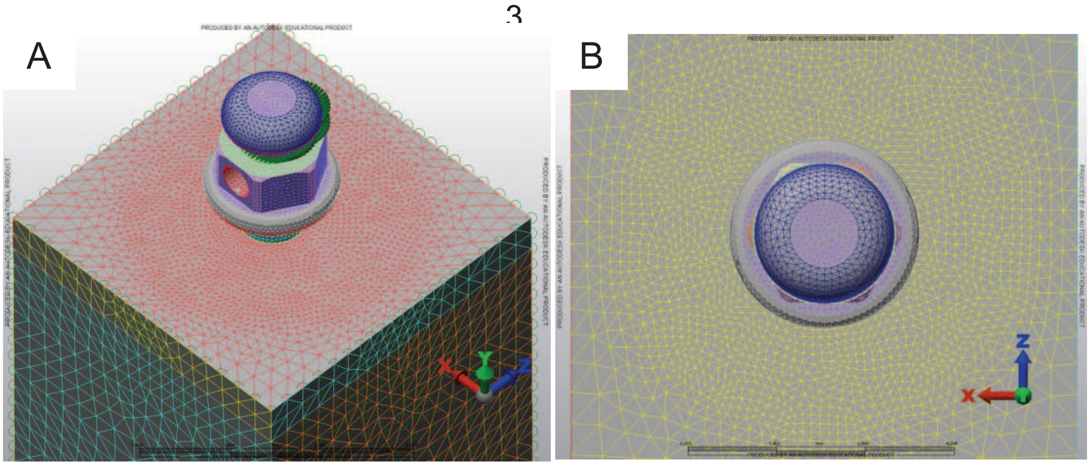

Fig 3. A, Refinamento da malha na região do osso circundante ao MI; **B,** Detalhe do refinamento da malha vista superior

Enquanto a coordenadas, o sistema da coordenadas foi determinado pelo programa Autodesk Simulation Multiphysics®.

O campo de tensões sobre o modelo de elementos finitos, é avaliado de acordo com a teoria da energia de distorção para materiais dúcteis, também conhecido como critério de falha de von Mises.[24,25] Estas tratativas vem sendo utilizada no campo da ortodontia. [20,26,27]

Resultados

1. *Avaliação das tensões*

Foram avaliadas a distribuição das tensões no bloco ósseo e no MI por meio de uma escala de cores. As cores que tendem ao azul indicam regiões com tensões baixas e as cores que tendem ao vermelho, referem-se a locais com tensões altas. Além disso, estas regiões indicam regiões concentradoras de tensões.

Observou-se que a distribuição das tensões na interface do osso e do MI em todos os ensaios mecânicos se concentraram na região do MI que se encontra em contato com a cortical óssea, e também na região do MI que se encontra em contato com o osso trabecular próximo ao osso cortical. Igualmente, em todos os carregamentos houve distribuição de tensões na região do MI localizado fora do osso logo acima da cortical óssea.

Os valores máximos de tensão de Von Mises foi encontrado no EM4 (76,94 MPa), seguido pelo EM3 (68,38 MPa), depois EM2 (53,09 MPa) e por último EM1 (43,64 MPa).

Tabela II. Valores máximos e mínimos de tensões de von Mises para os 4 modelos de carregamento

	EM1 (MPa)	*EM2 (MPa)*	*EM3 (MPa)*	*EM4 (MPa)*
MAX	43,64	53,09	68,38	76,94
MIN	0	0	0	0

Legendas: MAX (valor máximo de tensão); MIN (valor mínimo de tensão); EM (ensaios mecânicos)

1) MI com perfil transmucoso de 1 mm de aço inoxidável (EM1)

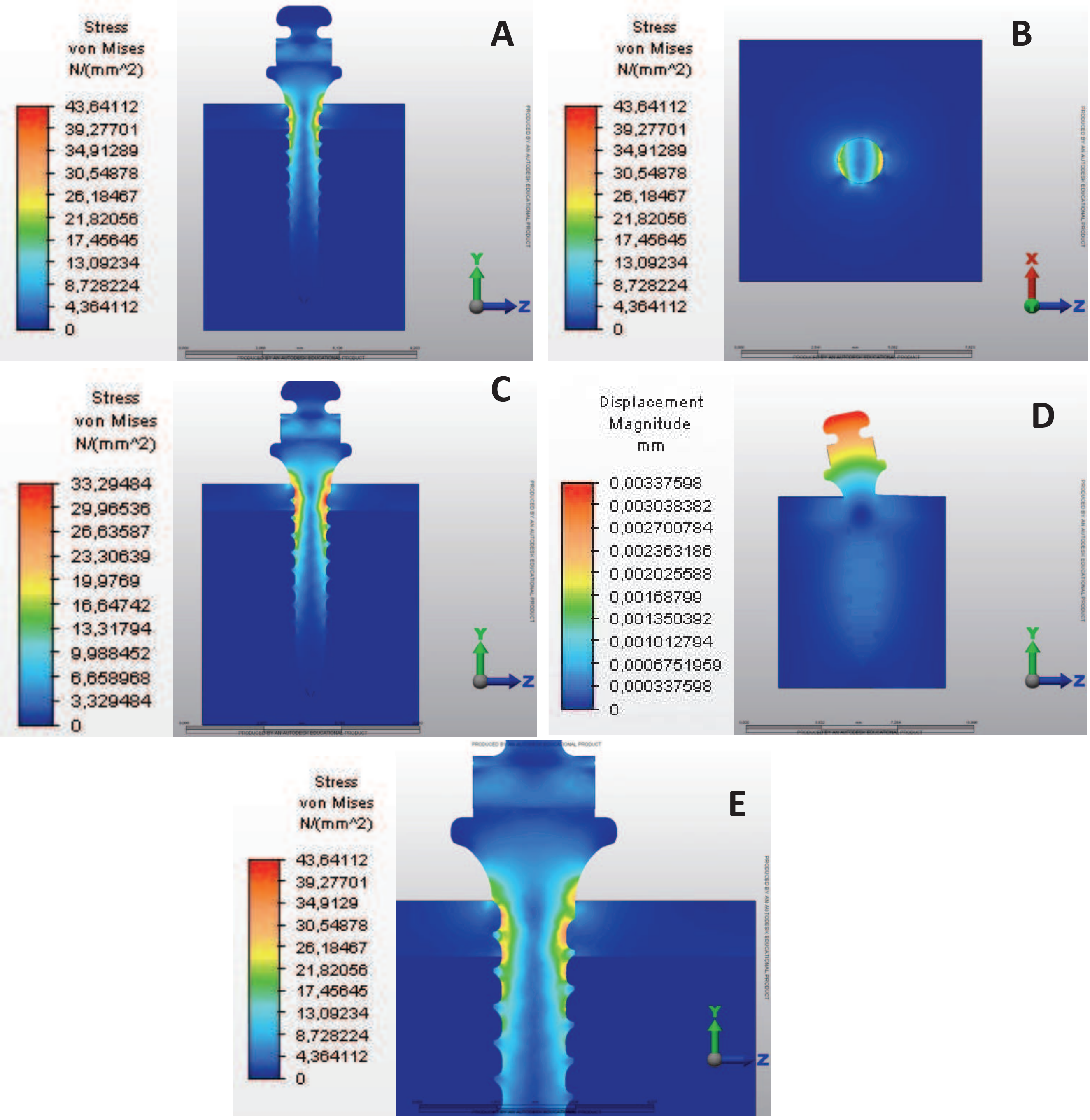

Fig 4. Distribuição das tensões para **EM1. A,** Corte transversal; **B,** Corte perpendicular; **C,** Corte transversal com menor escala dos valores máximos encontrados em todos ensaios mecânicos; **D,** Corte transversal com deflexão (escala: magnitude de deflexão); **E,** Detalhe da distribuição das tensões na região da espessura corresponde à cortical óssea.

2) MI com perfil transmucoso de 1 mm de titânio (EM2)

