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Determination of vibration acceleration mechanism and vibration load application duration from a non-biological perspective: Orthodontic Acceleration

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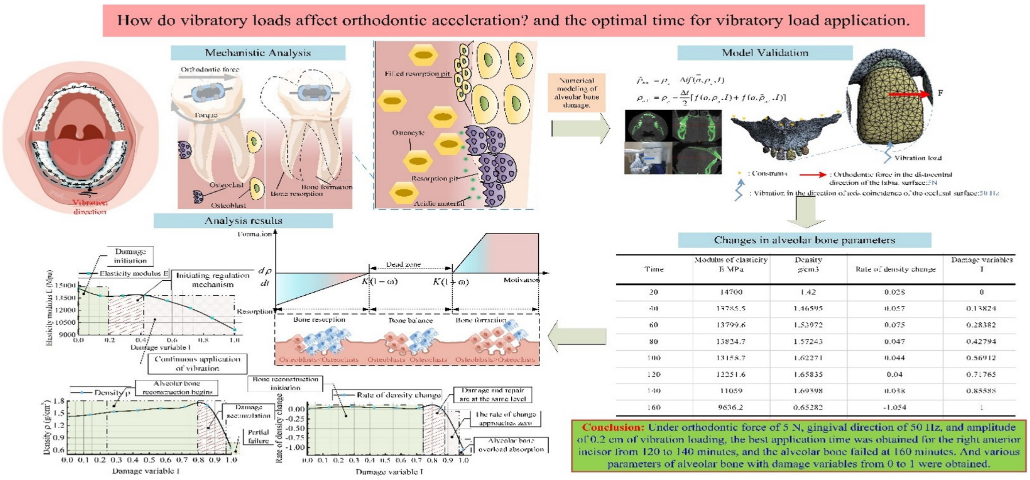
Abstract

Compared to other orthodontic acceleration methods such as drug, electric current, and laser, vibratory loading is less invasive and easier to use. But the optimal duration for vibratory load application has not been determined, nor can the alveolar bone parameters be predicted after vibratory load application. Therefore, this work examined the mechanism of vibration-accelerated alveolar bone reconstruction and established a numerical model for simulating alveolar bone damage caused by vibration loads. That is, the role of vibration load in orthodontic acceleration was analyzed, and a finite element model was established to validate the vibra-tion-accelerated orthodontic mechanism with a simulated numerical model of alveolar bone damage. The optimal duration of application was obtained for the right anterior incisor under vibratory loading of 5 N orthodontic force, 50 Hz in the gingival orientation and 0.2 cm amplitude for 120 to 140 minutes. This work is of guidance and reference significance in promoting the development of orthodontic treatment and shortening the orthodontic treatment cycle.

Keywords

Orthodontic acceleration; tooth movement; bone reconstruction; mechanical vibration; finite element analysis.

Graphical Abstract



Changes in alveolar bone parameters

Time	Modulus of elasticity E MPa	Density g/cm3	Rate of density change	Damage variables I
20	14700	1.42	0.028	0
40	13785.5	1.46595	0.057	0.13824
60	13799.6	1.53972	0.075	0.28382
80	13824.7	1.57243	0.047	0.42794
100	13158.7	1.62271	0.044	0.56912
120	12251.6	1.65835	0.04	0.71765
140	11059	1.69398	0.038	0.85588
160	9636.2	0.65282	-1.054	1

Conclusion: Under orthodontic force of 5 N, gingival direction of 50 Hz, and amplitude of 0.2 cm of vibration loading, the best application time was obtained for the right anterior incisor from 120 to 140 minutes, and the alveolar bone failed at 160 minutes. And various parameters of alveolar bone with damage variables from 0 to 1 were obtained.

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1. Introduction

Malocclusion endangers human health and is one of the three major oral diseases (Suhail Y, 2020), usually manifesting as symptoms such as abnormal tooth alignment and jaw surface abnormalities. Some reports (Asiri SN, 2019) showed that, of the 8804 untreated US adults evaluated by the NHANES III, almost two-thirds have at least one or more forms of malocclusion. There is age, ethnic, and sex differences in the prevalence of clinically meaningful malocclusions that characterize approximately two-thirds of untreated US adults. The reasons for the low treatment rate are divided into the following two main points. On the one hand, the rate of tooth movement during orthodontic treatment is generally one millimeter per month (Firth FA, 2019; Zhou Y, 2014), and the whole treatment cycle usually lasts 1-3 years (Briseno-Marroquin B, 2021). On the other hand, prolonged orthodontic treatment may cause side effects such as enamel demineralization, root aspiration, and periodontitis (Unnam D, 2018). Therefore, it is especially important to speed up the movement of teeth during orthodontic treatment and shorten the treatment period (Li SX, 2021). Currently, drug, electric current, weak laser, magnetic field, ultrasound and vibratory loading methods (Caccianiga G, 2019; Kalemaj Z, 2015; Babanouri N, 2020; Sonesson, M, 2017; Jing, D 2017) have been proven to be used as orthodontic acceleration, among which vibratory loading has the advantages of being less invasive, non-drug, and easy to adjust parameters relatively easy for patients to accept (Chidchanok L, 2018). Many biological experiments (Van SA, 2013; Cai YQ, 2021) have shown that applying a certain intensity of vibratory load during the movement of orthodontic teeth can increase the metabolism of alveolar bone and periodontal tissue, accelerate the movement of orthodontic teeth, and also make the structure of the teeth more stable after orthodontic treatment (Feng XD, 2014). Acceledent, a vibration-assisted orthodontic acceleration device from the United States, is the only orthodontically assisted acceleration device currently certified by the Food and Drug Administration. Although the effectiveness of vibration therapy has been confirmed in animal studies and some clinical studies have shown that this intervention accelerates tooth movement, several randomized controlled trials conducted in recent years have come to the negative conclusion that the vibration load of the adjunctive treatment does not accelerate tooth movement and shorten the treatment period. Since these clinical studies used the AcceleDent as the vibratory treatment device, the effectiveness of the AcceleDent itself is also not promising. It is speculated that the duration of vibration load application is too conservative, however, the longer the duration of vibration load application, the better. Therefore, there is a need to investigate the optimal duration of vibratory loading and when the rate of alveolar bone reconstruction is maximized in order to provide guidelines for clinical or numerical studies of orthodontic acceleration.

