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Biomechanical and histomorphometric evaluation of biodegradable mini-implants for orthodontic anchorage in the mandible of beagle dogs

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Abstract

Objective To evaluate the effectiveness of a mini-implant composed of unsintered hydroxyapatite, poly (L-lactic acid) and poly(lactic-co-glycolic acid) (u-HA/PLLA/PLGA) composites as an anchorage device under consistent orthodontic force (OF) loading in vivo.

Methods An mandible model in beagle dogs was introduced. 144 mini-implants were implanted in both sides of the mandibles. The mini-implants in the experimental group (left side) were loaded at the magnitude of 200 g to simulate the OF. At 2, 4 and 6 months after implantation, tissue specimens were taken from the implanted sites and biomechanical, histological and histomorphometrical analysis were performed.

Results Mini-implants in the group with the highest PLLA ratio showed a 27% non-fracture rate after 4 months and 20.83% after 6 months in beagle dogs, and the non-fractured mini-implants could maintain the tensile force of 200 g, while mini-implants in the other two groups were all fractured. Histomorphological analysis showed that there was no significant relationship between Bone Volume over Total Volume (BV/TV) and the implantation time among the most of the groups. The level of Bone-Implant Contact ratio (BIC) in Medium and Low ratio group were decreased gradually from 2 to 6 months.

Conclusions This study showed the biodegradable mini-implant could work as an alternative to the titanium alloy mini-implant by adjusting the proportion of its ingredients.

Clinical relevance Degradable mini-implants for orthodontic anchorage lie in their potential to revolutionize orthodontic treatments by offering a biodegradable alternative that minimizes the need for secondary surgeries for removal, thereby enhancing patient comfort and reducing overall treatment time.

Keywords Anchorage, Biodegradable composite, Orthodontic mini-implant, Poly(L-lactic acid)

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Background

The success of orthodontic treatment is closely related to the stability of anchorage. Generally, a set of several teeth is used as anchorage to minimize their unfavorable displacements during orthodontic treatment. However, traditional approaches may lead to the loss of anchorage in some cases which will compromise final results, especially in patients with severe occlusal discrepancies or multiple absent posterior teeth [1]. The introduction of titanium mini-implants has revolutionized orthodontics practice, enabling more precise control of tooth movement, minimizing or even eliminating undesirable tooth movement. Additionally, the application of mini-implants has increased the envelope of orthodontic treatment, providing an alternative to orthognathic surgery and allowing asymmetric tooth movement in three dimensions to resolve the challenging malocclusion [2, 3]. Nowadays, titanium alloy is widely used as mini-implant material due to its suitable mechanical properties and high biocompatibility. However, a secondary operation is required to remove the mini-implants at the end of orthodontic treatment. Additionally, there are concerns regarding the potential toxicity from metal ion released into bodily fluids [4]. Therefore, biodegradable materials have been investigated as alternatives to non-degradable materials as they can gradually degrade in vivo and non-toxic degradation products can be safely absorbed or excreted by surrounding tissue [5].

An ideal absorbable anchorage device for orthodontics should exhibit robust strength and an adequate modulus, maintain its strength throughout the necessary for tooth movement, and be absorbed without casuing any reactions that might impede bone healing [6]. Recently, bioabsorbable devices made by varieties of synthetic polymers such as poly(L-lactide) (PLLA) copolymers have been developed [7], and have been used increasingly in orthopedic surgery for operations such as fractured bone fixation [8, 9], reconstruction of the cruciate ligaments of the knee [10], and bone graft fixation [11]. A previous study showed that the PLLA mini-implant had favorable biocompatibility and mechanical strength, which successfully achieved mandibular molar distalization in 6 months without any sign of failure and inflammation in adjacent tissue [12]. However, PLLA's high crystallinity and hydrophobic nature present challenge, such as a considerably long degradation period (2–6 years), weak bone bonding capacity, and potential for immunological rejection [13, 14].

To address these drawbacks, poly(lactide-*co*-glycolide) (PLGA) with PLLA have attracted wide attention for their better biocompatibility, biodegradability and mechanical strengths [15, 16]. PLGA and PLLA, despite being insoluble in water, are both hydrolytically unstable. They undergo degradation through hydrolytic attack of their ester bonds, leading to the formation of lactic and glycolic acids. Conventionally, the rate of hydrolytic degradation of these biopolymers is modulated by modifying their physical properties; such as their molecular weights, degree of crystallinity and glass transition temperature (T_g) [17–19]. In addition, inorganic fillers, such as hydroxyapatite (HA) were introduced into biodegradable polymers to improve their mechanical properties and biocompatibility [20]. In this case, it is necessary to investigate the efficacy of the u-HA/PLLA/PLGA as a novel material for mini-implant anchorage application.

Therefore, the aims of this study was to evaluate the effectiveness of a mini-implant composed of u-HA/PLLA/PLGA composites as an anchorage device under consistent orthodontic force (OF) loading in vivo.

Methods

Animals

Twelve adult beagles (male, 13–15 kg, 12–15 months old) from Nanjing Chaimen Biotechnology Co., Ltd. were used in this study and kept at the animal care facility for 2 weeks before the surgical procedures. During this period, they were provided with a soft diet and accommodated in separate housing. The experimental protocols employed in this study were reviewed and approved by the Institutional Animal Care and Use Committee of GUANG DONG HUA WEI TESTING CO., LTD. (202301002).

Biodegradable mini-implants

The biodegradable mini-implants were designed with a square head, outer diameter of 2.0 mm, core diameter of 1.6 mm, thread pitch of 1.0 mm and length of 11.0 mm (Fig. 1A). Medical grade polymer materials, PLGA (75/25) and PLLA were acquired from Sino Biomaterials (Changchun, China), and their average molecular weight were 18000 and 89000 Daltons respectively. The ratio of PLLA to PLGA in the mini-implants was set to 2:1 (H, high), 1:1 (M, medium) or 1:2 (L, low) to adjust the degradation rate and mechanical strength. Uncalcined HA (u-HA) particles with the size of 100 nm (length) × 30 nm (width) (Emperornano material, Nanjing, China) accounted for 10 wt% in u-HA/PLLA/PLGA composites (Table 1). The mixture of u-HA, PLLA and PLGA was processed into mini-implants using an injection moulding machine (Haitian, Ningbo, China) within the Sino Biomaterials GMP plant under ISO13485 standards.

