




# Biomechanical analysis of occlusal modes on the periodontal ligament while orthodontic force applied

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## Abstract

**Objective** The study objective was to investigate four common occlusal modes by using the finite element (FE) method and to conduct a biomechanical analysis of the periodontal ligament (PDL) and surrounding bone when orthodontic force is applied.

**Materials and methods** A complete mandibular FE model including teeth and the PDL was established on the basis of cone-beam computed tomography images of an artificial mandible. In the FE model, the left and right mandibular first premolars were not modeled because both canines required distal movement. In addition, four occlusal modes were simulated: incisal clench (INC), intercuspal position (ICP), right unilateral molar clench (RMOL), and right group function (RGF). The effects of these four occlusal modes on the von Mises stress and strain of the canine PDLs and bone were analyzed.

**Results** Occlusal mode strongly influenced the distribution and value of von Mises strain in the canine PDLs. The maximum von Mises strain values on the canine PDLs were 0.396, 1.811, 0.398, and 1.121 for INC, ICP, RMOL, and RGF, respectively. The four occlusal modes had smaller effects on strain distribution in the cortical bone, cancellous bone, and miniscrews.

**Conclusion** Occlusal mode strongly influenced von Mises strain on the canine PDLs when orthodontic force was applied.

**Clinical relevance** When an FE model is used to analyze the biomechanical behavior of orthodontic treatments, the effect of muscle forces caused by occlusion must be considered.

**Keywords** Finite element method · Orthodontics · Occlusal mode · Periodontal ligament

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## Introduction

Orthodontic treatment can align patients' teeth and change their occlusions in addition to improving their facial appearances. Several mechanisms can cause tooth movement, and the most common theory of tooth movement is pressure–tension theory [1–4]. Through the force exerted by dental brackets, teeth can compress or stretch the periodontal ligaments (PDLs) surrounding their roots. The compression side of the PDL contracts because of pressure on its blood vessels, which reduces bloodstream volume and thus causes the resorption of surrounding bone. By contrast, the tension side of the PDL causes bone deposition. In other words, the goal of tooth movement is achieved by bone remodeling [1].

Researchers typically use animal model experiments to perform histomorphometry to study and verify the pressure–tension theory [5–7]. The rapid development of computer hardware and software in the last three decades has enabled more researchers to use the finite element (FE) method to study orthodontic treatment. During this time, the FE method has been commonly used to investigate the biomechanical responses of roots, the PDLs, and surrounding bone during

orthodontic treatments. Bouton et al. [8], Sardarian et al. [9], and Sugii et al. [10] have created FE models of an individual tooth and simulated the PDL as a linear elastic material to study the distribution of stress and strain on the PDL and surrounding bones after exerting orthodontic force. Field et al. [11], Caballero et al. [12], and Gerami et al. [13] have also explored similar topics using partial mandible models with multiple teeth created by the FE method.

Numerous studies have simulated the PDL as a linear elastic material. However, researchers who have conducted experiments on rat [14], miniature pig [15], or human cadavers have suggested that the PDL more closely resembles a bilinear elastic material. Therefore, to more closely approximate the human PDL, several studies using FE simulations of orthodontic treatment have assumed that the PDL has bilinear elasticity [12, 14, 16, 17]. Hartmann et al. [17] examined the effect of a tooth's size on its initial mobility. Papageorgiou et al. [15, 18] examined the effect of orthodontic appliance material on the force exerted on teeth. Cattaneo et al. [16] and Melsen et al. [19] compared the effects of modeling the PDL as a linear or nonlinear elastic material on tooth movement.

Most FE studies [8–13, 16, 20] have adopted only partial mandibular models to study biomechanical responses to simulated orthodontic tooth movement. Consequently, regardless of whether the PDL is simulated as a linear or nonlinear elastic material, these studies have used complete constraints on the mesial and distal sides of their partial mandibular models as the boundary conditions in FE simulations, and they have simply simulated orthodontic force on the teeth as a loading condition. In other words, these studies have not considered the influence of occlusal muscular forces in daily life on the loading of the PDL. Therefore, the objective of the present study was to investigate four common occlusal modes by using the FE method and to conduct a biomechanical analysis of the PDL and surrounding bone when orthodontic force is applied. In this study, we established a complete mandibular model and simulated the PDL as a bilinear elastic material. We referred to the anatomy of the human mandible to divide it into muscle attachment areas to simulate the muscle forces of four occlusal modes and conduct a biomechanical analysis of the PDL and surrounding bone when orthodontic force is applied.