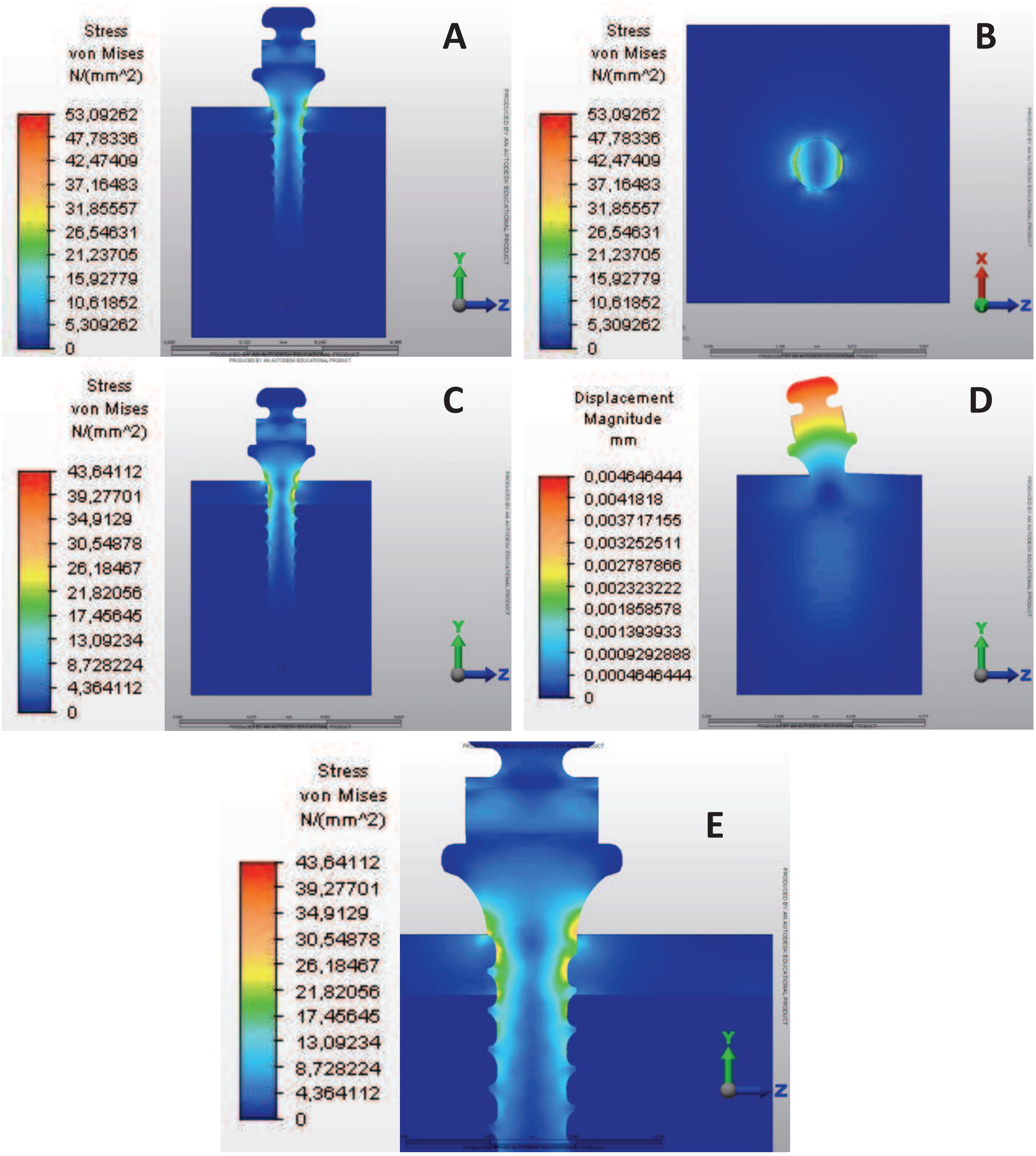

Fig 5. Distribuição das tensões para **EM2**. **A,** Corte transversal; **B,** Corte perpendicular; **C,** Corte transversal com menor escala dos valores máximos encontrados em todos ensaios mecânicos; **D,** Corte transversal com deflexão (escala: magnitude de deflexão); **E,** Detalhe da distribuição das tensões na região da espessura corresponde à cortical óssea

3) MI com perfil transmucoso de 2 mm de aço inoxidável (EM3)

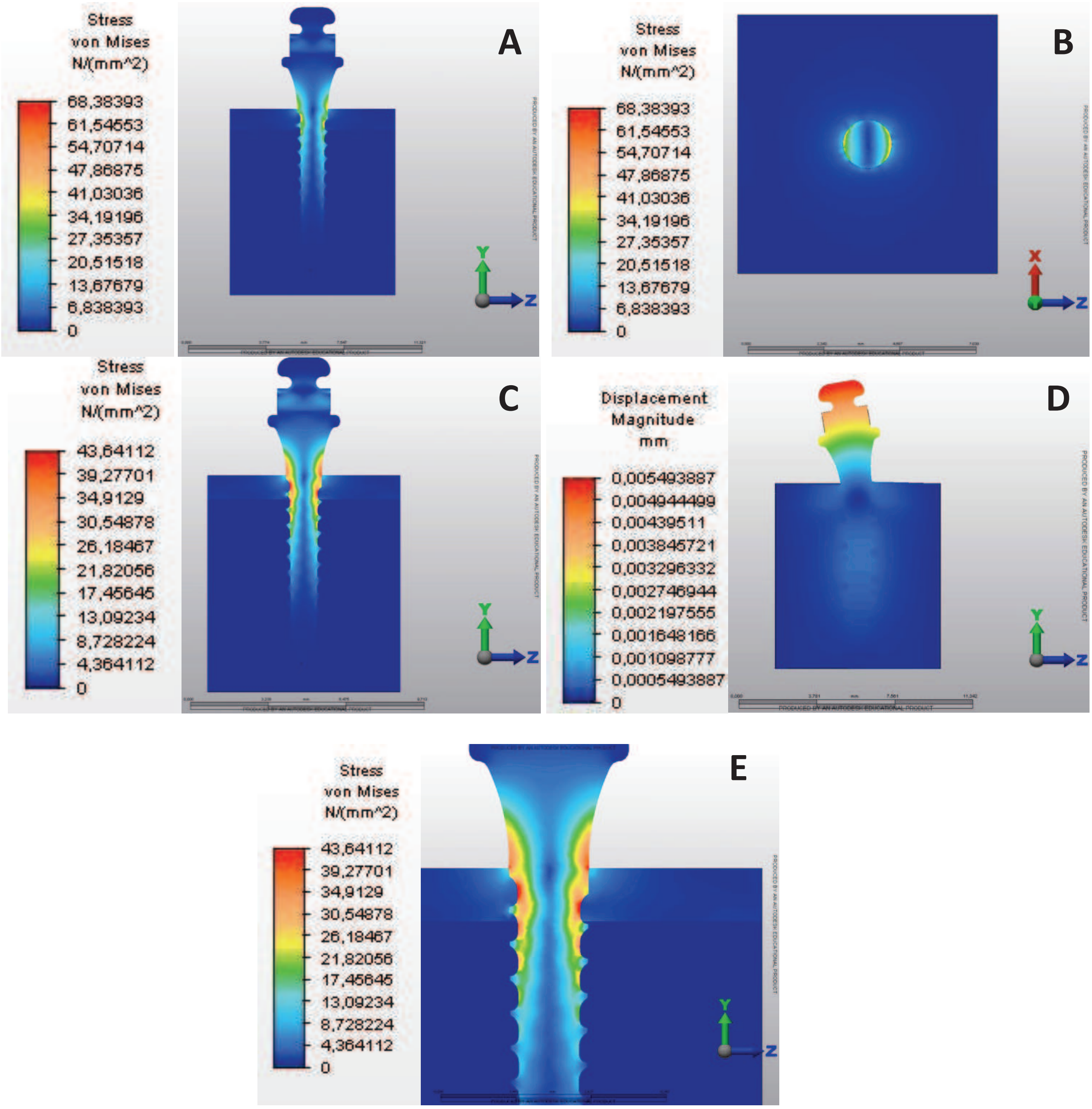

Fig 6. Distribuição das tensões para **EM3. A,** Corte transversal; **B,** Corte perpendicular; **C,** Corte transversal com menor escala dos valores máximos encontrados em todos ensaios mecânicos; **D,** Corte transversal com deflexão (escala: magnitude de deflexão); **E,** Detalhe da distribuição das tensões na região da espessura corresponde à cortical óssea

4) MI com perfil transmucoso de 2 mm de titânio (EM4)

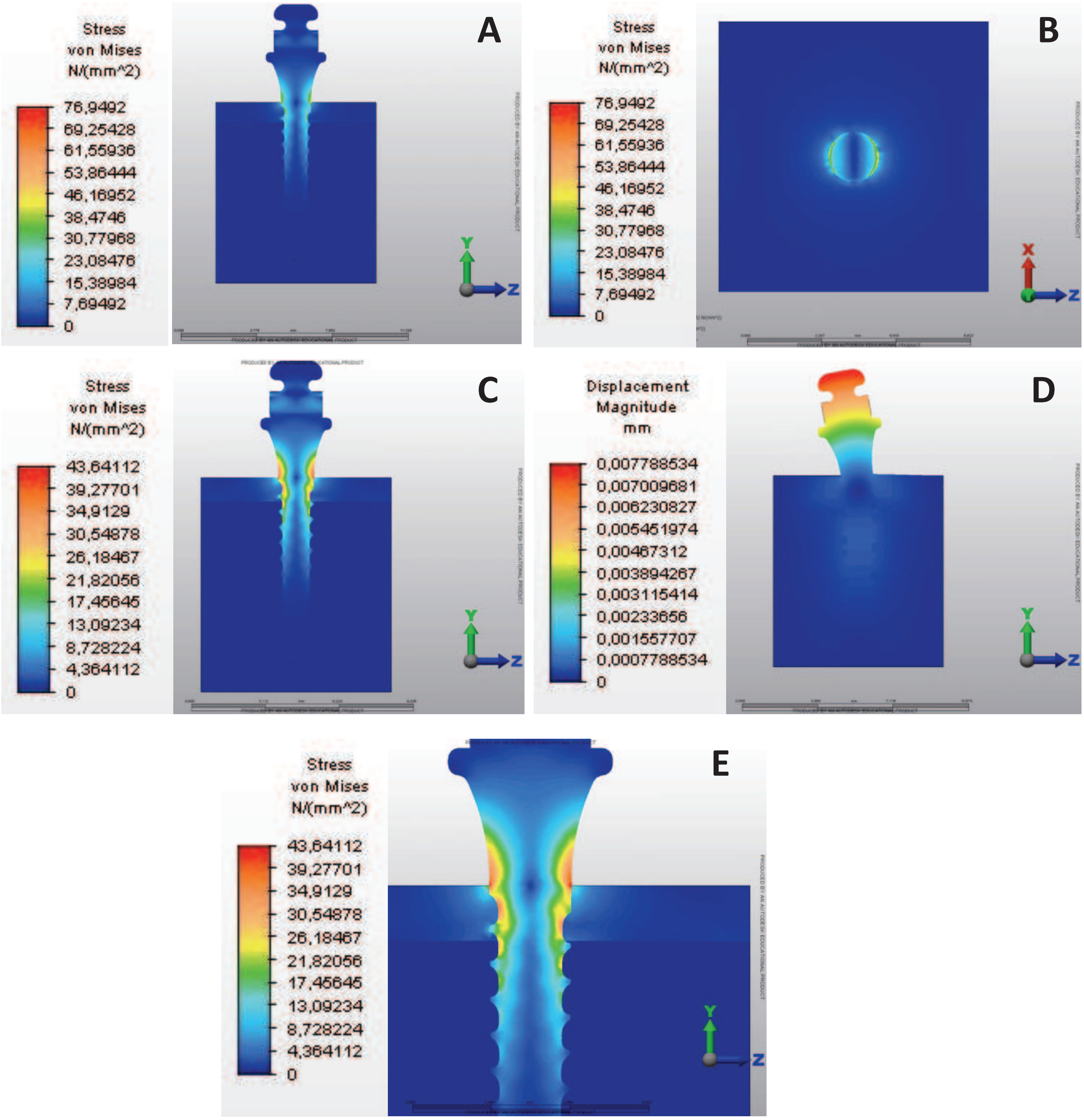

Fig 7. Distribuição das tensões para **EM4**. **A,** Corte transversal; **B,** Corte perpendicular; **C,** Corte transversal com menor escala dos valores máximos encontrados em todos ensaios mecânicos; **D,** Corte transversal com deflexão (escala: magnitude de deflexão); **E,** Detalhe da distribuição das tensões na região da espessura corresponde à cortical óssea