Mini-implant placement

Beagles were assigned into 3 groups randomly associated with the 3 different loading period (2 months, 4 months and 6 months, respectively). All surgical procedures were conducted in sterile conditions. The animals

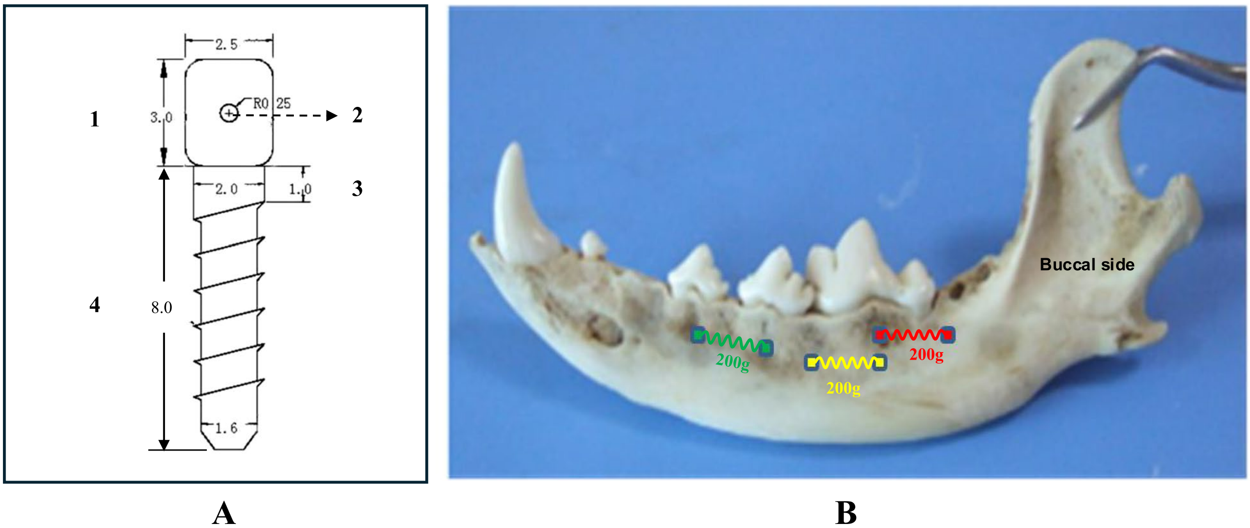

Fig. 1 **A** Parameter of designed mini-implant.1: Head of Mini-implant; 2: Round hole for linking tension spring; 3: Neck of Mini-implant; 4: Mini-implant body. **B** Sites of mini-implant. Mini-implant sites are indicated by red, yellow and green squircle. On the left sides, approximately 200 g of constant force was loaded immediately by activating the super elastic Ni-Ti closed coil springs between the paired mini-implants on 36 pairs of mini-implants. Another 36 pairs of mini-implants on the right sides without coil springs were regarded as control groups

Table 1 Mini-implant grouping by the ratio of the materials (wt%)

Description	Numbers of implants (n)	PLLA	PLGA	u-HA
H (High)	48	60	30	10
M (Medium)	48	45	45	10
L (Low)	48	30	60	10

were pre-anesthetized by a 0.044 mg/kg intramuscular injection of atropine sulphate (Northeast Pharmaceutical Group Corp., Shenyang, China) followed by an intramuscular injection of ketamine hydrochloride (16 mg/kg) and xylazine hydrochloride (8 mg/kg) (North China Pharmaceutical Group Corp., Shijiazhuang, China) for full anesthesia. Totally 144 mini-implants were used in this study. A split-mouth design was adopted in this study, designating the left side of the mandible as the experimental group (with force loading) and right side as control group (without force loading). Two pairs from each mini-implant type were randomly inserted into the anterior, middle and posterior regions of the both sides of mandible, and three types of implants were equally distributed across different regions. The mini-implants on the experimental side (left side) were subjected to a magnitude of 200 g force with a Nickl-titanium tension spring (SHENZHEN SUPERLINE TECHNOLOGY CO., LTD, Shenzhen, China) between the paired mini-implants to simulate the orthodontic force (OF) (Fig. 1B). This spring is designed to consistently provide a stable traction force of 200 g over an extended period.

Specimen preparation

Following 2, 4, and 6 months of force loading, the animals were humanely euthanized through an overdose of anesthesia. Subsequently, the mandibles were meticulously excised and segmented into smaller blocks, ensuring each block contained one mini-implant encased by a minimum of 2 mm of surrounding bone. All bone/mini-implant specimens were then fixed in 10% formaldehyde for 48 hours at ambient temperature.

Biomechanical test

The tensile strength of the mini-implants were evaluated by using a universal testing machine (SHIMADZU, Japan). A total of 54 blocks (emcompassing each type and time point from both experimental and control sides) were trimmed to make the longitudinal axis of mini-implants parallel to the testing surface (Supplementary figure 1). The applied load was monitored, and the peak load at breaking (Fmax) recorded from the data file. Once the head of mini-implant fracture prior to the mechanical test, then tensile strength for that particular mini-implant was documented as zero.

SEM (Scanning Electron Microscope) imaging of mini-implants

The mini-implants under went a post-removal processing protocol. Initially, they were rinsed with Phosphate Buffered Saline (PBS; Sigma Aldrich, St. Louis, MO, USA), subsequently dehydrated through successive graded ethanol baths and chemically dried with hexamethyldisilazane

(Sigma Aldrich, St. Louis, MO, USA). The mini-implants were left to dry on the sample holder and then coated with gold prior to imaging in SEM (S- 3400 N; Hitachi, Japan).

Histological analysis

The Bone blocks were bisected along the mesio-distal axis parallel to the load vector, and were washed in PBS and decalcified in formic acid (Sigma Aldrich, St. Louis, MO, USA) followed by EDTA (Sigma Aldrich, St. Louis, MO, USA) for durations of 72 hours and 28 days, respectively. Post-decalcification, bone specimens were then embedded in wax after dehydration in a graded series of ethanol, and 5-um thick slices along the mini-implant's long axis were obtained. The specimens were stained with hematoxylin and eosin (H&E) (abcam, Cambridge, UK) for detailed histological examination via microscopic.