## Materials and methods

### Mandibular bone, teeth, PDL, orthodontic bracket, and miniscrew models

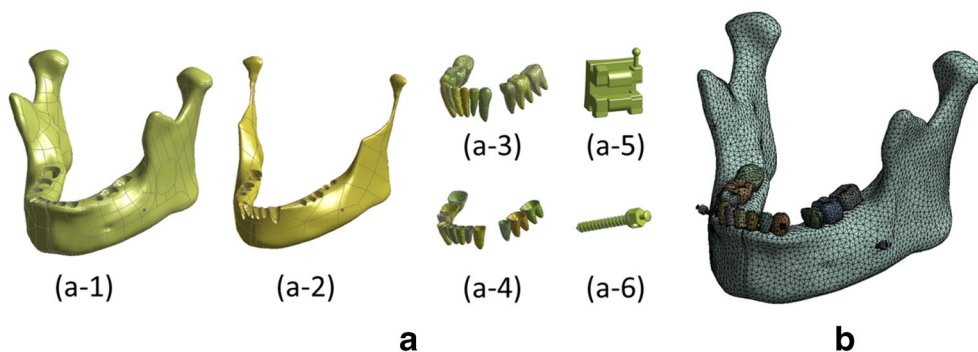
Three-dimensional solid models of the artificial mandible, which included cortical bone, cancellous bone, teeth, and the PDL, were established on the basis of dental cone-beam computed tomography images by

using Mimics 15.0 (Materialise, Leuven, Belgium) and Geomagic Design X (3D Systems, Inc., Rock Hill, SC, USA; Fig. 1). In the solid model, the left and right mandibular first premolars were not modeled because both canines required distal movement. The three-dimensional solid models of orthodontic brackets and miniscrews were created using SolidWorks (Swanson Analysis, Canonsburg, PA, USA; Fig. 1). In addition, to simplify the computer model, only two orthodontic brackets were created, and they were placed on the buccal side of the crown of both mandibular canines. Furthermore, two orthodontic miniscrews were placed in the inferior region between the first and second molars. The solid models were imported to the ANSYS Workbench FE package (Swanson Analysis, Canonsburg, PA, USA) for mesh and solution modeling. The interfaces between all components (cortical bone–cancellous bone, cancellous bone–PDL, PDL–tooth, tooth–bracket, and miniscrew–bone) were assumed to have full bonding. The material properties employed in the FE components are listed in Table 1. Except for the PDL, which was modeled as a bilinear elastic material, all other components were assumed to have linear elasticity. On the basis of our experience, we simulated five different element sizes for the mesh in the models' convergence tests. The results of the convergence analysis suggested that the group using element sizes of 2.3 mm for cortical bone, 2.3 mm for cancellous bone, 1.7 mm for the miniscrews, 1.7 mm for the brackets, and 1.4 mm for the PDL achieved convergence. Elements representing all components were created using 10-node tetrahedral elements. This FE model had a total of 235,603 nodes and 151,296 elements. Analyses were performed using the nonlinear static solver because of the nonlinear material properties of the PDL.

### Four occlusal modes

For both sides of the mandible, orthodontic forces of 2 N (approximately 200 g) were applied in opposite directions between the miniscrew and the top of the bracket. To study the effects of different occlusal modes on the biomechanical analysis, six muscle attachment areas on each mandibular side were defined. The loading conditions comprised six principal muscles: the superficial masseter, deep masseter, medial pterygoid, anterior temporalis, middle temporalis, and posterior temporalis. Four static clenching tasks were simulated: incisal clench (INC), intercuspal position (ICP), right unilateral molar clench (RMOL), and right group function (RGF; Fig. 2). The loading forces and constraints are listed in Table 2. Furthermore, to evaluate the effect of the muscles, another FE model including orthodontic force without muscle force was developed.

**Fig. 1** **a** Solid model, a1–a6: mandibular cortical bone, mandibular cancellous bone, teeth, periodontal ligament, bracket, and miniscrews. **b** Finite element mesh model



**Evaluation parameters**

We conducted a biomechanical analysis of the effects of four occlusal modes on the PDL, miniscrew, and bones. Studies have suggested that orthodontic treatment triggers tooth movement because of strain on the PDL [16, 18, 27, 28]. Therefore, von Mises strain was selected as an evaluating factor for the PDL. Additionally, von Mises stress has been used as an evaluating factor for the cortical bone, cancellous bone, and miniscrews [29–32].