2. *Composição do material (Ti-6Al-4V e aço inoxidável 316)*

A maior tensão de von Mises foi encontrada nos MI feitos de titânio (EM2 53,09 MPa; EM4 76,94 MPa) em comparação com o MI de aço inoxidável (EM1 43,64 MPa; EM3 68,38 MPa).

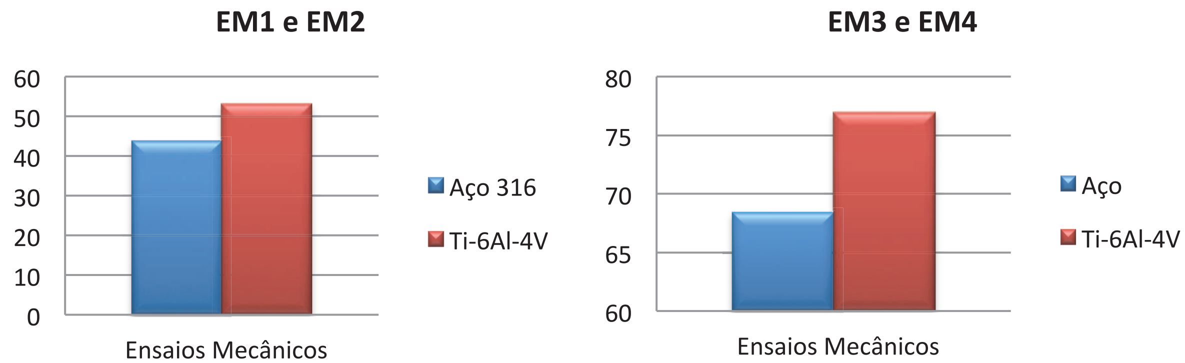

Fig 8. Valores máximos de von Mises de acordo com a composição do material

3. *Perfil transmucoso*

A maior tensão de von Mises foi encontrado no MI com perfil transmucoso de 2 mm (EM3 68,38 MPa; EM4 76,94 MPa) em comparação com o MI com perfil transmucoso de 1 mm (EM1 43,64 MPa; EM2 53,09 MPa)

4. *Deflexão*

A maior deflexão foi encontrada no EM4 (7,78 μm), seguido pelo EM3 (5,49 μm) que correspondem a MI com perfil transmucoso de 2 mm, seguido por EM2 (4,64 μm) e por ultimo EM1 (3,37 μm) que são MI com perfil transmucoso de 1 mm.

Discussão

Modelos de elementos finitos

O MEF é um método numérico que auxilia a solucionar problemas complexos em diversos campos da ciência, dividindo domínios complexos em pequenos domínios. Este método representa tanto a deformação como a distribuição de tensões tridimensionalmente, de corpos que estão expostos a carregamentos, sendo útil para simular a distribuição de tensões nas áreas das ciências biológicas e médicas.[20,28] Os resultados dos estudos com MEF fornecem informações importantes que ajudam a compreender reações biológicas complexas com maior precisão em estudos que são difíceis de desenvolver em humanos devido as grandes variações entre as amostras.

A eficácia do MEF depende do tipo de elemento e do grau de refino de malha. Neste estudo, o modelo final da malha dos EM1 e EM2 foi composto de 416.889 elementos e 81.421 nós, enquanto o modelo final do EM3 e EM4 foi composto de 577.429 elementos e 109.645 nós. Além disso, este estudo apresentou limitações na simulação. A geometria do bloco, simulando um bloco osso, foi um bloco cuboide para sua simplificação, as propriedades do material foram consideradas homogéneas, isotrópicas e linearmente elástico. O tecido mole não foi simulado pois seu impacto teria sido insignificante. Igualmente, o MI foi colocado no bloco ósseo perpendicular à superfície do osso e em perfeitas condições, e a interface entre o osso e o MI era de contato perfeito. No entanto, essas simplificações não devem alterar as analises do campo de tensões quantitativas das simulações numericas.[29]

Nos resultados deste estudo, observou-se que a maior tensão de von de Mises se concentrou na região do MI que se encontra em contato com a cortical óssea. Igualmente, em todos os carregamentos houve distribuição de tensões na região do MI localizado fora do osso logo acima da cortical óssea. Estes resultados estão em concordância com o estudo de Suzuki et al.[20] onde a tensão máxima do MI concentrou-se entre a superfície do osso de suporte e o MI.

Perfil transmucoso do MI relacionado com a tensão

O perfil transmucoso tem como objetivo manter a saúde dos tecidos peri-implantares, principalmente em áreas com pequena faixa de gengiva inserida, pois a inflamação é fator que contribui para o insucesso do MI.[30] Os resultados indicaram que a tensão máxima foi significativamente menor nos MI com perfil transmucoso de 1 mm do que nos MI com perfil transmucoso de 2 mm. Não existem estudos comparando comprimentos de perfil transmucoso, no entanto o resultado tem relação com o conceito de momento fletor causado por uma força. O momento é o efeito produzido pelo braço de alavanca de força em relação ao vínculo gerando uma tendência de rotação em um eixo transversal ao da linha de aplicação da foça na região do vínculo.[31-33]

Seguindo este princípio, quanto maior a distância do ponto de aplicação da força, maior será momento; consequentemente maior as tensões. Assim, os MI com perfil transmucoso de 2 mm possuem um braço de alavanca maior que os MI de perfil transmucoso de 1 mm. Sendo aplicada a mesma força, o resultando foi maior momento nos MI com perfil transmucoso de 2 mm, e consequentemente maior tensão de flexão.

O estudo de Lim et al,[34] com MEF e diferentes comprimentos de MI (6, 8, 10 e 12 mm), concluíram que quanto maior a exposição do comprimento do MI, maior será a tensão no osso cortical. Outro estudo de Nova et al, [13] compararam MI com e sem perfil transmucoso de duas marcas comerciais diferentes e concluíram que a presença ou ausência do perfil transmucoso parece não afetar os torques de inserção ou remoção, uma vez que somente uma marca (NEODENT) apresentou valores significativos, enquanto a outra marca (SIN), não apresentou diferenças. Outros estudos avaliaram o efeito do comprimento do MI e apresentavam resultados inconsistentes ou inconclusivos.[35,36] O motivo é que comprimento do MI não é um fator dominante na distribuição das tensões. O comprimento do MI exposto fora do osso, a distância do braço de alavanca do momento, são os fatores que influenciam na distribuição das tensões e deflexão. Por conseguinte, tanto o comprimento do MI e a profundidade da implantação deve ser considerados.[29]

Outros fatores como o diâmetro, a inflamação, infecção, inserção em locais não-queratinizados e o tamanho reduzido do MI também podem afetar na estabilidade do MI.[37] Miyawaki et al.[37] e Lim et al.[34] relataram que o diâmetro afetou as taxas de sucesso, mas Park et al.[7] relataram que não teve nenhum efeito. Miyawaki et al. em 2003, examinaram a taxa de sucesso de três tipos de

MI de titânio e os fatores associados à estabilidade dos MI usados e concluíram que MI com diâmetro maior de 1 mm tem maior sucesso.

Deformação do MI

Sempre que uma força é aplicada a um corpo, esta tende a mudar a forma e o tamanho dele. Essas mudanças são dominadas deformações e podem ser altamente visíveis ou praticamente imperceptíveis.[38] As deformações sofridas pelos MI com a aplicação da força perpendicular, neste estudo, foram leves e imperceptíveis.

Deflexão do MI

A região superior do MI foi o local de maior deflexão, como pode-se observar nas Figuras 4, 5, 6, 7 D (Pag. 8, 9, 10, 11). A maior deflexão ocorreu nos ensaios mecânicos com MI de perfil transmucoso de 2 mm do que em MI com perfil transmucoso de 1 mm. Neste estudo, a deflexão foi mínima e imperceptíveis.