Histomorphometric analysis

Undecalcified bone blocks underwent systematic dehydration process using a series of ethanol of increasing concentration, followed by a gradual immersion in styrene before being embedded in polyester resin. Subsequently, the embedded blocks with implants and bone were sectioned sagittally along the force vector plane into 200um thick slices with a low-speed diamond saw (Isomet 2000; Buehler, LakeBluff, IL) and then refined down to a thickness of 70 um with a grinding/polishing machine (Metaserv 3000; Buehler) under a continuous water irrigation system. For histological analysis, the sections were stained with toluidine blue and referred to optical microscopy evaluation. To obtain a comprehensive image of the entire longitudinal section of the mini-implant interface, six microscopic images at 40x magnification were digitally combined. The combined mini-implant images were quantitatively analyzed by using BioQuant Osteoversion 7.20.10 (BIOQUANT Image Analysis Corporation, Nashville, USA) to calculate bone volume fraction which defined as Bone Volume over Total Volume (BV/TV), and Bone-Implant Contact ratio (BIC).

Statistical analysis

Paired *t* test was used to evaluate the effect of mechanical loading on breaking strength values. For BV/TV and BIC values, the mean and standard error of mean (SE) were calculated using SPSS 12.0 (Lead Tech, Chicago, IL, USA). The Kruskal-Wallis H test was used to compare BIC and BV/TV for the different loading periods (2, 4, or 6 months) and loading methods.

Results

Macroscopic inspection of degradable mini-implants

There were no discernible differences in color or texture between the force loaded and unloaded mini-implants across all three groups. However, it was observed that the heads of some implants were damaged or had become dislodged. The number of intact mini-implants within each group was shown in Table 2.

Biomechanical test

Statistical analysis revealed different in tensile strength among the three different types of mini-implants at all measured time points, except for six-month. There was a statistically significant gradual decline in the tensile strength of the mini-implants, which can be attributed to the degradation of the polymers, as illustrated in Fig. 2.

SEM images of mini-implants

Initially, the surfaces of all mini-implants were uniformly smooth, looking similar to each other prior to placement. After 2 months, the surfaces of all mini-implants retained their smoothness, although some shallow cracks could be observed on the surface of mini-implants with low PLLA ratio and medium PLLA ratio compared with group of high PLLA ratio. By the 4 months, the mini-implant with a high PLLA ratio started to show a few surface cracks. For mini-implants with a medium PLLA ratio, these cracks had expanded, merging to form significant grooves. While, the mini-implants with a low PLLA ratio exhibited deep pits and extensive surface roughness. At 6 months, all three types of mini-implants displayed an obvious increase in surface roughness compared with that of 4 months due to progressive degradation (Fig. 3).

Histological analysis

The findings from the histological sections showed an absence of inflammatory cells infiltration and foreign body granulomatous reactions at any sites. Spongy bone consists of trabeculae with osteocytes within the bone matrix and osteoblasts lining the surface. Since

Table 2 The number of unfractured mini-implants in each group

Loading Time (Month)	Loading Methods	Description		
		H	M	L
2	Loading	8	6	0
	Un-loading	8	7	0
4	Loading	6	5	0
	Un-loading	7	6	0
6	Loading	5	0	0
	Un-loading	5	0	0

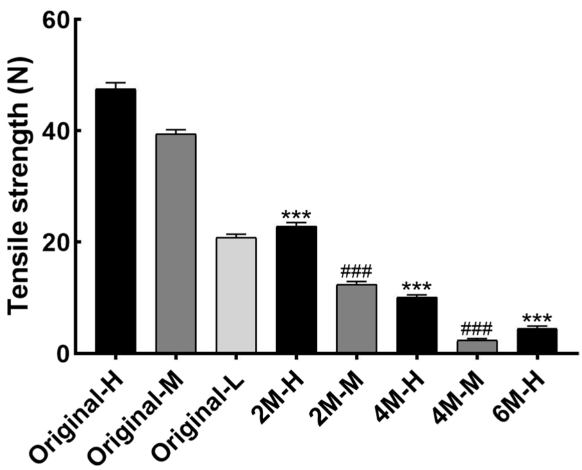

Fig. 2 The tensile strength of mini-implant with high PLLA ratio (H) in black, medium PLLA ratio (M) in gray and low PLLA ratio (L) in light gray. The tensile strength of mini-implant with high and medium PLLA ratio had almost the same tendency of decrease, and dropped to about 5 N and 0 N at the end of the study respectively. The mini-implant with low PLLA ratio lost its tensile strength after 2 month of placement. * there is significantly statistic difference between the group with original-H group. # there is significantly statistic difference between the group with original-M group

the osteocytes shrink caused by fixation, the lacunae were visible. Evidence of bone formation at trabecular regions was observed. Undifferentiated connective tissue containing osteoprogenitor cells were present within the spaces between the trabeculae. The majority of mini-implant surfaces were surrounded by undifferentiated connective tissue (Fig. 4A-C). A remarkable feature found in some experimental groups was the apparent cortical bone loss along with soft-tissue apical migration at the side opposite to the static load vector direction (Fig. 4D).

Histomorphometric analysis

In the toluidine blue-stained longitudinal sections, the comprehensive visualization of the mini-implant interface facilitated the assessment of BIC and the peri-implant BV/TV across all 54 implants, as depicted in Fig. 4E-F.

Comparison of BIC and BV/TV with respect to the different force loading period after mini-implants implantation

Comparison were made of tissue samples subjected to 2, 4, or 6 months after the mini-implants implantation. In the histomorephometric analysis, significant differences were observed in BIC and BV/TV depending on the time (Table 3). There was a trend of increasing BV/TV from 2 months to 6 months across all three groups, a statistically significant difference was only observed in the H group ($p<.05$) between the 2 month and 6 month group. In contrast, the histomorephometric analysis showed a significant decrease in BIC from 2 months to 6 months in M and L group while an increase between 2 and 4 months in H group was observed ($p<.05$). At 2 months, the BV/TV of L group was 17.98%, which was significantly higher than H group. At 4 months, L group was 20.37%, which was significantly higher than H group and M group ($p<.05$). The mean of BIC as measured via histology showed a significant difference among each ratio group ($p< 0.05$).