Vibratory loads are dynamic loads, and a work (Yadav S, 2015) has concluded that dynamic loads are more beneficial to bone growth than static loads. The low-intensity mechanical signal formed by the stimulated bone depends on the frequency of the mechanical signal rather than the strain amplitude (Judex S, 2018; Shipley T, 2019). In addition to accelerating orthodontic teeth movement (Chen Y, 2020), vibrations can also affect osteoclast activity in the maxilla and is sensitive to the intensity of strain localized in that zone (Fujiki K, 2013). There is an obvious lack of data on the changes in alveolar bone parameters during orthodontic acceleration in most existing research studies using biological models. Neither the mechanism nor the efficacy of vibration-accelerated orthodontic treatment has been determined (Takano-Yamamoto T, 2017; Woodhouse NR, 2015), leading to controversy regarding its efficacy. So the vibration load application time is more conservative, about 20 minutes (Dubravko P, 2012). The optimal duration of vibration load application must be determined by simulating the alveolar bone state. Based on a mechanical steady-state theory (Frost HM, 1987), existing studies (Carter DR, 1987) simulated the density and alignment of bone trabeculae by numerical equations to derive expressions for bone density changes. Considering the process of bone reconstruction as an interval function in the form of an interval, when the excitation value is in the interval range, the alveolar bone will not be reconstructed, and the concept of the dead zone was proposed (Weinans H, 1992). Some scholars have also proposed damage variable formulas (Hambli R, 2011) using stress as an excitation, but none of them has been described for the relationship between vibratory load parameters, alveolar bone parameters, and applied vibration time. In summary, existing studies have failed to determine the variation of parameters in the alveolar bone at different durations of vibratory load application and therefore cannot determine the optimal duration of vibratory load application, which is often the most pressing need for physicians and patients in clinical applications.

In response to the shortcomings of the current works, the present work aims to better understand the mechanism of alveolar bone reconstruction under vibratory loading and its optimal time for applying vibratory loading. To do this, we start with the analysis of orthodontic correction to determine the role of vibratory loads in orthodontics. More in detail, we believe that vibratory loads play a role in orthodontically accelerated alveolar bone reconstruction as a mechanical signal. In addition, to obtain the parameter changes of the alveolar bone after the applied vibration load, we introduce the damage variables and establish the fatigue damage theory to establish the numerical equations of alveolar bone damage simulation. By integrating all these elements, we first built a dental and jaw model to confirm that the dental and jaw model can support the later experiments by comparing the theoretical results obtained with the real dental and jaw model (Horina JL, 2018). Then, the role of vibration load in orthodontic acceleration is verified by finite element analysis. Then the finite element simulation model will be established by the simulation software combined with the numerical equation of alveolar bone damage simulation. We explore the trend of vibration load application time on parameters such as bone density and elastic modulus of alveolar bone. The validity of the orthodontic mechanism of vibration load acceleration and the numerical model of alveolar bone damage simulation is demonstrated, and the optimal duration of vibration load application is determined by analyzing the parameters of alveolar bone.

2. Materials and Methods

2.1. Analysis of the orthodontic mechanism accelerated by vibratory loading

The movement of orthodontic teeth is a mechanical adaptation of the alveolar bone and periodontal tissues to orthodontic forces under the action of orthodontic forces, as a result of the mechanical adaptation of the orthodontic teeth to alterations in a new position (Garlet TP, 2007). The alveolar bone is part of the human skeleton, which is regulated mechanically by the mechanism depicted in Fig. 1. When orthodontic teeth undergo torque and orthodontic forces exerted by the archwire, the bone tissue modulation mechanism is triggered. At this stage, the bone tissue error and stress state are detected; the increased and decreased bone mass are then determined. The alveolar bone is the most changeable part of the bone tissue in the whole body, therefore the alveolar bone adapts to the existing mechanical state and undergoes reconstruction when stimulated by excitation, bone formation occurs on the tension side, and bone resorption on the pressure side. The principle of orthodontic treatment is adapting to the mechanics of this time and restoring bone homeostasis. Reconstruction of bone is a force biological process that occurs when the alveolar bone is remodeled. From the mechanical point of view, the alveolar bone adjusts the mechanical environment to the new orthodontic position by increasing or decreasing the bone mass, and the mechanical stimulus determines the direction of adjustment of the bone mass increase or decrease. From a biological point of view, the alveolar bone tissue senses micromechanical signals, transduces them into biochemical signals, and then transmits them to the effector cells to complete the adjustment of bone mass increase or decrease through the bone reconstruction mechanism. Therefore, accelerating orthodontics means accelerating the adjustment of bone mass increase or decrease, and mechanical signal perception conduction is the key to orthodontic tooth movement.

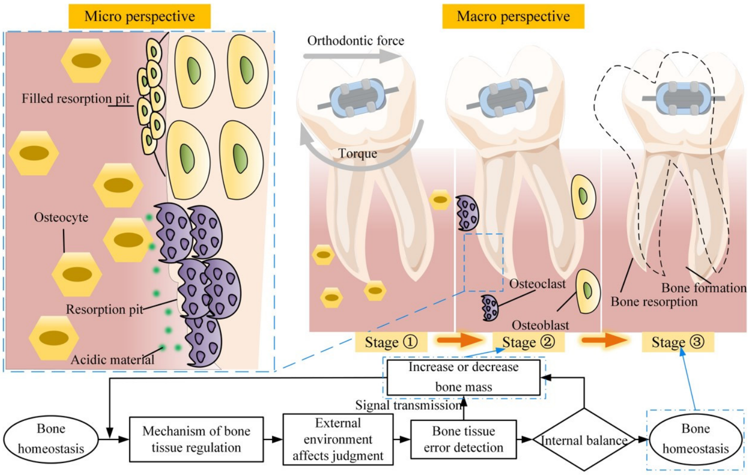

Figure 1 (Draw by Figdraw). Schematic diagram of orthodontic treatment and mechanism of mechanical regulation.

Vibratory loading serves to enhance the mechanical signal in orthodontically accelerated alveolar bone reconstruction. It has been suggested that the mechanical sensitivity of skeletal cells to dynamic loading is higher than static loading and would promote the opening of the alveolar bone for reconstruction (Lanyon LE, 1984; Judex S, 2007). Under the action of external dynamic loads, bone materials can also be fatigued like ordinary materials. The accumulation of fatigue damage in bone tissue over a long period can cause a loss of bone mass and result in a decrease in elastic modulus, which in turn can affect the biological properties of bone tissue. Unlike ordinary materials, bone material is a "living" material that can adjust its structure to adapt to changes in the external environment without exceeding its physiological load limit, which means that bone tissue has a strong self-healing ability. After applying the orthodontic force and torque shown in Figure 2a, it is known that tension zone AD and pressure zone BC are generated on the PDL (periodontal ligament) and CAB (cancellous bone) of the alveolar bone according to the movement trend of the orthodontic teeth. On a microscopic level, orthodontic teeth increase the concentration of osteoclasts and osteoblasts within the bone tissue when subjected to vibratory loads. In the tension zone, the number of osteoblasts is greater than the number of osteoclasts, and osteoblasts cells promote bone accumulation in the tension zone of the tooth. In the pressure zone, the number of osteoblasts is greater than the number of osteoclasts, and osteoclasts promote bone tissue dissolution in the pressure zone of the tooth. From the macroscopic point of view, the high-frequency vibration of the vibratory load on the orthodontic teeth will cause fatigue and damage to the alveolar bone, and we think that the process of promoting the reconstruction of the alveolar bone under the action of the vibratory load is the process of fatigue and damage to the alveolar bone after being stimulated by external excitation and repair. After bone tissue fatigue, from a material point of view, the bone in the pressure zone loosens and the orthodontic teeth move under orthodontic forces, thus adapting to the new mechanical state, and the tension zone increases as the mass of bone in the pressure zone decreases. Combining the concept of the dead zone (Weinans H, 1992) with the physiological properties of the alveolar bone, it is easy to see that the vibratory load is not applied to the alveolar bone for as long as possible. In this work, we argue that there must be an optimal solution for the time of vibration load application, i.e., when the number of osteoclasts and osteoblasts in the tension and pressure zones reaches the most suitable state for the modification of the alveolar bone in the new mechanical state. Since the properties of bone material are constantly changing with its metabolism, we can only observe the macroscopic results of the alveolar bone reconstruction process, so this work also addresses the material properties in a macroscopic sense. Depending on the location of load application which also affects osteoclast resorption (Fujiki K, 2013), the direction of vibratory load application is divided into the lingual, lip, and gingival orientation in this work, as shown in Figure 2b.