Materiais do MI relacionado com a tensão

Neste estudo foi usado MI de dois materiais distintos, um de aço inoxidável (ASTM F138) e outro de titânio (ASTM F136). Aços inoxidáveis utilizados em implantes dentários precisam ter propriedades mecânicas e físicas adequadas para seu uso como baixa permeabilidade, alta resistência, baixo teor de impurezas, resistência à corrosão e variedade de técnicas de esterilização.[39,40] O titânio (Ti-6Al-4V) é o mais conhecido por possuir duas fases: uma fase alfa de estruturas cristalinas hexagonal compacta e outra fase beta que é cúbica de corpo centrado as quais se fazem presentes a temperatura ambiente, combinando resistência à corrosão, resistência mecânica específica alta, boas propriedades para altas temperaturas, conformabilidade e usinabilidade.[40,41] Embora o aço apresenta maior resistência à fratura que o titânio, a qualidade “performance” dos MI de aço poderia ser menor do que a dos de titânio.[42]

Os resultados do presente estudo mostram que a tensão máxima de von Mises foi encontrado nos MI feitos com liga de titânio Ti-6Al-4V em comparação com os MI de aço inoxidável 316. A magnitude do campo de tensões é maior no MI fabricado com liga de titânio porque a sua rigidez é aproximadamente 47%

menor que a do MI do aço inoxidável 316. Para carregamento de flexão a rigidez está diretamente relacionada com o modulo de Young.[33,43]

Considerações finais

A abordagem experimental em humanos possui limitações biológicas, éticas e de privacidade; por outro lado, o MEF proporciona uma abordagem mais manejável para a avaliação da biomecânica dentária.[29,44] Portanto, sugere-se que o presente modelo baseado em trabalhos anteriores de simulação proporciona os mesmos padrões de distribuição de tensões no MI e no osso circundante, e pode ser uma ferramenta, clinicamente, útil para estimar o efeito da distribuição de tensões e, consequentemente, prever a reação do tecido contra a força ortodôntica. Para Lim et al[34] a tensão óssea excessiva induzida por maiores comprimentos de exposição do MI permite a reabsorção óssea local.

Conclusão

Neste estudo ao se analisar as tensões geradas no MI e no osso circundante pela aplicação de uma força perpendicular ao MI de acordo com variações de comprimento de perfil transmucoso (1 mm e 2 mm) e composição do MI (Ti-6Al-4V e aço inoxidável 316) pelo método de elementos finitos. Por tanto, concluiu-se que:

1. O valor mais alto de tensão de Von Mises foi encontrado no EM4, seguido pelo EM3, depois EM2 e por último EM1;
2. As tensões estão mais concentradas no MI mais do que no osso;
3. As tensões foram menores nos MI fabricados de aço inoxidável que nos de titânio;
4. MI com perfil transmucoso de 2 mm mostraram maiores tensões que os MI com perfil transmucoso de 1 mm.

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ARTIGO EM INGLÊS

TITLE PAGE

Stress in the mini-screw and surrounding bone according to the transmucosal profile and mini-screw composition: A finite element analysis

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Abstract

Introduction: This study aimed to analyze the finite element method (FEM) stress generated in the mini-screw (MS) and surrounding bone upon applying a force perpendicular to the MS with varying transmucosal profile lengths and compositions. **Methods:** Mini-screws with two transmucosal profile lengths (1 mm and 2 mm) were modeled in using Computer Aided Design (CAD) and analyzed by FEM using the mechanical properties of stainless steel (ASTM F138) and titanium (ASTM F136). A 3.5-N (356.90 gf) force was applied perpendicularly to the MS during four mechanical tests as follows: EM1, 1-mm transmucosal profile and stainless steel MS; EM2, 1-mm transmucosal profile and titanium MS; EM3, 2-mm transmucosal profile and stainless steel MS; and EM4, 2-mm transmucosal profile and titanium MS. **Results:** The stress was most concentrated at the MS in all mechanical tests, especially at the region contacting the cortical bone. The von Mises voltage was highest in EM4, followed by EM3, EM2, and EM1. **Conclusions:** The highest stress was observed in MS with a 2-mm transmucosal profile and titanium composition compared with a stainless steel MS and identical transmucosal profile.

Keywords: dental implant, orthodontic anchorage procedures, finite element analysis

Introduction

Orthodontics has developed rapidly in recent years, and several devices are currently available to assist the orthodontist in performing treatment, thus facilitating therapy, reducing treatment time, and minimizing patient involvement.[1]

Anchoring is the determinant factor for the success or failure of biomechanical application. A variety of temporary anchorage devices (TADs) are used to optimize skeletal anchorage devices, including mini-screws (MS), micro-implants, orthodontic mini-screws, and micro-screws.[1] TAD is defined as "all the variations of implants that are installed to promote orthodontic anchorage and are removed after the biomechanics".[1,2] From a scientific point-of-view, the most appropriate term is MS, although there is no consensus.[1]

The MS was developed during the search for a suitable anchorage for orthodontic treatment.[3,4] Installation and removal is simple, inexpensive, and comfortable for patients.[5-11] Its small size allows installation in various locations of the alveolar bone, including the interradicular space.[12,13] The screws are manufactured from titanium grade 5 (Ti6Al-4V)[14,15] and alloy ASTM F136. This alloy has bioactive characteristics inferior than pure titanium and greater resistance than pure titanium. This system relies on the primary mechanical stability arising from osseointegration, rather than secondary stability.[16] A few MS are also manufactured from stainless steel 316 grade II (ASTM F138).[14] These devices have a higher mechanical resistance when used in osseointegrated dental implants compared with commercially pure titanium.[14]

MS comprise three distinct parts as follows:

a) Head design: couples to the clinically exposed orthodontic device

b) Transmucosal profile: area between the intraosseous portion and screw head that accommodates the peri-implant soft tissue

c) Long position: intraosseous portion corresponding to the MS threads.[2,14]

MS have different sizes, shapes, diameters, and transmucosal profiles, and these variations may compromise its performance. The MS form must provide mechanical locking into bone, which directly influences the implant stability. It should also ensure that the load distribution is not detrimental to the bone physiology and limits trauma beyond that occurring during insertion.[16-18]

Various clinical scenarios can be simulated using the finite element method (FEM) to evaluate the stress distribution in the areas surrounding bone.[19] However, the stress distribution in MS with different transmucosal profiles has not been examined, nor has the stress distribution in mini-implants with different profile lengths, transmucosal profiles, and materials. An understanding of the stress distribution would help in achieving greater strength and stability of MI. Therefore, the objective of this study was to analyze, using FEM, the stress that are generated when a perpendicular force is applied to the MS and surrounding bone, according to changes in the transmucosal profile length (1 mm and 2 mm) and MS composition (Ti6Al-4V and stainless steel 316).

Material and Methods

An MS (Morelli, SP) measuring 1.5 mm in diameter and 8 mm in length with a 1-mm to 2-mm transmucosal profile (Morelli 37.10.102 and 37.10.202; Figure 1) was three-dimensionally simulated in a Computer Aided Design (CAD) application for creating solid models based on the measurements provided by the manufacturer. The MS was assigned the mechanical properties of two materials, titanium (ASTM F136) and stainless steel (ASTM F138) Table 1.

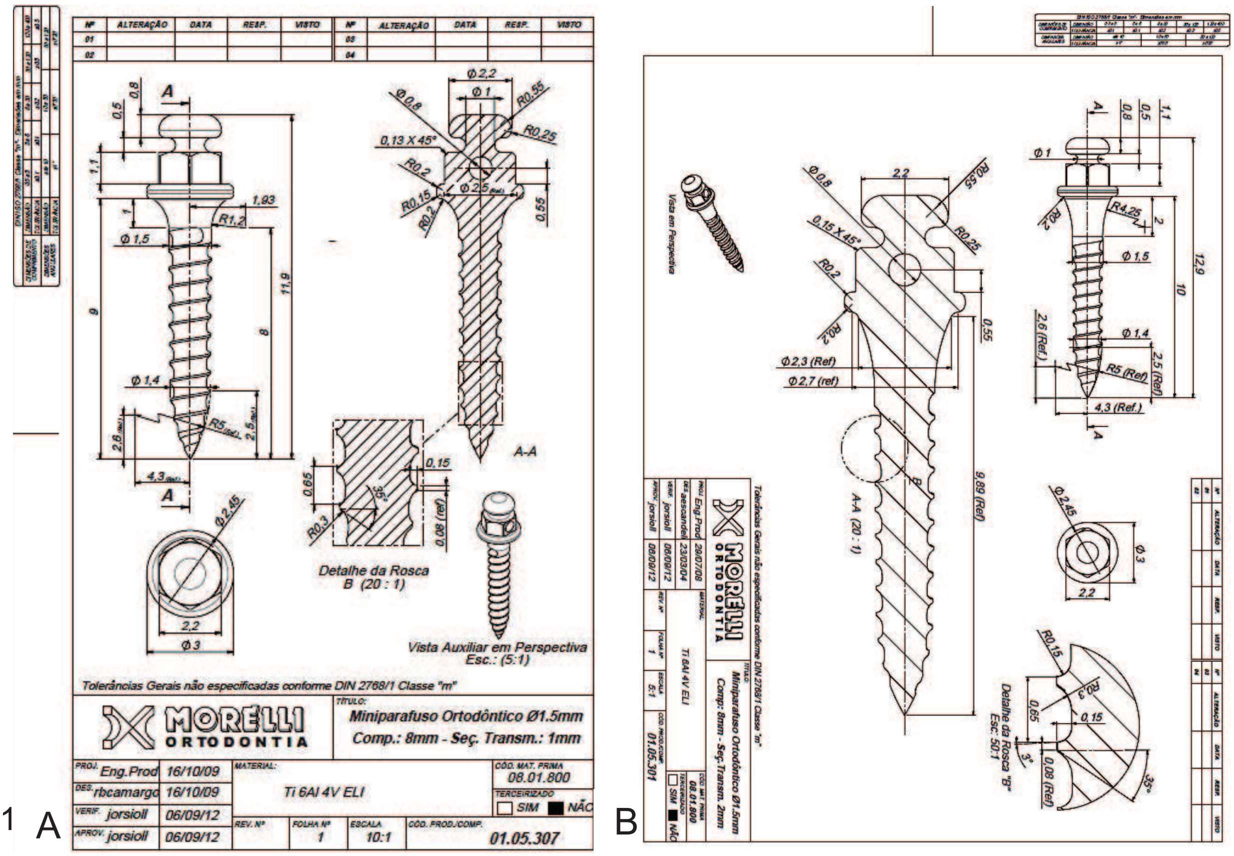

Fig 1. Mini-screw (MS) design. **A,** MS with a 1-mm transmucosal profile and **B,** MS with a 2-mm transmucosal profile.