Comparison of subgroups for each loaded and unloaded mini-implants

The mean BV/TV values, as determined through histomorphometric analysis, exhibited significant differences in the H-group observed over a 4-month period and 6-month period between the loading and without loading group. The values were 15.43% with loading and 18.89% without loading in the 4-month period. Similarly, after 6 months, the H-group demonstrated values of 17.47% with loading and 19.92% without loading (Table 4). In contrast, the mean Bone-to-Implant Contact (BIC) values, as assessed via histological analysis, did not show significant differences across the groups, as detailed in Table 5.

Discussion

The main advantage of the degradable mini-implant used in the present study is to avoid the secondary removal of implant without imaging interference [21], although its mechanical properties and adverse tissue reactions during the degradation process are the issues which we are worrying about. With the development of materials science, however, the application of these degradable fixtures in the craniomaxillofacial position has become more and more frequently studied [4]. In this study, we investigated the mechanical properties and histomorphology of a novel degradable mini-implant made of u-HA/PLLA/PLGA composite material.

The clinical applicability of our study results is directly related to the similarity between canine skeleton and human skeleton. According to the studies of Melsen and Parfitt, the bone formation in canines is about twice as fast as that in human beings [22], and the bone remodeling in canines is approximately 42% faster than that in human beings [23]. Except for other primates, canine skeleton is most similar to human skeleton [24].

The study results indicated that mini-implants in the group with the highest PLLA content showed a 20.83% non-fracture rate after 4–6 months in beagles, and the non-fractured mini-implants could maintain the tensile force of 200 g, while mini-implants in the other two groups were all fractured. It is mainly due to the

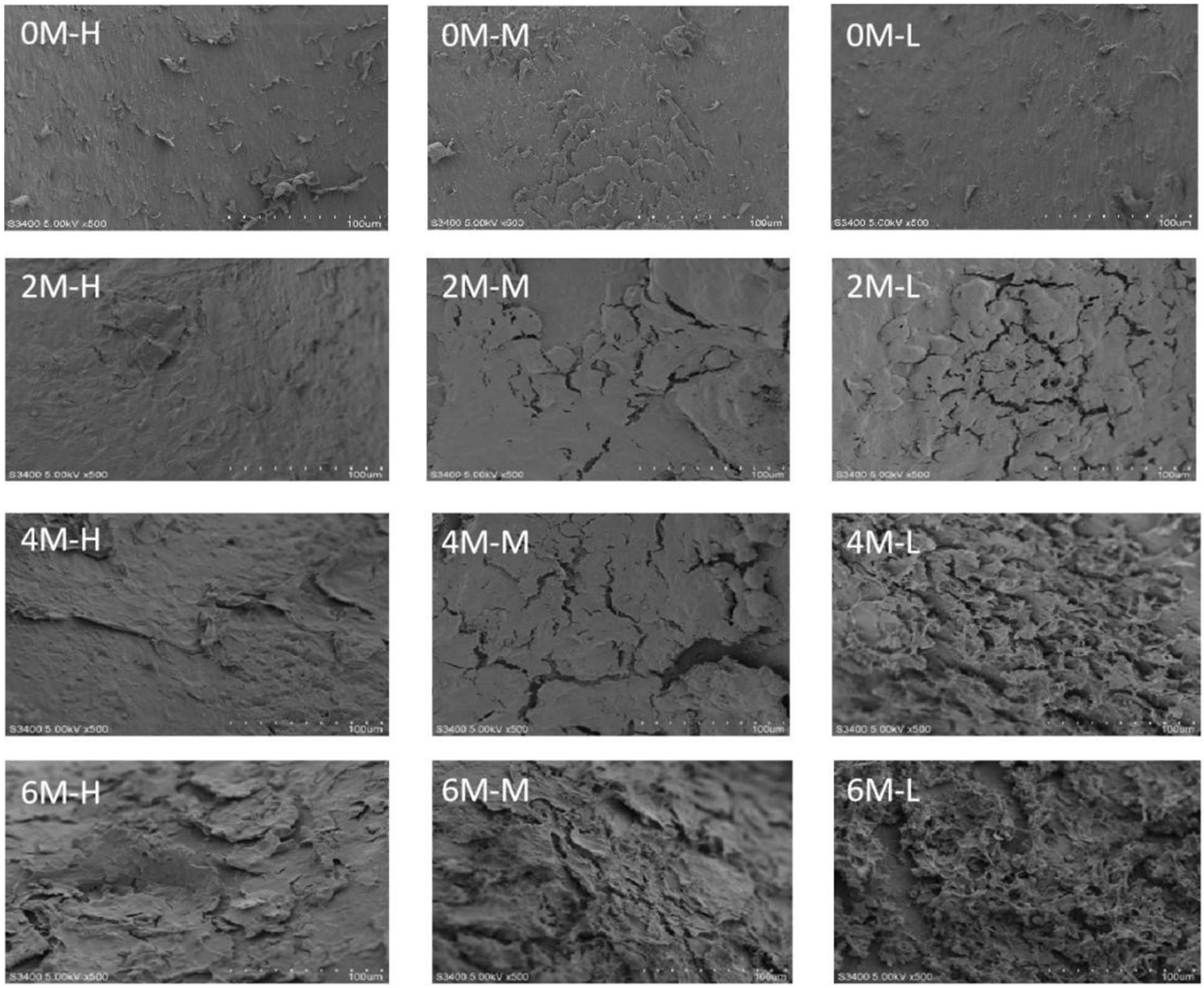

Fig. 3 Surface morphology of the mini-implants under scanning electron microscopy (SEM). H, high ratio of PLLA to PLGA(2:1); M, medium ratio of PLLA to PLGA (1:1); L, low ratio of PLLA to PLGA (1:2)

relatively slow degradation of PLLA in vivo, which can maintain the strength of mini-implants [25]. The material used in this study was made by the physical mixing of three components. The uneven mixing of these three components may lead to some structural defects and affect its overall strength, which can explain the 80% fracture rate in the group with the highest PLLA content. In addition, the nano-sized HA has poor dispersibility in PLA matrix and poor interfacial adhesion to other components [26, 27], and there may be a phenomenon of serious HA aggregation, thus affecting the mechanical properties of the composite material [28] In general, mini-implants can bear a load of no more than 2 N and work in patients for 8–12 months, so the strength of this high-content PLLA material is sufficient to be used for mini-implant. At present, some studies have demonstrated the safety and mechanical adequacy of biodegradable devices in low-load applications, such as the reconstruction of craniofacial bones, maxilla and mandible [29–31]. Furthermore, the effects of load and non-load on mini-implants strength were also compared in the present study, and no significant difference was found, which is consistent with other study results. As shown by the studies of Farrar and Deng, the molecular weight and component distribution of degradable polylactic acid material is the key factor affecting its strength, and the load cannot cause the remarkable changes in its molecular weight [32, 33]. Although the degradable mini-implant has poorer mechanical strength compared with the metal mini-implant [34, 35]. the composite material has obvious advantages in terms of biodegradability, bone conductivity and bone substitution potential [35], and it does not require secondary surgical removal. Therefore, such