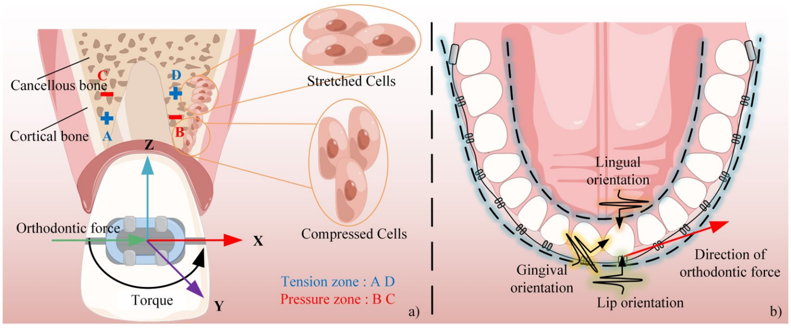

Figure 2 (Draw by Figdraw). Cell state analysis and dental orientation definition. (a) Theoretical tension zone and theoretical pressure zone distribution diagram; (b) Schematic diagram of the tooth orientation to which the vibration load is applied.

2.2. Numerical modeling of alveolar bone damage simulation

To establish a numerical model for simulating the damage of alveolar bone under the action of vibration load, the damage variable (Hambli R, 2011) is introduced into this work, and based on it, a fatigue damage control equation applicable to the alveolar bone under the action of vibration load is proposed to simulate the bone damage behavior of alveolar bone.

$$I = 1 - \left[1 - \left(\frac{Tf}{N_{\max}} \right)^{\frac{1}{1-\alpha}} \right]^{\frac{1}{1+\beta}} \tag{1}$$

where, I is the damage variable; T is the external vibration load duration; f is the frequency of vibration; $N_{\max}$ is the number of failure vibrations; α and β is a constant, $\alpha = 0.01$, $\beta = 0.001$.

Considering that the damage variable is determined by the number of vibrations and the time of applied vibrations, the damage variable is considered an isotropic parameter. Eq. (1) responds to the variation law of damage with the frequency of vibration and the time of applied vibration.

The alveolar bone reconstruction process conforms to the elastic mechanical characteristics of the material, therefore, from the analysis of elastic mechanics, stress is the basic unit to represent the force, based on which the mechanical parameters such as strain and strain energy density at that point can be obtained. Considering the mechanical characteristics of the alveolar bone tissue, we consider that stress, strain, strain energy, and strain energy density are all functions of excitation, and the equivalent force can be used as the most basic external mechanical excitation in the alveolar bone reconstruction process, so the Von. Mises yield criterion is introduced. The equation for the calculation of Von. Mises stress is:

$$\bar{a} = \frac{1}{\sqrt{2}}\sqrt{(a_1 - a_2)^2 + (a_2 - a_3)^2 + (a_3 - a_1)^2} \tag{2}$$

where, a_1, a_2, a_3 are the first, second, and third principal stresses, respectively. The meaning is that under certain deformation conditions when the elastic deformation energy of the material per unit volume shape change reaches a certain constant, the material will yield.

On this basis, we introduce the classical bone reconstruction control equation (Carter DR, 1987) proposed by Carter in this work, and on its basis, we propose a bone reconstruction theoretical equation based on Von. Mises stress as an external excitation representation quantity:

$$\frac{d\rho}{dt} = A(\bar{a} - K) \tag{3}$$

where, ρ is the apparent density, which can be associated with the elastic modulus to express the strength of the bone tissue; A is the reconstruction coefficient of the bone, which is time-dependent and whose expression is: $A(t) = p + qe^{-Rt}$, a constant in this work; $\bar{a}$ is the Von. Mises stress value of the bone tissue; K is the steady-state reference value, which refers to the stress value to which the bone tissue is subjected when its internal physiological environment is in equilibrium.

Based on this, we introduce the concept of dead zone (Weinans H, 1992), a theory that suggests that bone volume increases with external excitation during orthodontic treatment and that bone reconstruction does not occur when excitation $\bar{a}$ is between $K(1 \pm \omega)$. Eq. (3) is refined.

$$\frac{d\rho}{dt} = \begin{cases} A[\bar{a} - K(1-\omega)], \bar{a} < K(1-\omega) \\ 0, K(1-\omega) \le \bar{a} \le K(1+\omega) \\ A[\bar{a} - K(1+\omega)], \bar{a} > K(1+\omega) \end{cases} \tag{4}$$

Based on Eq. (4), the damage variable I is introduced as an influencing factor in the orthodontic accelerated reconstruction model of orthodontic bone under the action of vibration load, to express the effect of vibration load on the reconstruction process of orthodontic bone in the form of fatigue damage.

By alveolar bone fatigue and damage repair, the bone damage mechanism can start the bone reconstruction process and can accelerate bone production, considering that bone damage is difficult to repair when it is too large, it is believed that there should be a threshold value of damage, within the threshold value bone fatigue can accelerate the rate of change of bone density of alveolar bone, beyond the threshold value, the rate of change of bone density will gradually decrease to 0. Excessive application of vibratory loads may then cause the rate of change in bone density to become negative, i.e., produce overload resorption. Eq. (4) is improved, i.e., the numerical model for alveolar bone reconstruction under vibratory loading is:

$$\frac{d\rho}{dt} = \begin{cases} A[\bar{a} - K(1-\omega)] - BI[\bar{a} - K(1-\omega)]^2, \bar{a} < K(1-\omega) \\ 0, K(1-\omega) \le \bar{a} \le K(1+\omega) \\ A[\bar{a} - K(1+\omega)] - BI[\bar{a} - K(1+\omega)]^2, \bar{a} > K(1+\omega) \end{cases} \tag{5}$$

where, ρ is the apparent density; A and B are the reconstruction coefficients of bone, which are time-dependent and constant in this work; $\bar{a}$ is the Von. Mises stress value of bone tissue; K is the steady-state reference value, which refers to the stress value to which bone tissue is subjected when its internal physiological environment is in equilibrium. ω indicates the range of the dead zone, when the excitation $\bar{a}$ is between $K(1\pm\omega)$, no bone reconstruction occurs, and $\omega = 0.1$ in this work.