**Results**

**FE model validation**

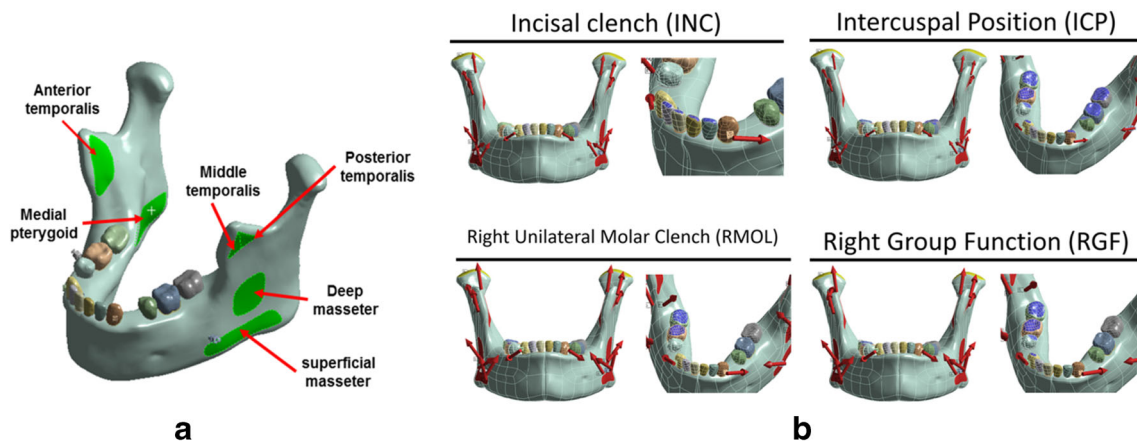
To verify the mandibular model developed in this study, we consulted research on the occlusal force by Kikuchi et al. [33] for comparison. Kikuchi et al. [33] used electronic devices to measure the occlusal force of the premolar and molar teeth in the human RGF occlusal mode. We modified the proposed FE model to be without missing teeth and simulated the same occlusal mode to examine the reaction forces on the fixed surfaces for the premolar and molar teeth. The reaction forces of the second molar, first molar, premolar, and canine teeth differed from actual humans’ occlusal forces as measured by Kikuchi et al. [33]; nevertheless, the ranking of the value of the four types of teeth is consistent (Table 3).

**von Mises strain on the PDL from different occlusal modes**

The FE model with only orthodontic force (i.e., the model excluding forces exerted by mandibular muscles) yielded von Mises strain values of 0.394 on the PDLs of the left and right canine teeth. In the INC and ICP occlusal modes, because of the symmetric occlusal force exerted by the mandibular muscles, the von Mises strains on the PDLs of the left and right canine teeth were equal. However, the von Mises strains on the PDLs of the canine teeth in the ICP occlusal mode were 4.57 times higher than those in the INC occlusal mode (Fig. 3). By contrast, in the RGF occlusal mode, the mandible is affected by asymmetric muscle force. Therefore, the von Mises strain on the PDL of the right canine (1.12) was clearly higher than that in the left canine (0.39; Fig. 3). The main reason for this difference is that in the RGF occlusal mode, the right canine tooth, premolars, and molars are completely constrained. Therefore, when the crown of the right canine tooth is immovable, the von Mises strain on the PDL of the tooth is higher than that on the PDL of the unfixed left canine tooth. The RMOL occlusal mode was also affected by left–right asymmetrical occlusal force in the simulation. However, because the RMOL occlusal mode was constrained at the crowns of the right molars, it did not yield a considerable difference between the PDLs of the left and right canine teeth (Fig. 3).

**Table 1** Material properties used in the FE models

Component	Elastic modulus (MPa)	Poisson’s ratio	Reference
Bracket	230,000	0.3	Wu et al. 2011 [21]
Miniscrew			
Cortical bone	14,900	0.3	Poppe et al. 2002 [22]
Cancellous bone	460	0.3	
Tooth	18,600	0.3	Leung et al. 2008 [23]
Periodontal ligament	Bilinear: 0.15/0.50 Ultimate strain E <sub>12</sub> : 6.3%	0.3	Kawarizadeh et al. 2003[24]



**Fig. 2** a Muscle attachment areas in the mandibular bone. b Four occlusal modes

**Table 2** Muscular forces and constraints of the six clenching tasks. All raw data were obtained from the study of Koriath and Hannam [25]. This table was modified from that in our previous study [26]