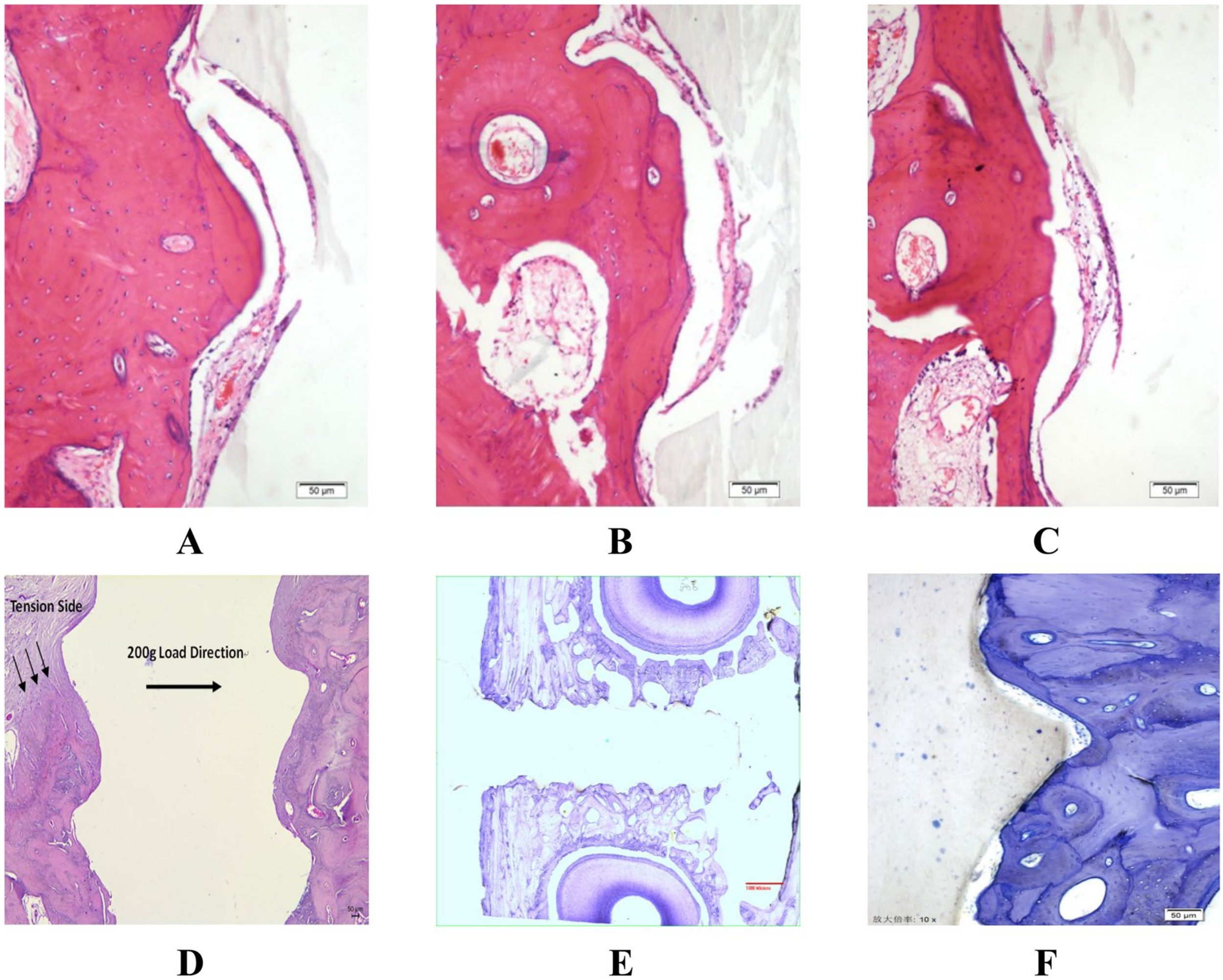

Fig. 4 Example of histological images of HE staining from different time points. **A** 2 months after implantation(High PLLA ratio). **B** 4 months after implantation(High PLLA ratio). **C** 6 months after implantation(High PLLA ratio). **D** Apparent cortical bone loss with soft-tissue apical migration at side opposite static load vector direction (tension side) was observed for all experimental groups. **E** Histology image of the entire longitudinal sectional mini-implant interface. **F** The enlarged view illustrates adequate osseointegration. The newly formed bone is in direct contact with the mini-implant surface. Bone quality shows lamellar bone

Table 3 BV/TV and BIC percentages of different types of mini-implants with respect to different time points

Time (Month)	N	BV/TV(%)						BIC(%)					
		High		Medium		Low		High		Medium		Low	
		Mean	SE	Mean	SE	Mean	SE	Mean	SE	Mean	SE	Mean	SE
2	18	16.14[b†]	0.75	17.40	2.31	17.98	1.73	39.97[ab*†]	1.00	41.78 [ab†]	1.22	22.75[ab]	3.99
4	18	17.16[†]	2.07	17.10[†]	1.81	20.37	2.52	48.10[*†]	7.76	30.16[b†]	2.99	18.12	1.49
6	18	18.69	1.38	18.50	6.46	21.19	6.46	45.78[*†]	1.72	24.41	6.46	17.36	1.47

[a] Significant difference with 4-month group, $P < 0.05$

[b] Significant difference with 6-month group, $P < 0.05$

[*] Significant difference with M group, $P < 0.05$

[†] Significant difference with L group, $P < 0.05$

Table 4 BV/TV percentages of different types of mini-implants with respect to different duration of force loading

	Methods	N	2M		4M		6M	
			Mean	SE	Mean	SE	Mean	SE
High	Loading	9	15.81	0.87	15.43 [a]	1.19	17.47 [a]	0.50
	Unloading	9	16.46	0.58	18.89	0.57	19.92	0.10
Medium	Loading	9	17.57	3.60	17.20	0.73	17.26	7.55
	Unloading	9	17.24	0.55	17.00	2.77	19.75	6.54
Low	Loading	9	16.97	0.73	18.49	1.18	23.48	0.90
	Unloading	9	19.00	1.98	22.26	1.94	18.89	9.37

[a] Significant difference with unloading group, $P < 0.05$.