In this work, based on the dead zone concept (Weinans H, 1992), bone tissue is considered as a continuous, isotropic material, and the strain energy density is selected as the mechanical excitation to simulate alveolar bone reconstruction based on the damage model under vibratory loading:

$$S = U/\rho \tag{6}$$

where, S is the mechanical excitation, can be equivalent effect force, equivalent effect variation, or strain energy density, this work takes the mechanical excitation as equivalent effect force.

The strain energy density is chosen to be expressed in terms of current stress and strain, where only the form of mechanical excitation applied on the gingival orientation of the tooth is considered, the strain energy can be expressed as:

$$U = \bar{a}\varepsilon/2 = \bar{a}^{2}/2E \tag{7}$$

The mechanical properties of bone are the focus of attention during bone reconstruction simulations. The relationship between the elastic modulus of bone tissue and the apparent density is expressed as:

$$E = E(\rho) = C\rho^{r} \tag{8}$$

where, C and r are constants. In this work, considering the influence of damage variables on the elastic modulus, the representation of the elastic modulus was rewritten as:

$$E = E(\rho) = C(1-I)\rho^{r} \tag{9}$$

where, $C = 3790Mpa/(g\cdot cm^{-3})^{3}$, $r = 3$. Bringing the modulus of elasticity Eq. (9) considering damage variables into Eq. (7), the expression for strain energy with damage variables is obtained as:

$$U = \frac{\bar{a}^{2}}{2\mathrm{C}(1\text{-}I)\,\rho^{3}} \tag{10}$$

Taking Eq. (10) into Eq. (6) to derive the excitation yields:

$$S = \frac{\bar{a}^{2}}{2C(1-I)\rho^{4}} \tag{11}$$

Bringing the excitation Eq. (11) into the numerical model of alveolar bone reconstruction under vibration load Eq. (5), we get:

$$\frac{d\rho}{dt} = f(\bar{a},\rho,I) = A\left(\frac{\bar{a}^{2}}{2C(1-I)\rho^{4}} - 1.1\times K\right) - BI\left(\frac{\bar{a}^{2}}{2C(1-I)\rho^{4}} - 1.1\times K\right)^{2} \tag{12}$$

To improve the accuracy of the calculation process and reduce the truncation error, the differential Eq. (12) is calculated using the modified Euler algorithm to obtain:

$$\tilde{\rho}_{n+1} = \rho_n + \Delta t f(\bar{a}, \rho_n, I) \tag{13}$$

$$\rho_{n+1} = \rho_n + \frac{\Delta t}{2}[f(\bar{a}, \rho_n, I) + f(\bar{a}, \tilde{\rho}_{n+1}, I)] \tag{14}$$

Eq. (14) is an iterative equation for the density of the alveolar bone tissue under vibratory loading, representing the change in bone density at each stage, which in turn exhibits the change in the internal elastic modulus of the bone tissue. Eq. (13) is the predicted value using Euler's formula, and Eq. (14) is the corrected value using the trapezoidal formula.

3. Results

3.1. Effective modeling of jaw bone simulation

The patient's CBCT images were first reconstructed in 3D using Mimics Medical 21.0, as shown in Figure 3a, and then optimized and smoothed using Geomagic Studio to obtain models of the maxillary teeth as well as the alveolar bone. The CAB model was divided and imported into the 3D processing software, and the CB (cortical bone) part of the alveolar bone was obtained by Boolean operation. The thickness of the PDL was set to 0.2 mm by real PDL thickness (Liu WY, 2013) and combined with the observation of CBCT images. The established parts were assembled to obtain the dental 3D model, as shown in Figure 3b.

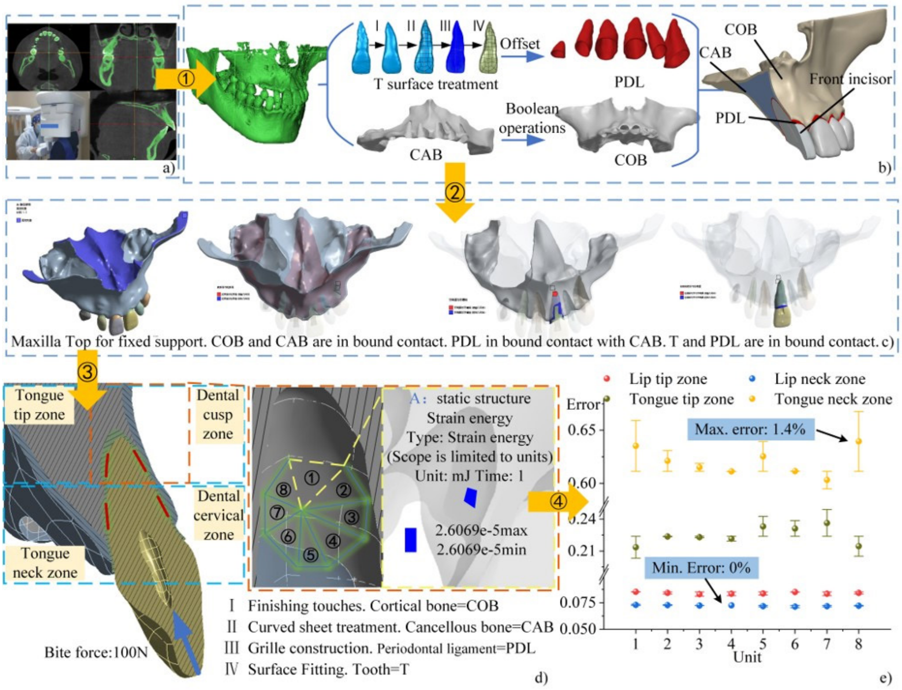

Figure 3 Model processing. (a) Data source; (b) Mandibular segmentation treatment; (c) Types of constraints between components; (d) Strain energy density method analysis; (e) Error analysis between 3D model and real model.

Table 1 Material parameters of jaw model.

Materials	Modulus of elasticity E (MPa)	Poisson's ratio µ
CAB	1470	0.3 (Holmes DC,1997; Helgason B, 2008)
CB	14700	0.3 (Staninec M, 2010; Lin CL, 2016)
Teeth	18600	0.3 (Staninec M, 2010; Lin CL, 2016)
PDL	0.69	0.45 (Holmes DC, 1997)

The established 3D model was imported into Ansys Workbench, and the material was assumed to be continuous, homogeneous, and isotropic, and the specific model material parameters were shown in Table 1. The biological connection relationship was analyzed and the connection type was set as shown in Figure 3c. The strain energy density method was used to verify the validity of the model, i.e., gingival orientation pressure was applied to the incisive end of the orthodontic tooth, as shown in Figure 3d, and the stress distribution on the dental model was observed. So the right anterior incisor, alveolar bone, and periodontal tissue were selected as the observed targets of the simulation, and a vertical load of 100 N was applied to the gingival orientation of the teeth. Analysis of the force situation showed that after applying the gingival orientation pressure, the alveolar bone force area was divided into four zones shown in Figure 2d, and each zone was divided into 8 units. The strain energy density was obtained as 0.0853 Kpa for the area of unit 1 in the lip tip zone, for example, after which the static structure simulation was performed for the remaining 31 units and the average value of the strain energy density was obtained. The error plot was plotted as shown in Figure 3e, and the maximum error of the obtained simulation results with the dental and jaw bone model data (Horina JL, 2018) was 1.4% and the minimum error was 0%, which can be known to provide model support for the subsequent simulation work.