Clenching tasks	Side	Direction	Muscular force (N)						Constraint*
			SM	DM	MP	AT	MT	PT	
INC	Right	<b>Force</b>	<b>76.2</b>	<b>21.2</b>	<b>136.3</b>	<b>12.6</b>	<b>5.7</b>	<b>3.0</b>	Constrained the incisor regions
		F <sub>X</sub>	-15.8	-11.6	66.3	-1.9	-1.3	-0.6	
		F <sub>Y</sub>	-31.9	7.6	-50.9	-0.6	2.9	2.6	
	F <sub>Z</sub>	67.3	16.1	107.8	12.5	4.8	1.4		
	Left	<b>Force</b>	<b>76.2</b>	<b>21.2</b>	<b>136.3</b>	<b>12.6</b>	<b>5.7</b>	<b>3.0</b>	
		F <sub>X</sub>	15.8	11.6	-66.3	1.9	1.3	0.6	
F <sub>Y</sub>		-31.9	7.6	-50.9	-0.6	2.9	2.6		
ICP	Right	<b>Force</b>	<b>190.4</b>	<b>81.6</b>	<b>132.8</b>	<b>154.8</b>	<b>91.8</b>	<b>71.1</b>	Constrained the canine and premolar regions
		F <sub>X</sub>	-39.4	-44.6	64.6	-23.1	-20.4	-14.8	
		F <sub>Y</sub>	-79.8	29.2	-49.6	-6.8	45.9	60.8	
	F <sub>Z</sub>	168.3	61.9	105.1	153.0	76.8	33.7		
	Left	<b>Force</b>	<b>190.4</b>	<b>81.6</b>	<b>132.8</b>	<b>154.8</b>	<b>91.8</b>	<b>71.1</b>	
		F <sub>X</sub>	39.4	44.6	-64.6	23.1	20.4	14.8	
F <sub>Y</sub>		-79.8	29.2	-49.6	-6.8	45.9	60.8		
RMOL	Right	<b>Force</b>	<b>137.1</b>	<b>58.8</b>	<b>146.8</b>	<b>115.3</b>	<b>63.1</b>	<b>44.6</b>	Constrained the right molars
		F <sub>X</sub>	-28.4	-32.1	71.4	-17.2	-14.0	-9.3	
		F <sub>Y</sub>	-57.4	21.0	-54.8	-5.1	31.5	38.1	
	F <sub>Z</sub>	121.2	44.5	116.1	114.0	52.8	21.1		
	Left	<b>Force</b>	<b>114.2</b>	<b>49.0</b>	<b>104.9</b>	<b>91.6</b>	<b>64.1</b>	<b>29.5</b>	
		F <sub>X</sub>	23.6	26.7	-51.0	13.7	14.2	6.1	
F <sub>Y</sub>		-47.9	17.5	-39.1	-4.0	32.0	25.2		
RGF	Right	<b>Force</b>	<b>34.3</b>	<b>29.4</b>	<b>12.2</b>	<b>104.3</b>	<b>61.2</b>	<b>46.9</b>	Constrained the right canine, premolars, and molars
		F <sub>X</sub>	-7.1	-16.0	6.0	-15.5	-13.6	9.8	
		F <sub>Y</sub>	-14.4	10.5	-4.6	-4.6	30.6	40.1	
	F <sub>Z</sub>	30.3	22.3	9.7	103.0	51.2	22.2		
	Left	<b>Force</b>	<b>51.4</b>	<b>21.2</b>	<b>132.8</b>	<b>11.1</b>	<b>5.7</b>	<b>4.5</b>	
		F <sub>X</sub>	10.6	11.6	-64.6	1.7	1.3	0.9	
F <sub>Y</sub>		-21.5	7.6	-49.6	-0.5	2.9	3.9		
F <sub>Z</sub>	45.4	16.1	105.1	10.9	4.8	2.2			

Muscular forces: SM, superficial masseter; DM, deep masseter; MP, medial pterygoid; AT, anterior temporalis; MT, middle temporalis; PT, posterior temporalis. Clenching tasks: INC, incisal clench; ICP, intercuspal position; RMOL, right unilateral molar clench; RGF, right group function

\*The models were constrained in all directions at the nodes on the top of the condyle in all clenching tasks

**Table 3** Occlusal force comparison between previous human experimental data and our simulation results

Tooth	Kikuchi et al. 1997 [33]						Present study	
	Sub #1	Sub #2	Sub #3	Sub #4	Mean ± SD	Ranking	Value	Ranking
Canine	81.2	37.5	25.5	55.4	49.9 ± 21.0	4	27.2	4
Premolar	41.0	85.3	62.4	108.7	74.4 ± 25.3	1	71.8	1
First molar	86.7	76.7	21.2	93.6	69.6 ± 28.6	3	37.1	3
Second molar	111.5	71.4	57.2	53.0	73.3 ± 23.1	2	42.6	2

Sub: Subject; Unit: N

**von Mises stress on mandibular bone from different occlusal modes**

For cortical bone, the maximum von Mises stress for all four occlusal modes was concentrated in the mandibular neck (Fig. 4). For the INC and ICP occlusal modes, in which occlusal forces are applied in the anterior teeth region, relatively high von Mises stress occurred in the cortical bone of the region (Fig. 4). Additionally, for the RMOL and RGF occlusal modes, in which occlusal forces are concentrated in the right posterior teeth region, relatively high von Mises stress occurred in the cortical bone of the region (Fig. 4). In cancellous bone, although the von Mises stress distribution differed according to the occlusal mode, these differences in the cancellous bone surrounding the canine PDLs were negligible (Fig. 5).