Table 5 Bone-Implant Contact (BIC) percentages of different types of mini-implants with respect to different duration of force loading

	Methods	N	2M		4M		6M	
			Mean	SE	Mean	SE	Mean	SE
High	Loading	9	39.58	0.82	44.25	0.76	45.51	1.43
	Unloading	9	40.37	1.16	43.64	3.47	46.04	2.26
Medium	Loading	9	41.69	1.13	28.93	1.55	22.55	0.99
	Unloading	9	41.88	1.55	31.40	3.92	26.27	4.53
Low	Loading	9	23.80	1.81	18.28	0.93	18.46	1.02
	Unloading	9	21.70	5.75	17.97	2.14	16.26	0.86

[a] Significant difference with unloading group, $P < 0.05$

composite material will still be the hotspot in the future research.

Through the SEM observation of mini-implant surface, we found that the cracks gradually appeared on the surface of mini-implant as time went on, and the most cracks were observed in the group with a low PLLA content. However, no attachment of surrounding tissues was observed on the surface of mini-implants, and no inward growth of tissues was found in the cracks. As revealed by the previous studies, the initial stage of polylactic acid degradation is the cleavage of polymer chains in the amorphous region, which produces more chains with a lower molecular weight. Since occurring inside the polymers, these changes cannot be detected in the surrounding tissues, and thus it is impossible to observe the attachment of surrounding tissues in the cracks. Furthermore, with the further degradation of polylactic acid, the shortened polymer chains in the amorphous region are small enough to diffuse out from the polymers, thus causing the appearance of macrophages on the implant surface. Then, these cells absorb the fine particle products from the hydrolyzation of polylactic acid to gradually reduce the volume of the material [25]. Therefore, it can be seen that the polymer in the group with a low-content PLLA has a fast internal cleavage speed, manifested as a large number of cracks on the surface, which is also consistent with the rapid attenuation of its mechanical strength. A longer in vivo degradation time may be needed to observe the absorption or even complete absorption of the material.

The results of histological observation showed no significant difference in the morphological structure of bone tissues around mini-implants over time, which is similar to the titanium alloy micro-implant widely used in the clinical practice [36]. Some scholars also conducted several studies similar to our study, and they found that there was no significant difference of local tissues in 4 years after the implantation of u-HA/PLLA implant in experimental animals, and no inflammatory response was observed [37]. Although it has been reported that the acidic products from the degradation of polylactic acid material may cause local inflammatory response [38, 39], this phenomenon was not observed in the present study, which may be related to the low degradation of composite materials [40]. If they are further degraded into the fine particle products by the hydrolyzation of polylactic acid, macrophage phagocytosis will occur, which may lead to inflammatory response [35]. In this study, the relevant in-depth discussion was not made due to the time limitation, and we can perform the corresponding exploration in the future research. Moreover, there was no significant difference in histology between the loading group and the non-loading group, which is consistent with other studies [41]. We believe that the static load have no obvious effect on the tissues around mini-implants. Meanwhile, we also observed a phenomenon similar to that in the metal mini-implants, i.e., the common feature of all the load groups was that there were apparent resorption of cortical bone and apical migration of soft tissues on the opposite side of the static load vector. Some studies have shown that this may be due to the increased distance between the surface of mini-implant and the bone caused by adding a static load, and resultantly the migration of soft tissues in the early stage of mini-implants load activation [42].

The stability of mini-implant is the key for the success of orthodontic treatment. Mini-implants stability can be divided into initial stability and long-term stability. The initial stability is the immediate stability after the implantation of mini-implants in bone and depends on the mechanical bonding between mini-implant and bone tissues, while the long-term stability depends on the integration of mini-implant and the surrounding bone tissues [43]. Since the earliest observation in this study was the osseointegration 2 months after implantation, it was difficult to evaluate the initial stability; however, the long-term stability of degradable mini-implant was reflected by the observation of new bone formation and bone bonding around mini-implant. The histomorphological analysis showed that there was no significant relationship between BV/TV and the implantation time. BV/TV tended to be stable, and the value of BV/TV was much lower than that of titanium alloy mini-implant reported in the previous study [38]. The reason may be that the measurement began at the second month in our study, during which the initial healing of the bone surrounding mini-implant was completed and the new bone became stable. Besides, our study results have indicated that the degradable materials HA and PLLA may cause local allergy or inflammatory response [38–40], thereby affecting the formation of new bone, which may also be the reason why the BV/TV of mini-implants in this study is smaller than that of titanium alloy implant. However, there was no significant inflammatory response found by histological observation, which may be related to the slow degradation of the material [40]. We found that the level of BIC was decreased gradually from 2–6 months, and the fastest decrease of BIC level was observed in L group. Our finding was different from the study results of titanium alloy implant showing the level BIC was increased gradually over time [3, 44]. The results of the present study have suggested that the degradation rate of materials can affect the level of BIC, i.e., the cracks on the surface can destroy the original bonding between the material and the implant with the continuous degradation of the material; however, the surrounding tissues cannot immediately fill in these cracks, which was clearly observed in the SEM results of our study and eventually led to a gradual decrease of BIC level. This is consistent with the study results on the degradation of magnesium alloy implant: with the degradation of magnesium alloy, the BIC level of the implant with corrosion-resistant coating was significantly increased as compared with that of the implant without coating, and the BIC level and BV/TV were both reduced due to the surface loosening and hydrogen evolution after the degradation of magnesium alloy [39]. Our study results have demonstrated that the immediate loading of low-intensity static load cannot significantly affect the osseointegration of the degradable mini-implant. It is consistent with the conclusions of the studies on metal implants that the inflammation of soft tissues around the implant, other than the immediate loading of orthodontic static load, is a key factor influencing the osseointegration [45].