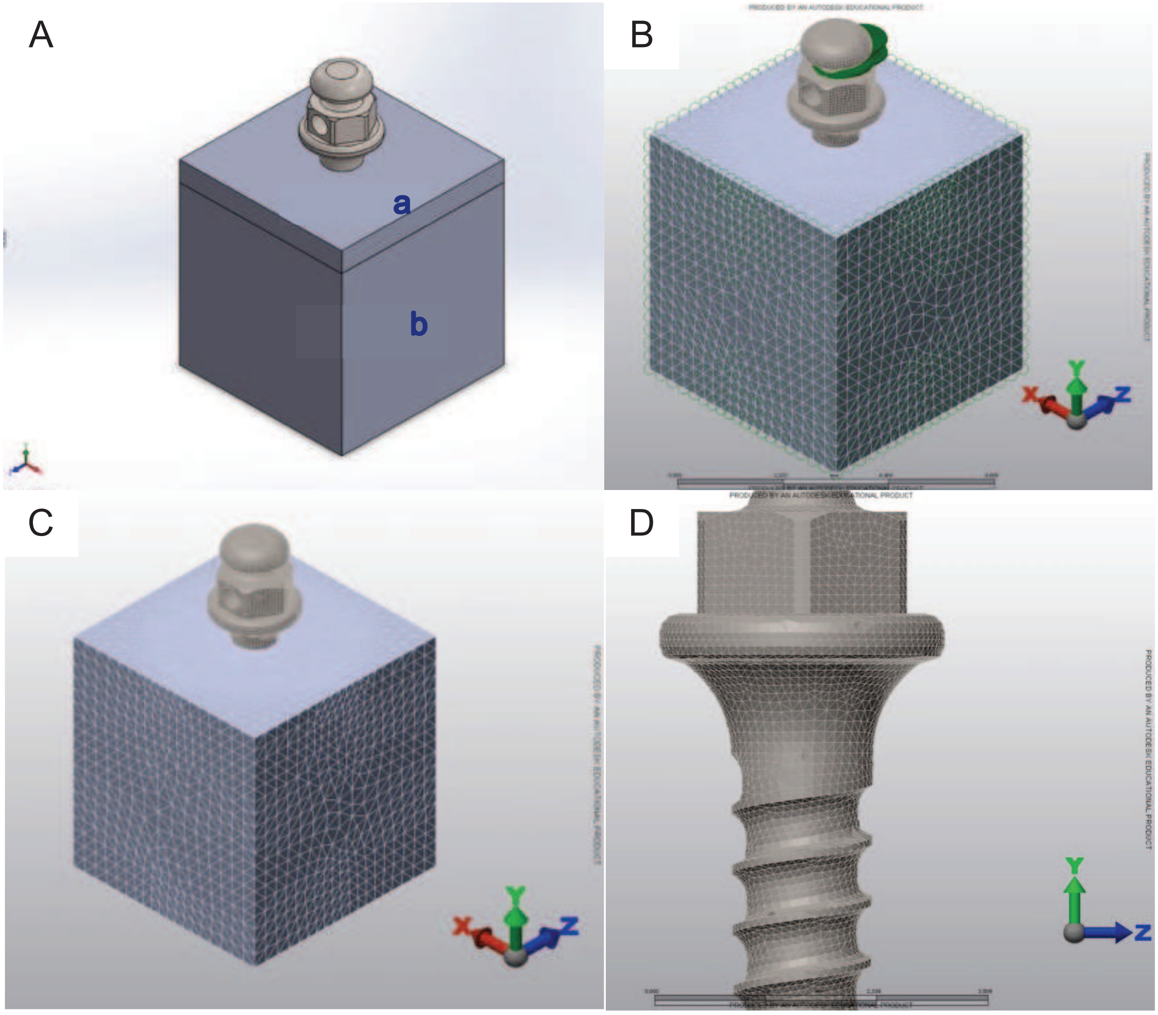

Fig 2. Three-dimensional modeling used in the mechanical tests **A,** Cortical bone measuring 1-mm thick (top line) and trabecular bone. **B,** Three-dimensional view of boundary conditions and load application view at the mini-screw (MS) head; **C,** Three-dimensional tetrahedral mesh; **D,** MS mesh detail

Table 1. Material properties used in finite element analysis

Material	*Young´s modulus (MPa)*	*Poisson´s ratio*
Titanium	110,000[a]	0.33[a]
Stainless steel	205,000[b]	0.29[b]
Cortical bone	13,800[c]	0.26[c]
Cancellous bone	345[c]	0.31[c]

[a]:Suzuki et al.[20]; [b]Kojima and Fukui[21]; [c]Jones et al.[22]

All computational MS models were inserted into a cuboid block format (9-mm height and 8-mm width and depth) and modeled three-dimensionally in the same program. The blocks were formed into two layers reflecting the relevant performance properties of cortical and trabecular bone (Figure 2A). The cortical bone layer was fixed at 1-mm thick. The interface between the bone and MS was perfect contact. Once complete, the solid model was transferred to the finite element program (Autodesk Simulation Multiphysics® 2013).

Concerning the model geometry, we generated a finite element mesh comprising linear tetrahedral elements of high order (first order) and then performed a convergence analysis of the stress field defined by the mesh in order to allow the FEM analysis. The length of the element edges in the analysis ranged 0.375–0.500 mm (Figure 2C and D). The block exhibited higher mesh refinement in the 3-mm bone region surrounding the MS (Figure 3A and B). Thus, the final mesh in the EM1 and EM2 models comprised 416,889 elements and 81,421 nodes, while the final EM3 and EM4 models comprised 577,429 elements and 109,645 nodes. On the side faces of each cuboid, the translations were restricted in the x-, y-, and z-directions as boundary conditions (Figure 2C, green circles). These conditions were necessary to simulate the rest of the jaw. A 3.5-N load was applied to the MS head directed along the z axis (Figure 2B, green lines). This force magnitude is realistic and applied clinically.[23]

All materials (cortical bone, cancellous bone, titanium MS, and stainless steel MS) were assumed to have homogeneous, isotropic and linearly elastic behavior with a specific Young's modulus and Poisson's ratio (Table 1). Furthermore, the stress field and strain analysis of the finite element occurred within the elastic range of stress.

The simulations were divided into two length groups (1 mm and 2 mm) and subjected to the following MS and transmucosal profile material combinations:

1) EM1: stainless steel MS and 1-mm transmucosal profile

2) EM2: titanium MS and 1-mm transmucosal profile

3) EM3: stainless steel MS and 2-mm transmucosal profile

4) EM4: titanium MS and 2-mm transmucosal profile

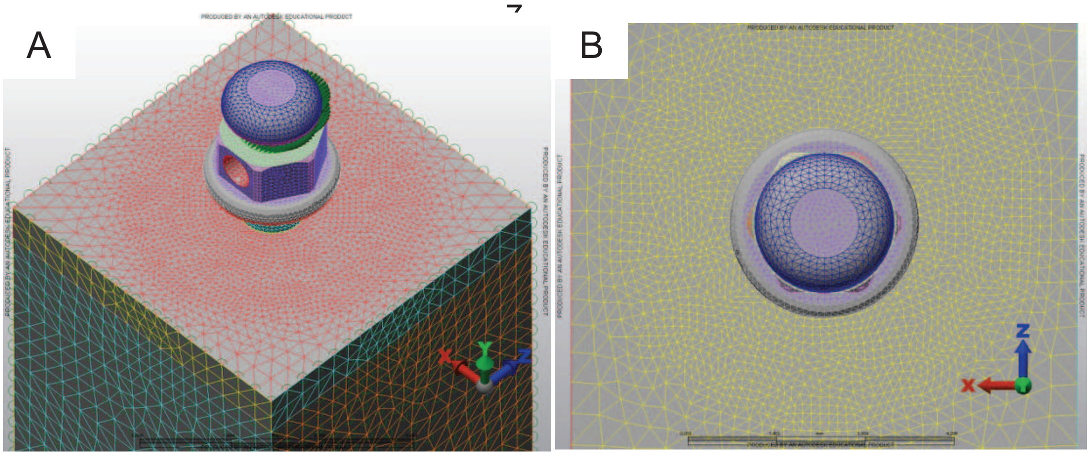

Fig 3. A, Mesh refinement in the bone region surrounding the mini-screw (MS); **B,** Magnified top view of the mesh.

The coordinates were determined automatically by the Autodesk Simulation Multiphysics® program. The field stress on the FEM was assessed according to the distortion energy theory for ductile materials, also known as the von Mises[24,25] failure criterion,[24,25] which has been used previously in orthodontics.[20,26,27]

The stress on the bone and MS elements was evaluated by the von Mises yield criterion. Maximum stress in the cortical bone was measured and compared between each model.

Results

1. Stress evaluation

The stress distribution in the bone and MS was evaluated according to a color scale, with blue indicating regions of low stress and red indicating regions of high stress. These regions also indicated areas of stress concentration.

We found that the stress distribution at the bone-MS interface in all four test conditions was highest at the MS region in contact with the cortical bone. In addition, in all loading conditions for the MS, the stress was the most concentrated outside the bone just above the cortical bone.