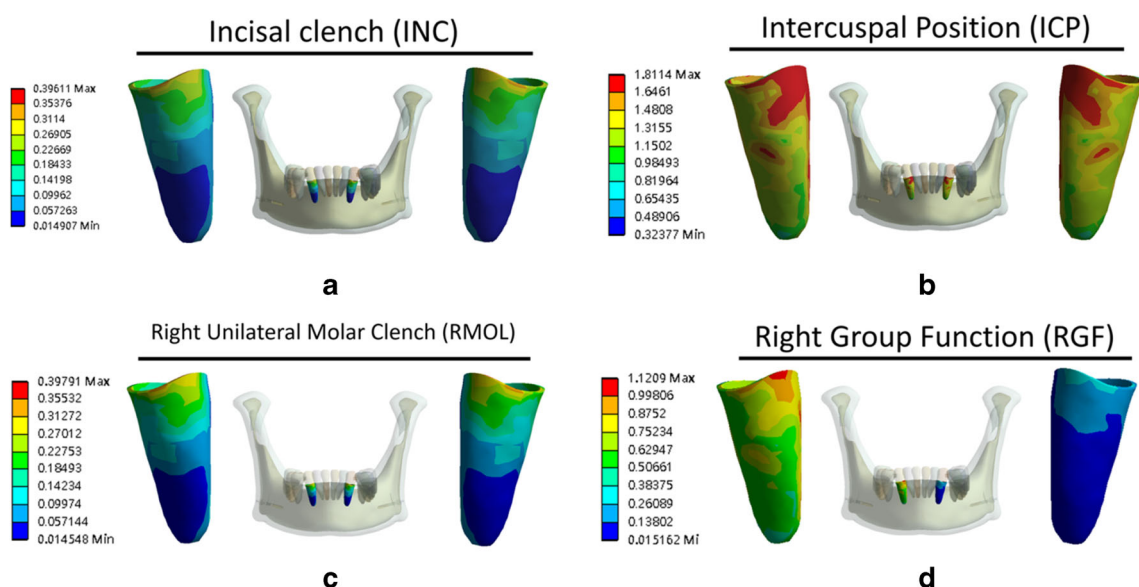
**von Mises stress on the miniscrew from different occlusal modes**

No considerable differences were observed regarding the effect of the four occlusal modes on von Mises stress values of

the miniscrews (Fig. 6). Specifically, the maximum von Mises stress on the miniscrews in all modes was less than 30 MPa, which is similar to that in models using only orthodontic force. This indicated that occlusal modes do not considerably affect miniscrews.

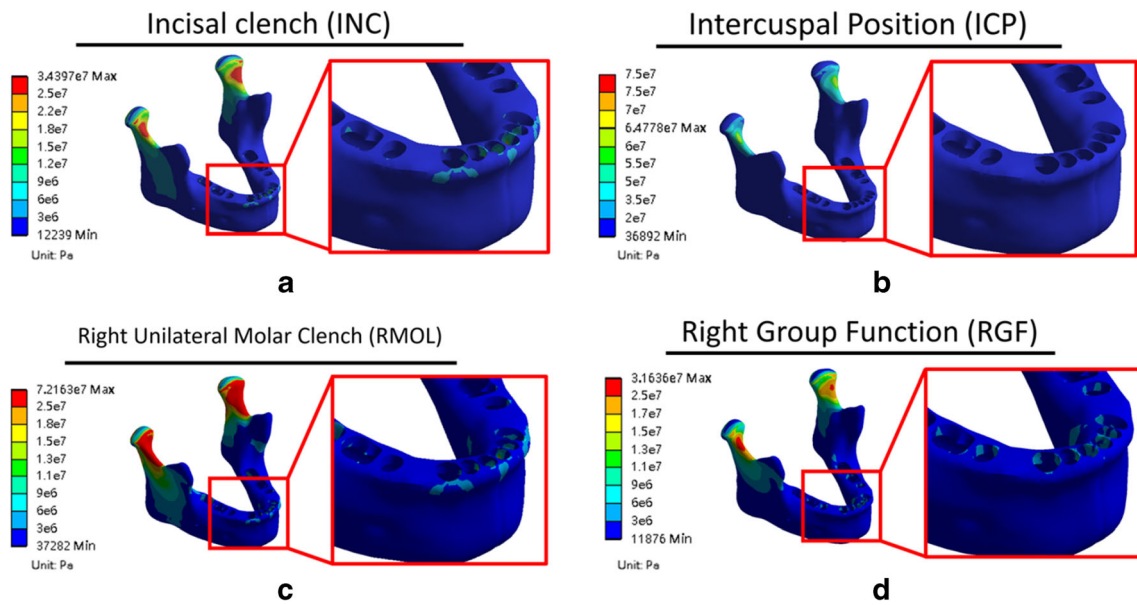
**Discussion**

Most FE model simulations of orthodontic treatments have used partial mandibular models to simulate biomechanical effects generated by exerting orthodontic force on teeth. However, in real-life situations, patients’ daily chewing behaviors during orthodontic treatment also generate occlusal forces that can generate effects. Therefore, we developed a complete mandibular model to simulate the occlusal force generated by chewing. Additionally, we set the PDL as a bilinear elastic material to more realistically simulate the effect of occlusal forces of daily life on orthodontic treatments. The simulation results revealed that although the same orthodontic force was exerted, the PDL surrounding the targeted canine tooth was affected differently under different occlusal modes.