The study has several limitations that should be addressed in future research. Firstly, the relatively short follow-up period of six months may not adequately reflect the long-term behavior of the biodegradable implants. While the initial results are promising, they necessitate further validation through extended studies. Additionally, the research could benefit from a larger sample size and the inclusion of more diverse animal models to enhance the robustness of data concerning the implants' performance and biocompatibility. Besides, different locations of implants should also be taken into consideration. Future research should also focus on optimizing the composition of these biodegradable mini-implants, particularly by exploring different ratios of PLLA and PLGA to achieve an optimal balance between mechanical strength and degradation rates. Ultimately, clinical trials in humans, are essential to fully ascertain the feasibility and safety of these implants for routine orthodontic use.

In conclusion, the degradable mini-implant used in the present study also needs to be further improved in terms of mechanical properties and biocompatibility, but the advantages of this composite material in biodegradability and bone substitution potential can bring new insight into the mini-implant application in the clinical senarios. We believe that with the continuous development of materials science and the improvement of surface denaturation and modification technologies, the performance of new degradable materials will be greatly enhanced, and these materials will be widely applied in the field of orthodontics.

Abbreviations

OF	Orthodontic force
PLLA	Poly(L-lactide)

PLGA	Poly (lactide-*co*-glycolide)
HA	Hydroxyapatite
PBS	Phosphate Buffered Saline
SEM	Scanning Electron Microscope
BV/TV	Bone volume over total volume
BIC	Bone-implant contact ratio

Supplementary Information

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Supplementary Material 1.

Acknowledgements
Not applicable.

Authors' contributions
SW carried out the data collection of the study, performed the measurements and drafted the manuscript. ZZ gave statistical analysis guidance, did formal analysis and revised the manuscript. QT participated in the investigation and revised the manuscript. LC gave the study concept and design and supervised this study. All authors gave final approval and agreed to be accountable for all aspects of the work. All authors read and approved the final manuscript.

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Data availability
The datasets used and/or analysed during the current study are available from the corresponding author on reasonable request.

Declarations

Ethics approval and consent to participate
The animals used in this study were obtained from Nanjing Chaimen Biotechnology Co., Ltd. Informed consent for their use was obtained from the company, ensuring full compliance with ethical guidelines. We followed the Basel Declaration and the International Council for Laboratory Animal Science (ICLAS) ethical guidelines and relevant guidelines of Directive 2010/63/EU for animal care and use. The animal experiment was conducted according to ARRIVE guidelines, and the experimental protocols were approved by the Institutional Animal Care and Use Committee of GUANG DONG HUA WEI TESTING CO., LTD. (202301002).

Consent for publication
Not applicable.

Competing interests
The authors declare no competing interests.

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References
1. Proffit WR, Fields HW. The biological basis of orthodontic therapy. In: Contemporary orthodontics. 2nd ed. St. Louis: Mosby-Year Book; 2000. p. 296–352.
2. Bucur SM, Vaida LL, Olteanu CD, Checchi V. A brief review on micro-implants and their use in orthodontics and dentofacial orthopaedics. Appl Sci (Switzerland). 2021;11(22):10719.
3. Choi SH, Jeon JY, Lee KJ, Hwang CJ. Clinical applications of miniscrews that broaden the scope of non-surgical orthodontic treatment. Orthod Craniofac Res. 2021;24:48–58.
4. Ashammakhi N, Peltoniemi H, Waris E, et al. Developments in cranio-maxillofacial surgery: use of self-reinforced bioabsorbable osteofixation devices. Plast Reconstr Surg. 2001;108:167–80.
5. Tipan N, Pandey A, Mishra P. Selection and preparation strategies of Mg-alloys and other biodegradable materials for orthopaedic applications: a review. Mater Today Commun. 2022;31:3083–94.
6. Pietrzak WS, Sarver DR, Verstynen ML. Bioabsorbable polymer science for the practicing surgeon. J Craniofac Surg. 1997;8:87–91.
7. Lu Z, Zhang B, Gong H, Li J. Fabrication of hierarchical porous poly (l-lactide) (PLLA) fibrous membrane by electrospinning. Polymer (Guildf). 2021;226:123797.
8. Podgorbunsky AB, Imshinetskiy IM, Mashtalyar DV, et al. Bioresorbable composites based on magnesium and hydroxyapatite for use in bone tissue engineering: focus on controlling and minimizing corrosion activity. Ceram Int. 2025;51:423–36.
9. Joshi A, Dias GJ, Staiger MP. Surgically-induced deformation in biodegradable orthopaedic implant devices. Acta Biomater. 2022;154:667–75.
10. Ficek K, Rajca J, Stolarz M, et al. Bioresorbable stent in anterior cruciate ligament reconstruction. Polymers (Basel). 2019;11(12):1961.
11. On SW, Cho SW, Byun SH, Yang BE. Bioabsorbable osteofixation materials for maxillofacial bone surgery: a review on polymers and magnesium-based materials. Biomedicines. 2020;8(9):300.
12. Aoki T, Ogawa K, Miyazawa K, et al. The use of bioabsorbable implants as orthodontic anchorage in dogs. Dent Mater J. 2005;24:628–35.
13. Zaaba NF, Jaafar M. A review on degradation mechanisms of polylactic acid: hydrolytic, photodegradative, microbial, and enzymatic degradation. Polym Eng Sci. 2020;60:2061–75.
14. Malone LP, Best SM, Cameron RE. Accelerated degradation testing impacts the degradation processes in 3D printed amorphous PLLA. Front Bioeng Biotechnol. 2024;12:1419654.
15. Chen F, Han J, Guo Z, et al. Antibacterial 3D-printed silver nanoparticle/poly lactic-co-glycolic acid (PLGA) scaffolds for bone tissue engineering. Materials. 2023;16(11):3895.
16. González-González AM, Cruz R, Rosales-Ibáñez R, et al. In vitro and in vivo evaluation of a polycaprolactone (PCL)/polylactic-co-glycolic acid (PLGA) (80: 20) scaffold for improved treatment of chondral (cartilage) injuries. Polymers (Basel). 2023;15(10):2324.
17. Zheng Y, Pan P. Crystallization of biodegradable and biobased polyesters: polymorphism, cocrystallization, and structure-property relationship. Prog Polym Sci. 2020;109:101291.
18. Nifant'ev IE, Tavtorkin AN, Shlyakhtin AV, Ivchenko PV. Chemical features of the synthesis, degradation, molding and performance of poly(lactic-co-glycolic) acid (PLGA) and PLGA-based articles. Eur Polym J. 2024;215:113250.
19. Chor A, Gonçalves RP, Costa AM, et al. In vitro degradation of electrospun poly(lactic-co-glycolic acid) (PLGA) for oral mucosa regeneration. Polymers (Basel). 2020;12(8):1853.
20. Aoki K, Ideta H, Komatsu Y, et al. Bone-regeneration therapy using biodegradable scaffolds: calcium phosphate bioceramics and biodegradable polymers. Bioengineering. 2024;11(2):180.
21. Khodaei T, Schmitzer E, Suresh AP, Acharya AP. Immune response differences in degradable and non-degradable alloy implants. Bioact Mater. 2023;24:153–70.
22. Melsen F, Mosekilde L. Tetracycline double-labeling of iliac trabecular bone in 41 normal adults. Calcif Tissue Res. 1978;197(26):99–102.
23. Parfitt AM, Drezner MK, Glorieux FH, et al. Bone histomorphometry: standardization of nomenclature, symbols, and units: report of the asbmr histomorphometry nomenclature committee. J Bone Miner Res. 1987;2:595–610.
24. Pearce AI, Richards RG, Milz S, et al. Animal models for implant biomaterial research in bone: a review. Eur Cell Mater. 2007;13:1–10.