The EM4 model showed the highest von Mises stress value (76.94 MPa), followed by EM3 (68.38 MPa), EM2 (53.09 MPa), and EM1 (43.64 MPa).

Table 2. Maximum and minimum von Mises stress values in the four load models

	EM1 (MPa)	*EM2 (MPa)*	*EM3 (MPa)*	*EM4 (MPa)*
MAX	43.64	53.09	68.38	76.94
MIN	0	0	0	0

Subtitles: MAX (maximum voltage value); MIN (minimum voltage value)

1) Stainless steel MS with 1-mm transmucosal profile (EM1)

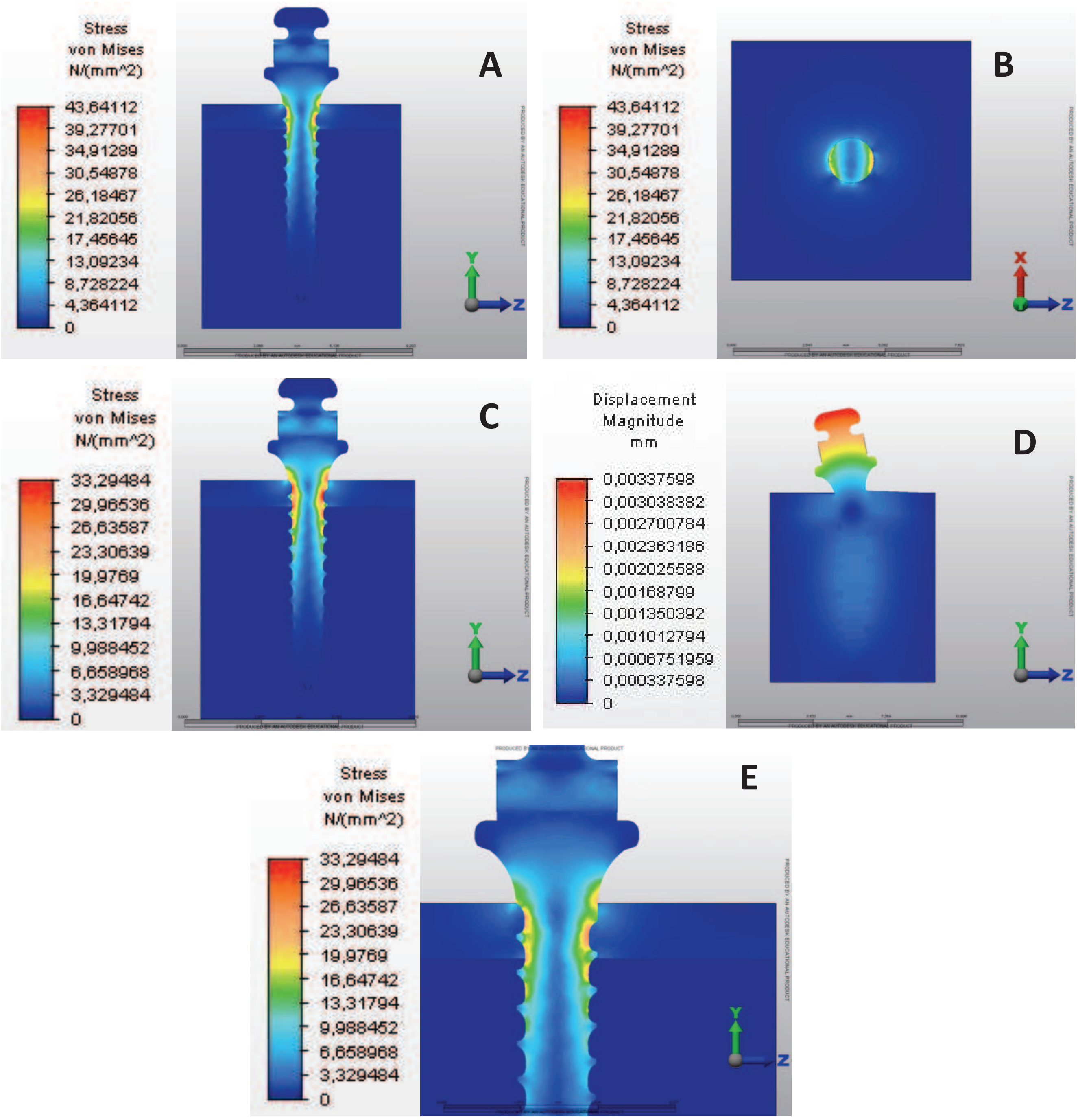

Fig 4. Stress distribution in EM1. **A,** Cross-section; **B,** Perpendicular court; **C,** Transverse section; the EM1 model had the lowest magnitude of stress among the mechanical test models; **D,** Transverse section deflection (scale: magnitude of deflection); **E,** Stress distribution in the region corresponding to the cortical bone

2) Titanium MS with 1-mm transmucosal profile (EM2)

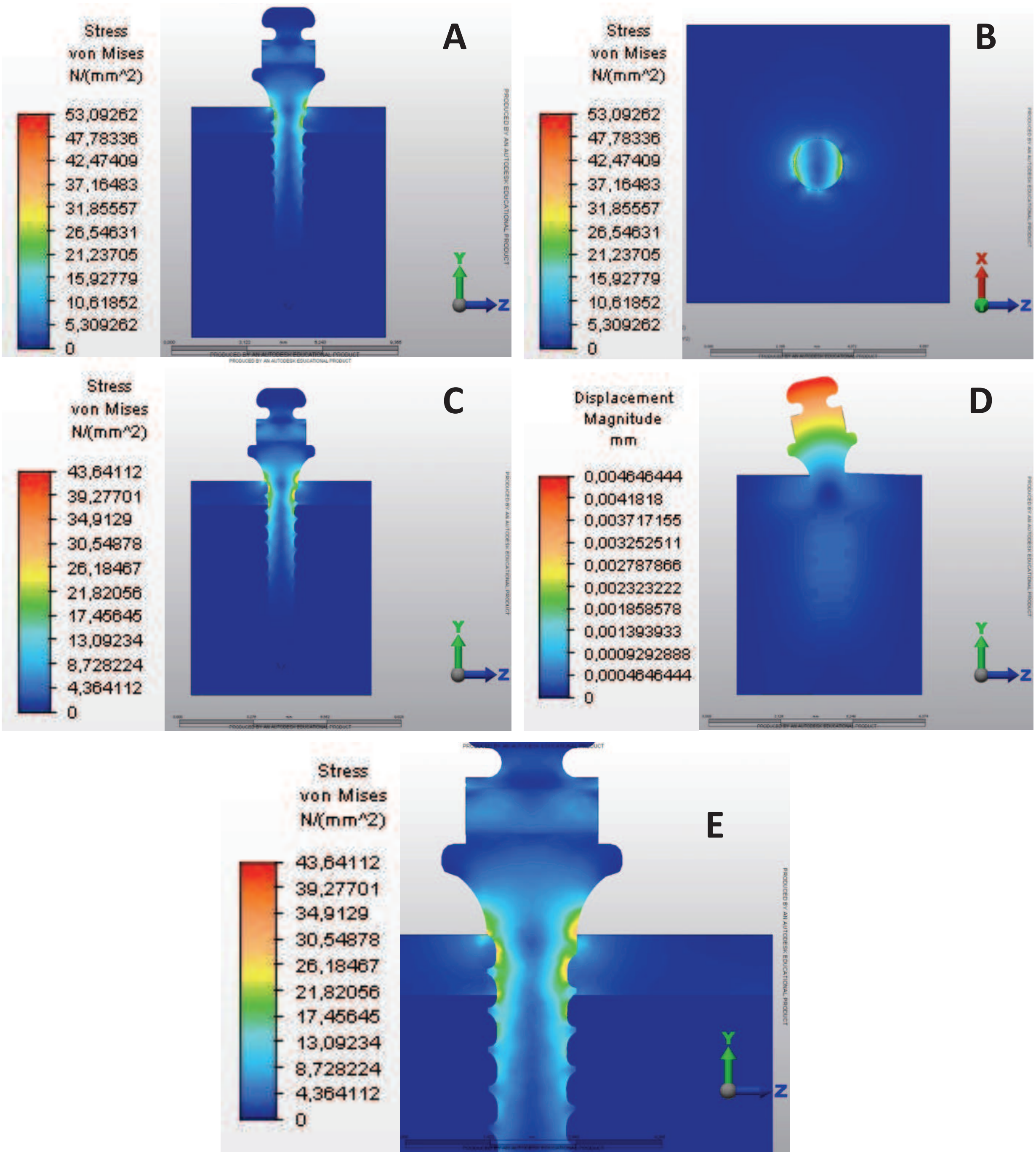

Fig 5. Stress distribution in EM2. **A,** Cross-section; **B,** Perpendicular court; **C,** Transverse section; the EM2 model had the lowest magnitude of stress among the mechanical test models; **D,** Transverse section deflection (scale: magnitude of deflection); **E,** Stress distribution in the region corresponding to the cortical bone

3) Stainless steel MS with a 2-mm transmucosal profile (EM3)