**Fig. 3** von Mises strain on the PDL of both mandibular canines in different occlusal modes: **a** INC, **b** ICP, **c** RMOL, and **d** RGF





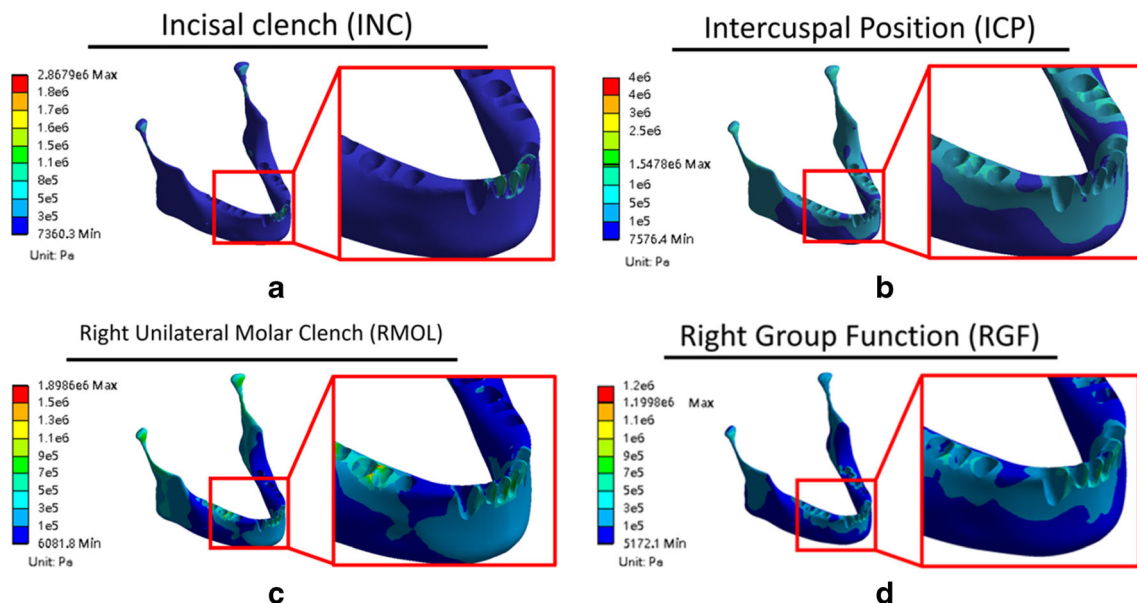
**Fig. 4** von Mises stress on cortical bone in different occlusal modes: **a** INC, **b** ICP, **c** RMOL, and **d** RGF

Thus, we recommend that when using an FE model to analyze the biomechanical response to orthodontic treatments, the effect of muscle forces caused by occlusion must also be considered.

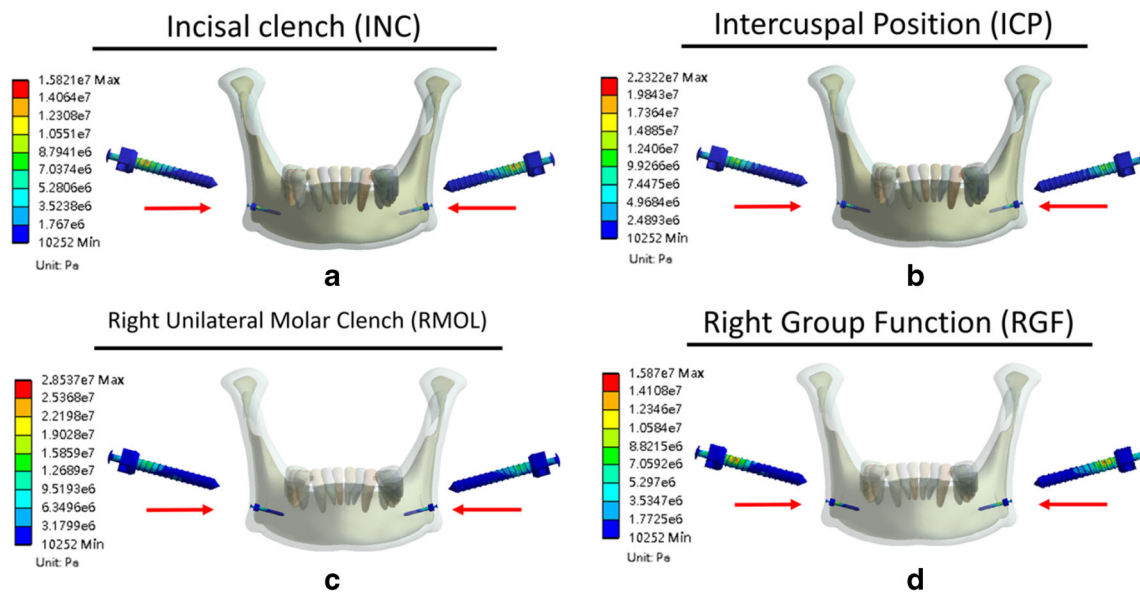
The human skull has several types of muscles with different functions. Chewing, mouth opening and closing, and occlusion mainly rely on forces generated by muscle contraction in the mandible. Mandibular muscles that are responsible for chewing are the masseter, temporalis, and pterygoid muscles. Anatomical information reveals the direction of each muscle contraction and the area that contacts with the mandible. To assess muscle strength, researchers have used electromyography to measure the signals generated by the chewing muscles