3.2. Vibration load accelerated orthodontic mechanism validation

To verify the mechanism of accelerated orthodontics by vibratory loading proposed in Section 2.1, the fatigue damage of periodontal tissue under dynamic and static loading was verified. The average stress is an important factor in the fatigue damage of materials caused by cyclic loading (Lee TC, 2010), so it is used as an observation. Since the process of alveolar bone reconstruction is a process of material reconstruction, the Von Mises stress is set as another observation.It was assumed that the teeth were subjected to the orthodontic forces generated by the sequence curvature, excluding the effect of torque. In the static analysis, the right anterior incisor was selected to apply an orthodontic force of size 5 N in the distal-central direction of the labial surface, and a high-value low-frequency vibration load of 50 Hz with a vibration amplitude of 0.2 cm was applied to the gingival orientation (Li SX, 2021). The stress distribution after applying the force was shown in Figure 4. It can be seen that the Von Mises stress distribution cannot clearly distinguish between the tension and pressure zones, so the X-direction stress was used as the external excitation characterization quantity.

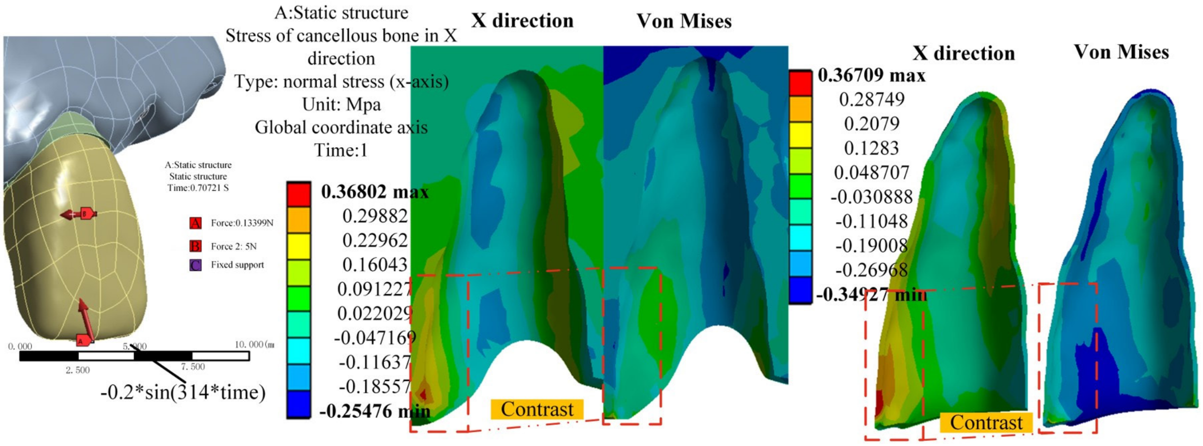

Figure 4 Cloud map of stress distribution in CAB and PDL.

The same load was applied again to the right anterior incisor for static analysis. To improve the calculation accuracy and reduce the simulation time, only the tension zone AD and the pressure zone BC were selected for the observation area. The stress results were extracted and the stress distribution clouds were shown in Figure 5a, which were consistent with the distribution of theoretical tension and pressure zones presented in Section 2.1. It can be seen that the application of vibratory loads did not change the forces on the orthodontic teeth. The average stress values in the four zones of the alveolar bone and PDL were recorded in Ansys Workbench, and the images were plotted as shown in Figure 5b, and the curves all showed asymmetric periodic variations. From a mechanical point of view, asymmetric cycling is more likely to lead to lower fatigue life of the material (Cheng JS, 2015). From an engineering perspective, the combined dynamic and static loading increases the degree of fatigue damage to the bone tissue, and fatigue damage initiates and promotes the bone reconstruction process. Biologically, in conventional orthodontic treatment, although the loaded lateral static orthodontic force initiates the orthodontic tooth movement mechanism, the modulation of bone reconstruction is weak, and the high mechanical sensitivity of skeletal cells to the loaded low-value high-frequency vibratory load enhances the active bone reconstruction, thus accelerating the orthodontic tooth movement. In contrast, the cyclic load of the asymmetric cycle is more easily perceived by the skeletal cells. Therefore, it is shown that the combined dynamic and static loading of lateral static orthodontic forces combined with low-value high-frequency mechanical vibration loads is a feasible method to accelerate orthodontic tooth movement.

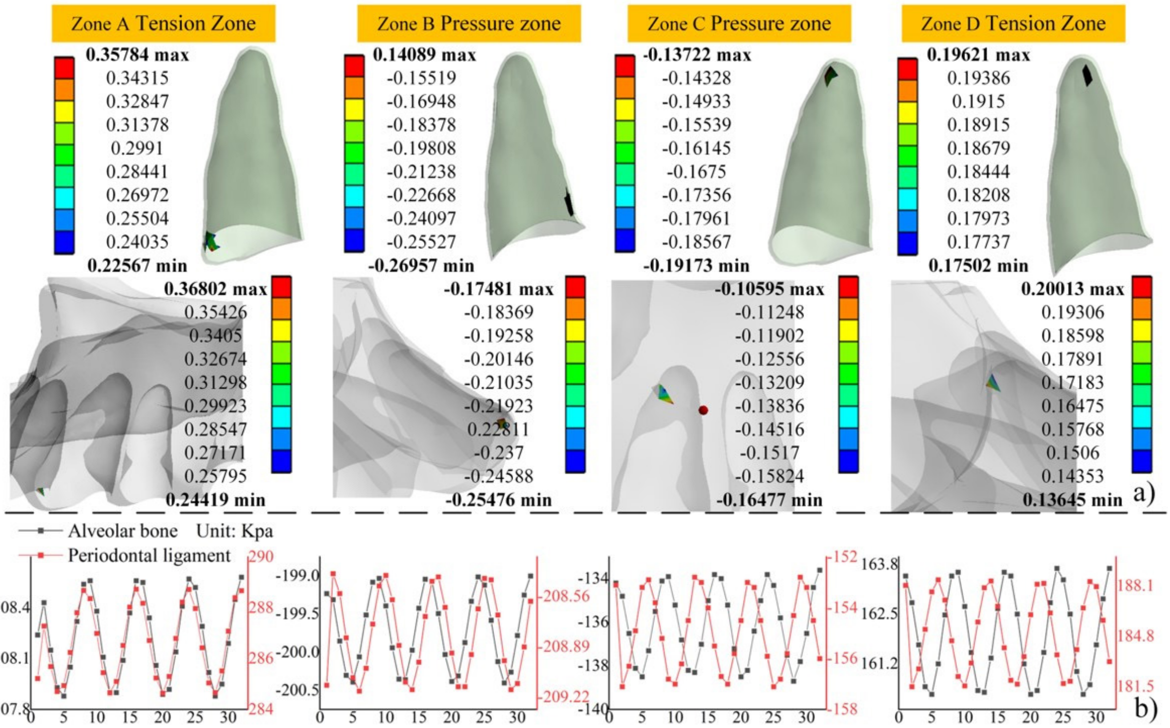

Figure 5 Simulation results. (a) X-directional stress cloud of the PDL and alveolar bone of orthodontic teeth; (b) Average stress curve of the PDL and alveolar bone in the X-direction of orthodontic teeth.