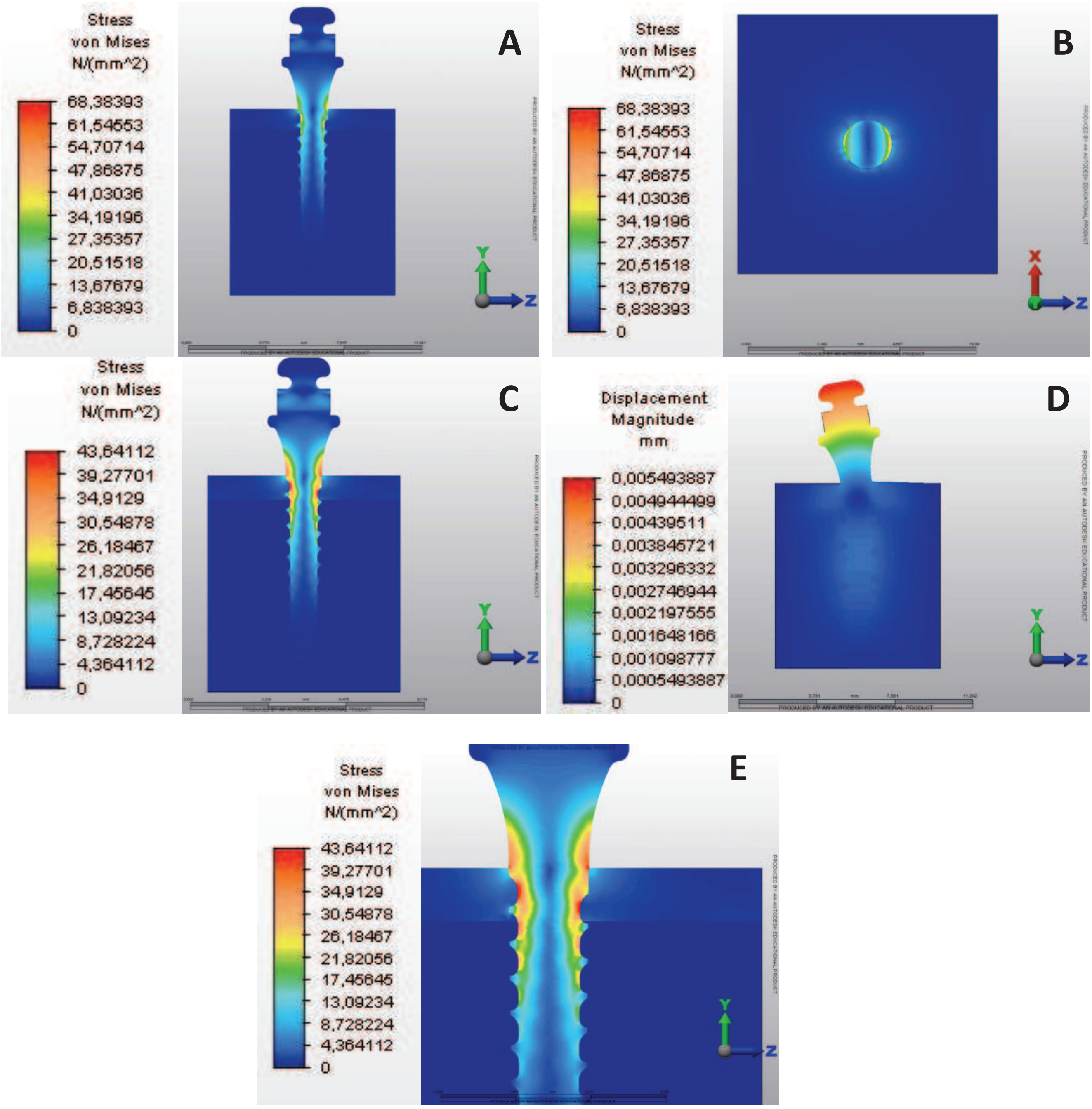

Fig 6. Stress distribution in EM3. **A,** Cross-section; **B,** Perpendicular court; **C,** Transverse section; the EM3 model had the lowest magnitude of stress among the mechanical test models; **D,** Transverse section deflection (scale: magnitude of deflection); **E,** Stress distribution in the region corresponding to the cortical bone

4) Titanium MS with a 2-mm transmucosal profile (EM4)

1

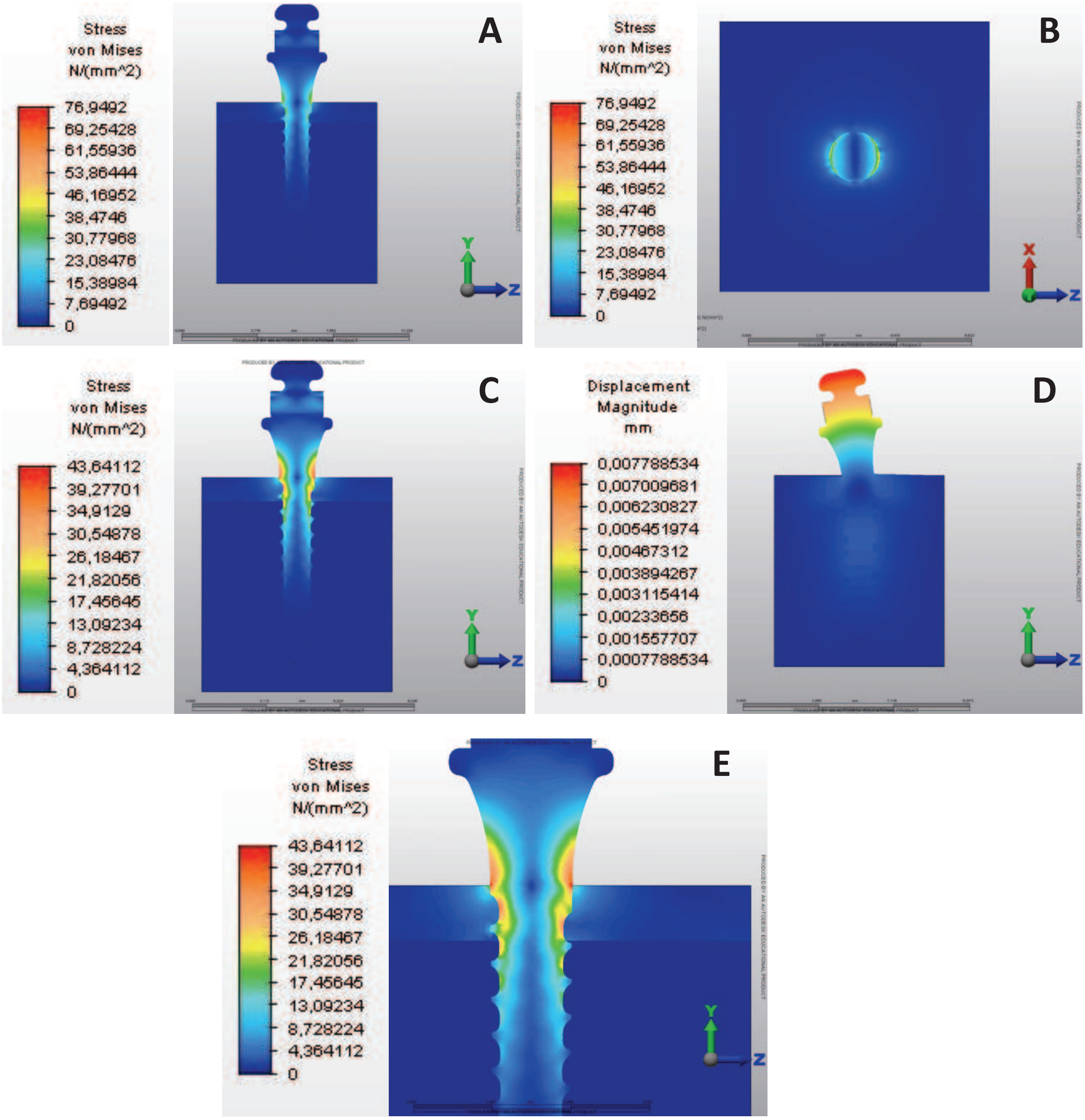

Fig 7. Stress distribution in EM4. **A,** Cross-section; **B,** Perpendicular court; **C,** Transverse section; the EM4 model had the lowest magnitude of stress among the mechanical test models; **D,** Transverse section deflection (scale: magnitude of deflection); **E,** Stress distribution in the region corresponding to the cortical bone

2. *Material composition (Ti-6Al-4V and 316 stainless steel)*

The von Mises stress was higher in the titanium MS (EM2 53.09 MPa; EM4 76.94 MPa) than in the stainless steel MS (EM1 43.64 MPa; EM3 68.38 MPa).

3. Transmucosal profile

The von Mises stress was higher in the MS with a 2-mm transmucosal profile (EM3 68.38 MPa; EM4 76.94 MPa) than in those with a 1-mm transmucosal profile (EM1 43.64 MPa; EM2 53.09 MPa).

4. Displacement

The largest displacement occurred in EM4 (7.78 μm), followed by EM3 (5.49 μm), which both have a 2-mm transmucosal profile, followed by EM2 (4.64 μm), and EM1 (3.37 μm), which both have a 1-mm transmucosal profile.

Discussion

Finite element models

FEM is a numerical method that helps solve complex problems in a variety of scientific fields by dividing complex areas into small, simple areas. This method represent the deformation as well as the distribution of stress in a tridimensional way, of bodies that are exposed to loads, making it useful for simulating the stress distribution in biological and medical science studies.[20,28] The results of FEM studies provide important information allowing investigators to accurately understand complex biological reactions that are otherwise difficult to study in humans due to the wide variations between subjects.

The effectiveness of FEM depends on the type and degree of mesh refinement. In this study, the end of the mesh model EM1 and EM2 comprised

416.889 elements and 81.421 nodes, while the end of the EM3 and EM4 models comprised 577.429 elements and 109.645 nodes. This study did have several limitations in the simulation. The geometry used to simulate a bone block was a cuboid pack for simplicity, the material properties were homogeneous and isotropic, and the soft tissue was not been simulated because its effect was surmised to be insignificant. Also, the MS was positioned within the bone block perpendicular to the bone surface and in perfect condition, and perfect contact was maintained between the bone-MS interface. However, these simplifications should not change the analysis of the field of quantitative strains of numerical simulations.[29]

We observed that most of the von Mises stress concentrated in the MS region, which is in contact with the cortical bone. Also, during loading, the concentrated stress region was located outside the bone just above the cortical bone in all models. These results are consistent with those of Suzuki et al.,[20] who found that the maximum voltage of the MS was concentrated between the surface of the support bone and MS.