25. Kawabata T, Ikeda M, Miyakoshi C, et al. Degradation of unsintered hydroxyapatite and Poly-L-lactide composite sheets in vivo and in vitro. J Craniofac Surg. 2024;35(8):2225–30.
26. Pietrzykowska E, Romelczyk-Baishya B, Chodara A, et al. Microstructure and mechanical properties of inverse nanocomposite made from poly-lactide and hydroxyapatite nanoparticles. Materials. 2022;15(1):184.
27. Pietrzykowska E, Romelczyk-Baishya B, Wojnarowicz J, et al. Preparation of a ceramic matrix composite made of hydroxyapatite nanoparticles and polylactic acid by consolidation of composite granules. Nanomaterials. 2020;10(6):1060.
28. Diao H, Si Y, Zhu A, et al. Surface modified nano-hydroxyapatite/poly(lactide acid) composite and its osteocyte compatibility. Mater Sci Eng C. 2012;32:1796–801.
29. Liao W, Jiang X. Mini-implant anchorage: much is still unknown. Am J Orthod Dentofacial Orthop. 2019;156:709.
30. Essig H, Lindhorst D, Gander T, et al. Patient-specific biodegradable implant in pediatric craniofacial surgery. J Cranio-Maxillofac Surg. 2017;45:216–22.
31. Singh A, Muthunagai R, Agarwal M, et al. Experience with resorbable plates for fixation of mandible fracture. A prospective study of 10 cases. J Pharm Bioallied Sci. 2022;14(45):216–22.
32. Deng M, Zhou J, Chen G, et al. Effect of load and temperature on in vitro degradation of poly(glycolide-co-L-lactide) multifilament braids. Biomaterials. 2005;26:4327–36.
33 Farrar DF. Gillson RK Hydrolytic degradation of polyglyconate B: the relationship between degradation time, strength and molecular weight. Biomaterials. 2002;23(18):3905–12.
34. Smith A, Hosein YK, Dunning CE, Tassi A. Fracture resistance of commonly used self-drilling orthodontic mini-implants. Angle Orthodontist. 2015;85:26–32.
35. Wu Y, Zeng W, Xu J, et al. Preparation, physicochemical characterization, and in vitro and in vivo osteogenic evaluation of a bioresorbable, moldable, hydroxyapatite/poly(caprolactone-co-lactide) bone substitute. J Biomed Mater Res A. 2023;111:367–77.
36. Ding M, Shi J, Wang W, et al. Early osseointegration of micro-arc oxidation coated titanium alloy implants containing Ag: a histomorphometric study. BMC Oral Health. 2022;22(1):628.
37. Hasegawa S, Ishii S, Tamura J, et al. A 5–7 year in vivo study of high-strength hydroxyapatite/poly(L-lactide) composite rods for the internal fixation of bone fractures. Biomaterials. 2006;27:1327–32.
38. Shikinami Y, Okuno M. Bioresorbable devices made of forged composites of hydroxyapatite (HA) particles and poly-L-lactide (PLLA): part I. Basic characteristics. Biomaterials. 1999;20:859–77.
39. Shikinami Y, Okuno M. Bioresorbable devices made of forged composites of hydroxyapatite (HA) particles and poly l-lactide (PLLA). Part II: practical properties of miniscrews and miniplates. Biomaterials. 2001;22(23):3197–211.
40. Sakamoto A, Okamoto T, Matsuda S. Unsintered hydroxyapatite and poly-L-lactide composite screws/plates for stabilizing β-tricalcium phosphate bone implants. Clin Orthop Surg. 2018;10:253–9.
41. Majzoub Z, Finotti M, Miotti F, et al. Bone response to orthodontic loading of endosseous implants in the rabbit calvaria: early continuous distalizing forces. Eur J Orthod. 1999;21:223–30.
42. Melsen B, Lang NP. Biological reactions of alveolar bone to orthodontic loading of oral implants. Clin Oral Implants Res. 2001;12:144–52.
43. Chatzigianni A, Keilig L, Reimann S, et al. Effect of mini-implant length and diameter on primary stability under loading with two force levels. Eur J Orthod. 2011;33:381–7.
44. Lang NP, Imber JC, Lang KN, et al. Sequential osseointegration of a novel implant system based on 3D printing in comparison with conventional titanium implants. Clin Oral Implants Res. 2023;34:627–38.
45. Freire JNO, Silva NRFA, Gil JN, et al. Histomorphologic and histomophometric evaluation of immediately and early loaded mini-implants for orthodontic anchorage. Am J Orthod Dentofacial Orthop. 2007;131:704.e1-704.e9.

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