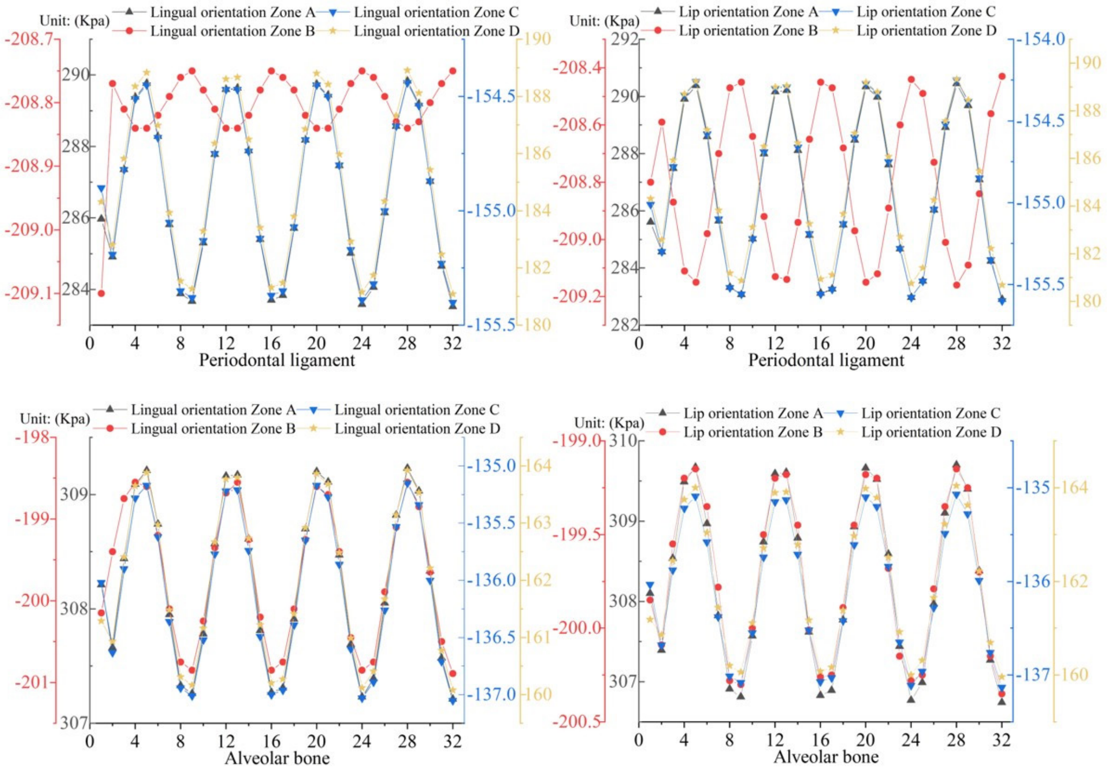

Figure 6 X-direction mean stress - timeline of alveolar bone and PDL.

Table 2 Displacement for different types of loads.

Type	Max. displacement (mm)	Min. displacement (mm)
Forward orthodontic force 5N	1.5962	0.02857
Forward orthodontic force + Gingival orientation	1.6683	0.02968
Forward orthodontic force + Lip orientation	1.6861	0.02968
Forward orthodontic force + Lingual orientation	1.6837	0.02971

To verify that the vibration applied to the labial and lingual orientations can also play the role of orthodontic acceleration, the same load as the gingival orientation was applied to them respectively, and their average stress-time curves were obtained with asymmetric periodic variations, as shown in Figure 6, yielding the same effect of orthodontic acceleration. For the undetermined effect of vibration applied to different tooth surfaces, this work solved the displacement magnitude for the combination of 5N forward orthodontic force and vibration loads of 50Hz and 0.2cm amplitude in different directions, and the specific data were shown in Table 2.

3.3. Simulation of alveolar bone reconstruction under vibratory loading

The established 3D model was imported into Ansys Workbench, and to reduce the computational effort of the simulation while ensuring the accuracy of the model, the model preserved the alveolar bone and the right anterior incisor. The main observation target was the change of the parameters of the alveolar bone, and a tetrahedral mesh with 1074 cells was used during the simulation. The initial elastic modulus of the alveolar bone and teeth during the simulation was shown in Figure 7. The nodes at the top of the jaw model were set as full constraints, the rest of the model was set as bound constraints, and the 5N orthodontic force in the distal-mid direction of the labial surface was set for the model. In this work, for safety and comfort reasons, the vibration derived in section 3.2 with the smallest displacement was selected to be applied in the orientation of the tooth surface, i.e., 50Hz and 0.2cm amplitude on the gingival orientation. By obtaining material properties such as density, elastic modulus, and Poisson's ratio to bring into the initial model, the number of properties after the updated iteration was obtained to carry out a new iteration, and so on until the bone tissue environment reached equilibrium. The finite element calculation flow was shown in Figure 7.

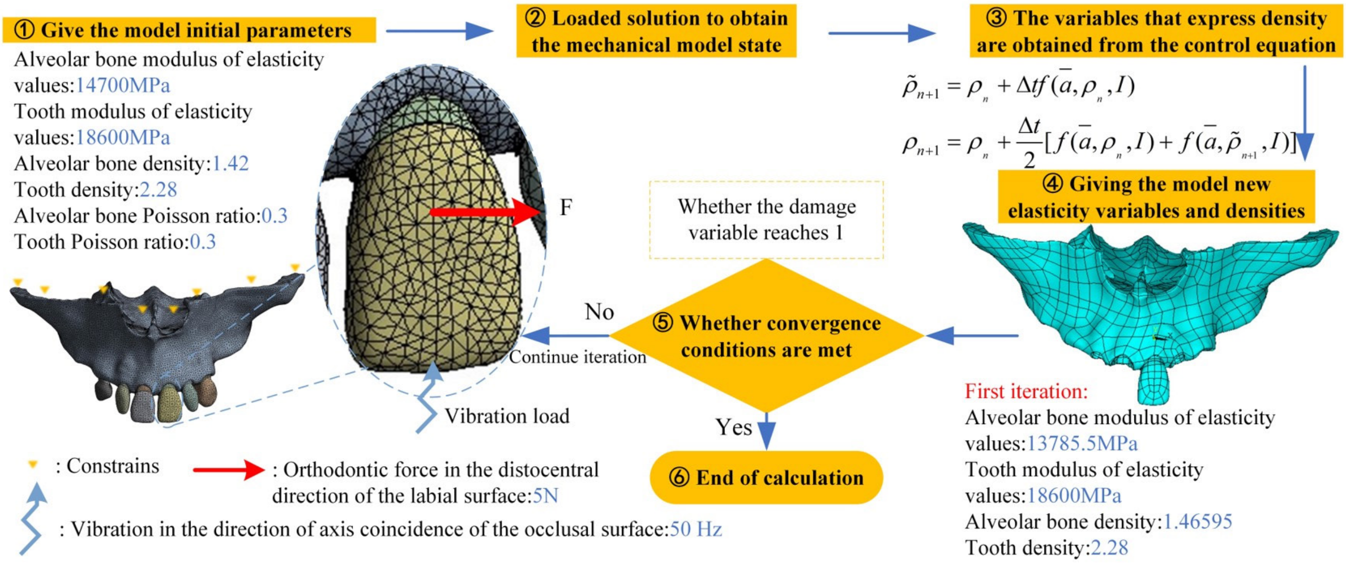

Figure 7 Finite element simulation calculation process.