in different occlusal modes and convert the signal into force to determine the force exerted by each muscle [25, 34–37].

The finite element method is a very powerful tool for the analysis of orthopedics and dental biomechanics [38–40]. Approximately two decades ago, scholars developed an FE model of the entire mandible to simulate the effects of different occlusal modes. Koriath and Hannam [25] modeled a complete mandible and nine muscles to investigate the biomechanical effect of muscle contraction strength on the mandible. Koolstra and Van Eijden [35] also generated a complete mandibular model and developed tooth and PDL models to simulate biomechanical effects of four different occlusal modes on the PDL and surrounding bone tissue. Huang



**Fig. 5** von Mises stress on cancellous bone in different occlusal modes: **a** INC, **b** ICP, **c** RMOL, and **d** RGF



**Fig. 6** von Mises stress on the miniscrew in different occlusal modes: **a** INC, **b** ICP, **c** RMOL, and **d** RGF

et al. [26] used an FE model to discuss the biomechanical effect of unilateral artificial temporomandibular joint replacement in different occlusal modes. Hijazi et al. [41] also created a complete mandibular model to simulate the biomechanical effects of different occlusal modes on bone nails and plates for repairing mandibular fractures.

The process of moving teeth in an orthodontic treatment is based on pressure–tension theory, which explained that the resorption or deposition of bones around the PDL is caused by compression or tension on the PDL [1–4]. Therefore, the mechanical behavior of the PDL is crucial. In FE models, material property settings have considerable effects on simulation results. Researchers have hypothesized that the PDL is a linear elastic material. For example, Motoyoshi et al. [31] established a partial mandibular model with teeth and the PDL to analyze the biomechanical effects of miniscrews on the peripheral bone; their PDL was hypothesized to be a linear elastic material. Sardarian et al. [9] established a partial mandibular FE model for teeth, the PDL, and orthodontic brackets and wires to analyze the effects orthodontic treatments on the PDL. Sardarian et al. [9] and Gupta et al. [42] also hypothesized that the PDL was a linear elastic material. Additionally, Caballero et al. [12] and Bouton et al. [8] used similar methods to analyze the biomechanical effect of orthodontic forces on tooth movement. Although most studies have hypothesized the PDL to be a linear elastic material, the physiological structure of the PDL is a connective tissue composed of a considerable amount of collagen fiber [38, 43–45]. Therefore, researchers have inferred that the PDL has nonlinear elasticity. Researchers have used human or animal experiments to directly measure PDL material properties. For example, Poppe et al. [22] conducted mechanical experiments on mandibular fragments of human cadavers to directly measure the elastic

modulus of the PDL. Experimental results of Poppe et al. have suggested that the PDL has bilinear elasticity. Kawarizadeh et al. [24] conducted mechanical experiments on rat mandibles to demonstrate that the PDL of rats has bilinear elasticity. Similarly, Ziegler et al. [46] conducted experiments on miniature pig bones and indicated that their PDL has bilinear elasticity. Therefore, researchers increasingly simulate the PDL as a bilinear elastic material in FE models. For example, Poppe et al. [22] established a partial mandibular, tooth, and PDL FE model to simulate the effect of tooth movement on the PDL. Similarly, Kawarizadeh et al. [24], Ziegler et al. [46], and Papageorgiou et al. [15, 18] established partial mandibular FE models to investigate the biomechanical effects on the PDL after exerting orthodontic force on teeth. Moreover, Kettenbeil et al. [47] developed a complete mandibular model to analyze the effect of orthodontic force on the PDL during orthodontic treatment.