MS transmucosal profile associated with stress

The transmucosal profile is designed to maintain the health of peri-implant tissues, especially in areas with minimal attached gingiva, because inflammation is a contributing factor to MS failure.[30] Our results showed that the maximum stress was significantly higher in MS with a 1-mm transmucosal profile than in those with a 2-mm transmucosal profile. There are no studies comparing MS according to the transmucosal profile thickness; however, our results reflect the principle of a bending moment caused by a force. The moment of force is the effect produced on a body by a force applied at a relatively distant point on the line of action of this force, ultimately generating rotational movement.[31,32]

Following this principle, the further away the applied force, the greater the moment, and consequently, the greater the stress. Thus, the MS with a 2-mm transmucosal profile has a longer lever arm than a MS with a 1-mm transmucosal profile. Application of the equivalent force generated a higher moment in the MS with a 2-mm transmucosal profile and consequently, greater bending stress.

In a FEM study assessing different MS lengths (8, 10, and 12 mm), Lin et al.[45] concluded that a longer MS length was associated with higher stress in the cortical bone. Nova et al.[13] compared two brands of MS with and without a transmucosal profile and concluded that the presence or absence of the transmucosal profile did not affect the insertion or removal torques because only one device (NEODENT) showed significant torque, while torque in the other device (SIN) did not vary. Some studies evaluating the effect of MS length have reported inconsistent or inconclusive results.[35,36] This is because the MS length is not a determining factor in stress distribution. The exposed MS length from the bone and the length of the moment lever arm are the factors that primarily influence the stress distribution and displacement. Therefore, both the MS length and depth of implantation must be considered. [29]

Other factors, such as the MS diameter, inflammation, infection, insertion into non-keratinized tissue, and a small MS size, may also affect the MS stability.[37] Miyawaki et al.[37] and Lim et al.[34] found that the MS diameter affected the success rate, but Park et al.[7] reported that the diameter had no effect. In 2003, Miyawaki et al. examined the success rate of three types of titanium MS and the factors associated with stability; they concluded that the 1-mm MS was more successful.

MS displacement

The upper region of the MS was the point of maximum deformation, as illustrated in Figures 4–7D. The greatest deformation occurred in the mechanical tests using a MS with a 2-mm transmucosal profile compared with the 1-mm transmucosal profile. In this study, the MS displacement was minimal.

MS materials associated with stress

The MS used in this study comprised two different materials, stainless steel (ASTM F138) and titanium (ASTM F136). The stainless steel used in dental implants must have suitable mechanical and physical properties, namely low-permeability, high strength, low percentage of impurities, corrosion resistance, and amenable to a variety of sterilization techniques.[39,40] Titanium (Ti-6Al-4V) is known to have two phases: an hexagonal close-packed alpha phase and a beta crystalline structure phase, which is a body centered cubic. Both phases are present at ambient temperature, ensuring corrosion resistance, high specific strength, good properties at high temperatures, formability, and usability.[40,41]

The present results showed that the maximum von Mises stress occurred in the MS comprising titanium alloy Ti-6Al-4V compared with the stainless steel 316 device. We surmise that the stress field was higher in the titanium alloy MS because it is approximately 47% less stiff than the stainless steel 316 MS, and the bending stiffness in response to a load is directly related to the Young's modulus.[33,43]

The experimental approach in humans has biological, ethical, and privacy limitations, and FEM provides a more practical approach to evaluating dental biomechanics.[29,44] Our simulation model appears to exhibit the same pattern of stress distribution in the surrounding bone and MS, and thus, may be a clinically useful tool for assessing the effect of stress distribution and predicting the tissue reaction in response to an orthodontic force. As observed by Lin et al.[45], excessive bone stress induced by a longer MS exposure length facilitates local bone resorption.

Conclusion

In the present study, we analyzed the stress generated in the MS and surrounding bone upon applying a perpendicular force to the MS according to the transmucosal profile length (1 mm and 2 mm) and MS composition (Ti-6Al-4V alloy steel and stainless 316) using the finite element method. We concluded the following:

1. The highest von Mises stress was observed in EM4, followed by EM3, EM2, and EM1.
2. The stress was more concentrated at the MS than in the bone.
3. The stress was less in the stainless steel MS than in the titanium MS;
4. MS with a 2-mm transmucosal profile showed higher stress than those with a 1-mm transmucosal profile.

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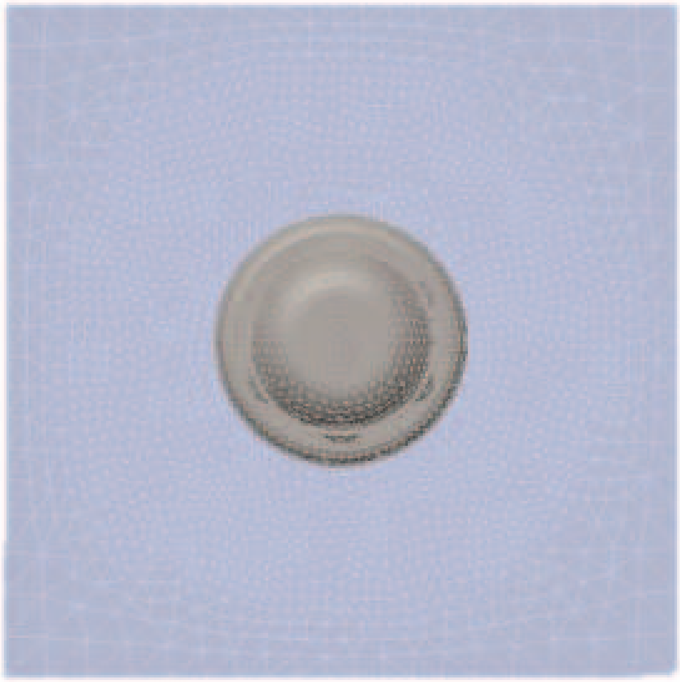
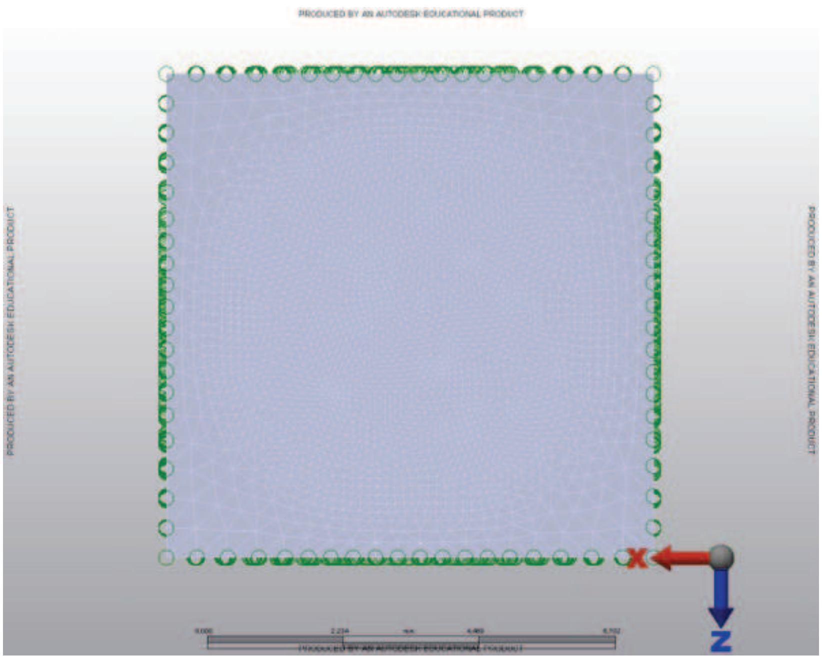
ANEXO A– Figuras adicionais

A.1- Malha tridimensional vista superior



A.2- Malha tridimensional vista inferior, condições de contorno nas paredes externas no bloco



ANEXO B- Autorização para utilização dos desenhos do MI da Empresa SAC Morelli

MORELLI ORTODONTIA

5 de Novembro de 2013

0800-121455
0800-7031455

SAC Morelli Finalizado

Prezada

Estephany Gordillo Yépez

Agradecemos seu contato bem como suas considerações sobre nossos produtos e serviços.

Segue abaixo os esclarecimentos sobre as SACs registradas na Morelli.

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Tipo	Serviço
Serviço	Informações Técnicas
Descrição fornecida pelo Cliente	
Ocorrência	Meu nome é Estephany Gordillo, estou cursando o mestrado de ortodontia na PUCPR, meu orientador é o professor Doutor Orlando Tanaka. Estou trabalhando junto com Renata Marangon e nossa disertação vai ser sobre mini-implantes e elementos finitos. E por isso que queríamos saber se vocês podem mandar para nos as dimensões dos MI de 8 mm de comprimento e com perfil transmucoso de 1 e 2 mm. O email anterior não chegou é por isso que estou entrando em contato de novo com vocês. Muito obrigado pela sua atenção e tempo investidos. Esperamos sua resposta nos siguientes emails: joanagordillo3@gmail.com; flacagordillo@hotmail.com; remarangon@hotmail.com. Atenciosamente Estephany Gordillo Cirurgiã Dentista
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These guidelines are supplemental to the Guidelines for Original Articles, which describe how to meet general submission requirements, such as figure formats, reference style, required releases, and blinding.

Systematic Review and Meta-Analysis Guide for Authors

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