Patients use the orthodontic acceleration device in daily situations at a frequency of approximately 20 minutes each time (Dubravko P, 2012). So a finite element analysis was performed for 160 minutes (At this point the convergence condition $I=1$ was reached), and the alveolar bone under vibratory loading was subjected to eight iterations, at which time the parameters were shown in Table 3.

Table 3 Parameters of 8-days iteration of alveolar bone under vibration loading.

Parameters / Times	Modulus of elasticity E MPa	Density ρ g/cm3	Rate of density change	Damage variables I
20	14700	1.42	0.028	0
40	13785.5	1.46595	0.057	0.13824
60	13799.6	1.53972	0.075	0.28382
80	13824.7	1.57243	0.047	0.42794
100	13158.7	1.62271	0.044	0.56912
120	12251.6	1.65835	0.040	0.71765
140	11059	1.69398	0.038	0.85588
160	9636.2	0.65282	-1.054	1

4. Discussion

The convergence condition at the end of the simulation was that the damage variable I reached 1, i.e., the alveolar bone tissue was about to reach a state of failure. The fitted curves of the damage variables I with the modulus of elasticity E, density ρ, and rate of density change, respectively, were plotted according to the data in Table 3 as shown in Fig. 8. In this work, based on the model developed for the reconstruction of alveolar bone under vibratory loading, it was known that the damage variable becomes progressively larger with the number of applied vibrations. Based on this theory, a constant frequency vibration load was applied and the trend of elastic modulus and damage variables were analyzed with time to verify the alveolar bone damage simulation said only model with vibration load accelerated orthodontic mechanism and to determine the optimal duration of continuously applied vibration.

As shown in Figure 8a, the change in the elastic modulus of the alveolar bone was divided into three stages when vibratory loads were applied during the orthodontic process. First, the elastic modulus decreases relatively quickly, which indicates that a state of alveolar bone damage begins to develop in the presence of vibration. Then the change in elastic modulus is slightly smoother, which is due to the activation of mechanical regulatory mechanisms within the alveolar bone tissue, which works to stabilize the state of the physiological environment within the alveolar bone, and therefore the change in elastic modulus is not significant. Finally, the damage repair mechanism within the alveolar bone has fallen below the rate of damage accumulation under continuous application of vibratory loads, indicating a decrease in bone tissue performance at this time. When the damage variable I reaches 1, i.e., when the application time is 160 minutes, the modulus of elasticity has already decreased significantly, so the alveolar bone is vulnerable to damage leading to tooth loss at this time. It can be shown that the numerical model can simulate the parameter changes of the real alveolar bone.

As shown in Figure 8c, the rate of change of the bone density of the alveolar bone continues to increase with the increase of the number of vibratory loads applied, which shows the beginning of the alveolar bone reconstruction, corresponding to the bone formation phase (Figure 8b). The rate of bone density reconstruction in the alveolar bone is maximized when the damage repair mechanism within the alveolar bone reaches just the same level as the damage accumulation. Biochemical indicators of bone metabolism can assess the rate of bone reconstruction, and changes in the rate of bone reconstruction are consistent with changes in bone mineral density, and changes in the rate of bone reconstruction are reliable indicators of changes in bone mineral density (Chen DC, 2000). The optimal time to apply vibration at this point is introduced as the time when the damage variable is close to 0.8, with a period of 120 minutes to 140 minutes. The subsequent curve continued to show a decreasing trend, indicating that the level of bone reconstruction and bone damage reached equilibrium, corresponding to a state of bone equilibrium (Figure 8b). The end of the alveolar bone reconstruction process is indicated when the density change rate is 0. To investigate the environmental changes within the bone tissue, the vibratory load continued to be applied at this point while the damage continued to accumulate, and the density change rate gradually decreased below 0, indicating the overload resorption of the alveolar bone tissue, corresponding to the bone resorption stage (Figure 8b).

As shown in Figure 8d, as the number of vibrations applied to the orthodontic teeth increased, the alveolar bone began to rebuild, and therefore its bone density continued to increase. The bone loss due to vibration load injury was supplemented by the alveolar bone reconstruction mechanism, and the curve direction was more consistent with the density change rate, which verified the mechanical regulation mechanism of alveolar bone. However, as the number of vibratory loads is applied, the damage variable becomes larger and the alveolar bone suffers more and more fatigue damage, resulting in a continuous decrease in density. When the damage variable I is 1, the density of the alveolar bone is approximately 0, which indicates that the alveolar bone may have failed locally due to the prolonged application of the vibration load.

The process of bone reconstruction is the result of the coupling effect of both repair damage and damage accumulation within the bone tissue. Under the conditions of applying an orthodontic force of size 5N in the far-medial direction of the labial surface on the right anterior incisor and applying a high-value low-frequency vibration load of 50 Hz and 0.2 cm amplitude on the gingival orientation, the optimal length of applied vibration time is about 120 min-140 min, and the limit of applied time is 160 min when the alveolar bone has failed. By analyzing the curve properties and directions, it is consistent with the vibration load accelerated orthodontic mechanism analyzed in section 2.1 of this work, and it is concluded that the simulated numerical model of alveolar bone damage can express the changes of each parameter property of the alveolar bone under the vibration action.