Some researchers have established complete mandibular FE models to conduct biomechanical analyses of different occlusal modes on the mandible or artificial replacements and bone plates [26, 41, 48]. Several researchers have also developed linear or nonlinear elastic partial mandibular, tooth, and PDL models to analyze the effect of orthodontic forces on the PDL [8–14, 17]. However, they have not used a complete mandibular model to simulate the biomechanical effects of orthodontic forces on the PDL in different occlusal modes. Therefore, we addressed this topic and used von Mises strain on the PDL as an evaluation factor in accordance with previous studies [1, 18, 27, 28], and von Mises stress was an evaluation indicator for bones and miniscrews [29–32].

The simulation results suggest that von Mises strain on the PDL was affected by various occlusal modes. The von Mises strains on the PDL of the right canine tooth in the ICP and

RGF occlusal modes were 4.64 and 2.87 times larger, respectively, than those in a model with orthodontic force alone. However, the effects of the INC and RMOL occlusal modes were relatively small because, in these occlusal modes, the canine tooth crown is not a constrained area. In other words, in different occlusal modes, when patients do not directly bite onto a canine tooth, the von Mises strain on the PDL of the tooth is small. Therefore, when researchers use the FE method to simulate orthodontic tooth movement, they should create a complete mandibular model and simulate muscle forces that approximate the behavior of occlusal modes used in daily life to realistically simulate an orthodontic treatment.

Regarding the effects of the four occlusal modes on the mandible, both cortical and cancellous bone exhibited high von Mises stress concentrated at the mandible neck. Although the distributions of von Mises stress on cortical and cancellous bone in the four occlusal modes differed slightly, the largest von Mises stresses on cancellous and cortical bones surrounding the canine tooth were less than 1 MPa and 10 MPa, respectively. The muscle force in INC and ICP was left–right symmetrical. Therefore, von Mises stress on the bone near the canine tooth was symmetrical. Because the RMOL occlusal mode does not involve the left canine tooth, almost no differences were observed in the von Mises stresses on the bones surrounding the left and right canine teeth. Because the muscle force of the RGF occlusion mode is concentrated on the right side, the von Mises stress on the bone near the right canine tooth was greater than that on the bone near the left canine tooth.

The simulation results indicated that the von Mises stress on the miniscrew was similar in the four occlusal modes. Because we completely constrained the top of the mandibular condyle, sizeable twists would not be generated after the occlusal force was exerted on the mandible, and the four occlusal modes all had a fixed area at the tooth crown. Therefore, the effect of the four occlusion modes on the von Mises stress on the miniscrew was extremely small. The von Mises stresses of the four occlusal modes on the miniscrew were approximately 30 MPa, which was considerably lower than the failure strength of the metal [49]. Therefore, the miniscrew was unlikely to break regardless of the occlusal force.

The FE simulation results were affected by many factors (e.g., the material properties and the boundary and loading conditions of the FE model). The main purpose of this study was to provide information for researchers evaluating the biomechanical response of the PDL under orthodontic force. The occlusal modes clearly affected the simulation results. Therefore, researchers should pay more attention to boundary and loading conditions.

As with any FE simulation approach, our study had several limitations. First, not only is the human PDL a nonlinear material, but it also exhibits viscoelasticity. However, similar to previous studies [2, 8, 11, 16, 24, 30, 34, 35, 41, 50], we assumed that the PDL is a nonviscoelastic material. Second, the loading

condition applied in the FE model mimicked long-term loading; however, biting is a short-term process. Nevertheless, avoiding this simplification is difficult, and many researchers who have conducted biomechanical analyses of dental implants during chewing also applied maximal force on the dental implant [51]. Third, the thickness of the outer cortical layer of the mandible in this study was a consistent 2 mm, and the material properties of cortical and cancellous bone in the model were homogeneous, isotropic, and linearly elastic. These properties differ from those of actual human bone. Fourth, we simulated only one type of orthodontic force in a specific direction. Similar to previous studies, the present study hypothesized that teeth are a homogeneous material, and their structure was not separated into dentin and enamel. In addition, only four occlusion modes common in daily life were selected, and only six major muscle strengths were simulated. Although the simulation results of the FE model did not exactly simulate real-life situations because of some simplifications, the results indicate a clear trend for the investigated topic.

**Conclusion** The von Mises strain on the PDL of a tooth undergoing orthodontic treatment varied considerably between four occlusal modes, namely INC, ICP, RMOL, and RGF. Among the four modes, ICP produced the highest maximum von Mises strain on the PDL (1.81). By contrast, RMOL exerted the lowest maximum von Mises strain on the PDL (0.39). Thus, during orthodontic treatment, the maximum von Mises strain on the PDL is affected by the occlusion modes. This implies that when researchers plan to simulate orthodontic tooth movement by using an FE model, they should develop a complete mandibular model and consider muscle forces to realistically simulate an orthodontic treatment.

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## Declarations

**Ethical approval** This article does not contain any studies with human participants or animals performed by any of the authors.

**Informed consent** For this type of study, formal consent is not required.

**Conflict of interest** The authors declare no conflict of interest.

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