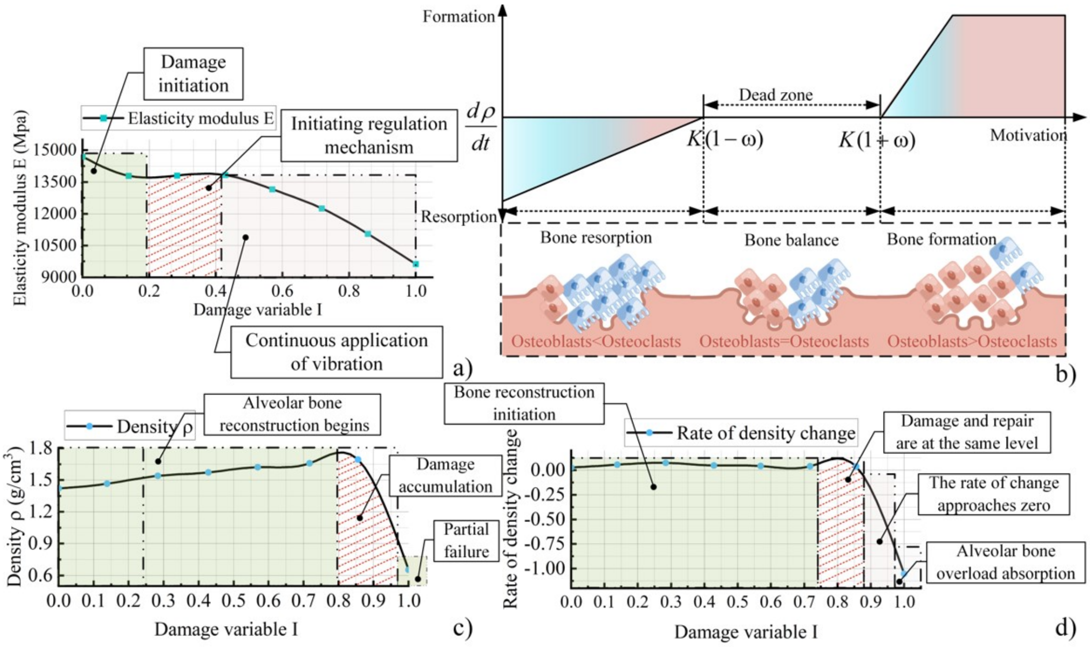

Figure 8 Analysis of simulation results. (a) Relationship between modulus of elasticity and damage variables; (b) Schematic diagram of the dead zone and cell for vibration load application time; (c) Relationship between bone mineral density and injury variables; (d) Relationship between the rate of change of bone mineral density and injury variables.

The simulation method of alveolar bone damage proposed in this paper yielded the distribution of the PDL pressure and tension zones, and the same stress distribution pattern was found in the previous study (Li SX, 2021). It is shown that different material models of PDL (Karimi A,2020) have an effect on the distribution of Von-Mises stress magnitude in PDL, and the stress distribution in the A-zone of PDL derived in this paper is close to the maximum stress of the elasto-plastic model in the previous study (Li SX, 2021), both of which are about 300 Kpa. The orthodontic process is the process of damage and reconstruction of alveolar bone, and this paper demonstrates the mechanism of alveolar bone reconstruction from an engineering point of view, which is consistent with previous studies (Du LL, 2017) showing that the application of cyclic loading reverses bone loss and allows improvement of bone quality in patients with osteoporosis. However, it has been shown that vibration increases tooth movement velocity only in the presence of sustained orthodontic forces (Yamamoto T, 2017), and therefore the magnitude of orthodontic force applied equally affects the effect of vibratory loading on orthodontic acceleration. In order to observe the damage effect on the alveolar bone more obviously, in this paper, a 5N orthodontic force larger than the clinical orthodontic force was selected based on the results of previous studies (Horina JL,2018; Ammar HH,2011), so that the continuous presence of orthodontic force could be simulated. In this paper, we only simulated the effect of orthodontic force on a single tooth, but the analysis showed that the change in the number of osteoclasts in orthodontic acceleration under orthodontic force on a single tooth was consistent with the experimental study in rat teeth (Uchida R, 2017). However, the orthodontic process generally involves multiple teeth subjected to orthodontic forces, and the damage to the alveolar bone during their interaction in the alveolar bone is not known and needs to be studied subsequently. In addition, it has been shown that (Alikhani M, 2018) only the direct application of vibratory load on the target teeth will increase the movement rate of orthodontic teeth, and the indirect application of vibratory load will not enhance the decomposition and metabolism of orthodontic forces, which was also selected to act directly on the target orthodontic teeth in this paper.

5. Conclusions

Existing studies have not determined the mechanism of vibration-accelerated orthodontic treatment and the optimal time of vibration load application. In this work, from the mechanism of alveolar bone reconstruction under vibration, the reconstruction process of alveolar bone under vibration load was expressed as a process of coupling the internal repair and damage accumulation of alveolar bone. Vibratory loads serve to reinforce the mechanical signal in orthodontic acceleration. The fatigue damage of alveolar bone was used as a factor to establish a numerical model of alveolar bone damage under the action of vibration load. Then a three-dimensional model of the maxilla was established from the patient's oral CBCT images, and the validity of the model was verified based on the strain energy density method, with a maximum error of 1.4% from the real data of the alveolar bone. By applying static orthodontic forces and periodic vibratory loads on the teeth, it was analyzed from the point of view of material fatigue damage (an engineering perspective), and it was demonstrated that after applying dynamic and static loads (archwire and vibratory orthodontic acceleration device), the alveolar bone will be subjected to asymmetric cyclic loading, and asymmetric cyclic loading will cause fatigue damage to the alveolar bone. In contrast, fatigue damage (caused by static orthodontic forces versus vibratory loading) acts as a mechanical signal to stimulate alveolar bone reconstruction, thus demonstrating the vibratory loading accelerated orthodontic mechanism proposed in section 2.1. And the displacements of orthodontic teeth under different directions of applied vibration load were determined as Lip orientation > Lingual orientation > gingival orientation. And based on the three-dimensional model and the alveolar bone damage simulation numerical model to simulate the alveolar bone damage process, specifically: the object is the right anterior incisor, loading conditions for 5N orthodontic force, gingival orientation 50Hz, 0.2cm amplitude of vibration load. Based on this, the numerical model of alveolar bone reconstruction mechanism and alveolar bone damage simulation under vibration analyzed in this work was validated, and it was concluded that the numerical model of alveolar bone damage simulation can express the changes of each parameter property of alveolar bone under vibration (the curves of bone density, elastic modulus, and bone density change rate of alveolar bone from 0 to 1 for the damage variables). By analyzing the rate of bone density reconstruction and damage accumulation in the alveolar bone, the analysis resulted in an optimal application duration of 120 to 140 minutes (with a damage variable of about 0.8, corresponding to the maximum bone density and the maximum rate of bone reconstruction change) and limit duration of 160 minutes (when the damage variable I reaches 1). To illustrate the validity of the results of this study, the geometric model was first verified according to the strain energy density method, with a maximum error of 1.4% with the real data of the alveolar bone. The distribution of the pressure and tension zones of the PDL obtained was consistent with previous studies (Li SX, 2021). The change in the number of osteoclasts and osteoblasts in orthodontics derived from the analysis of orthodontic acceleration mechanism was consistent with the trend of the rat test (Alikhani M, 2018), which provided a basis for simulating the damage and reconstruction of alveolar bone under vibration loading. The effect of vibration-accelerated orthodontics in alveolar bone is consistent with the findings of BMP-induced ectopic bone formation and BMP 2-induced ectopic bone formation and rib fracture healing models in rodents (Mahy PR, 1988; Takaoka K, 1988; Yoshikawa H, 2009; Hashimoto J, 1989). It can be shown that the numerical simulation model can also be applied to other parts of the bone reconstruction theory, but the specific microscopic parameters need to be modified to adapt to the response mechanisms of different bones (Uchida R, 2017) under vibratory loads.

The time of action of the vibration load can be indicated to the physician and the patient. According to the conclusions drawn in this paper, a guiding role can be provided for orthodontic acceleration in the clinic. However, the movement of orthodontic teeth, due to its biological nature, should be subsequently explored in depth in conjunction with clinical or biological experiments to validate microscopic specific parameters in combination with macroscopic expressions.

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