Effect of Mini-implant Diameters on Primary Stability and Viscoelastic Migration of Mini-implants under Orthodontic Loading

Thesis

Presented in Partial Fulfillment of the Requirements for the Degree Master of Science in the Graduate School of The Ohio State University

By

Jim Lee, D.D.S, M.P.H., M.S.

Graduate Program in Dentistry

The Ohio State University

2013

Thesis Committee:

Dr. Do-Gyoon Kim, Advisor

Dr. William M. Johnston

Dr. Henry Fields
ABSTRACT

Objectives The objectives of the current study were to examine effects of mini-implant diameters on 1) primary stability and 2) time-dependent migration of mini-implant systems under a tangential loading.

Materials and Methods Twenty one mini-implants (7 for each 1.4 mm, 1.6 mm, and 2 mm diameter mini-implant) were placed in mandibular sections of human cadavers (4 males and 3 females, 69.7±13.1 years). All mini-implants were 8 mm in length. A tangential load of 2N was applied to the mini-implant in bone under hydration for 2 hours. Migration was assessed as the change in displacement during initial loading followed by 2 hours constant loading. Mean and standard deviation (SD) of bone mineral density (BMD) were obtained using histogram of cone-beam computed tomography (CBCT) attenuation values surrounding the mini-implant hole. The cortical thickness along with the miniscrew hole was also measured using the CBCT image.

Results The mini-implant diameters did not have an effect on determining the primary stability and migration of mini-implant (p>0.147). Positive correlations of migration with SD and COV were found (p<0.073). These correlations were independent of the mini-implant diameters (ANCOVA, p>0.11).
Conclusions  Time-dependent viscoelastic creep migration following the static displacement at the small level of orthodontic constant loading suggested that more complications may be produced due to the micromotion at the immediate loading after implantation.
DEDICATION

Dedication to my family

Thank you for your support
ACKNOWLEDGEMENTS

I would like to thank my committee members: Dr. Do-Gyoon Kim, Dr. Henry Fields, and Dr. William Johnston, for their support and guidance in the development of this thesis. Their dedication to quality research is a pillar to our program.

I would also like to thank my family and friends for their love and support. It was their positive energy that revitalized me at times when I grew discouraged. I also want to thank my co-residents, Bethany Crawford, Jessica Liu, Charu Swamy, and Alpesh Patel, who help me through hard times with their comradery. I would like to give special thanks to Dr. Yonghoon Jeong and Joseph Pittman for their help in the lab and their instructions in developing methodology around my treatment. Also, a special thank you to the radiology department at OSU dental school.

I would like to thank the Delta Dental Foundation for providing financial support for this research through the Dental Master’s Thesis Award Program. And would also like to thank Dr. Amanda M. Agnew at Division of Anatomy, College of Medicine of OSU for the donation for the donation of the mandibles and the OSU Division of Orthodontics. I would like to also thank Shane Burden of Rocky Mountain Orthodontics for the miniscrews and miniscrew equipment.
VITA

January 21st, 1978.........................................................Born – Raleigh, NC

1996-2000.................................................................B.A History
George Washington University
Washington DC

2001-2003.................................................................Master in Public Health
Social and Behavioral Sciences
Boston University
School of Public Health
Boston MA

2006-2010.................................................................Doctor of Dental Surgery
University of Maryland
School of Dentistry
Baltimore MD

2010- Present ........................................................Orthodontic Residency
The Ohio State University
Columbus OH

FIELDS OF STUDY

Major Field: Dentistry
Specialty: Orthodontics
# TABLE OF CONTENTS

<table>
<thead>
<tr>
<th>Abstract</th>
<th>ii</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dedication</td>
<td>vi</td>
</tr>
<tr>
<td>Acknowledgements</td>
<td>v</td>
</tr>
<tr>
<td>Vita</td>
<td>vi</td>
</tr>
<tr>
<td>List of Tables</td>
<td>viii</td>
</tr>
<tr>
<td>List of Figures</td>
<td>ix</td>
</tr>
</tbody>
</table>

## Chapters:

1. Introduction .................................................................1
2. Materials and Methods..................................................6
3. Manuscript.................................................................12
4. Comprehensive Results and Discussion.................................36

Works Cited........................................................................41
# LIST OF TABLES

<table>
<thead>
<tr>
<th>Table</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Initial conditions and migration parameters measured for migration of the mini-implants with different diameters.</td>
<td>27</td>
</tr>
<tr>
<td>2. Significant or marginally significant correlations between initial conditions and migration parameters for the pooled group (n=21) and among migration parameters.</td>
<td>28</td>
</tr>
</tbody>
</table>
## LIST OF FIGURES

<table>
<thead>
<tr>
<th>Figure</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1. Site of a mini-implant installation in a human mandibular specimen, and b) a 3D CBCT image (200 micron voxel size) of a human mandibular specimen after removal of the mini-implant</td>
<td>29</td>
</tr>
<tr>
<td>2. Constant functional loading (2N) applied via a loading machine to a mini-implant installed in a wet human mandible. The red arrowhead and white arrow indicate the mini-implant head and loading direction, respectively</td>
<td>30</td>
</tr>
<tr>
<td>3. A typical force-displacement curve</td>
<td>31</td>
</tr>
<tr>
<td>4. No significant correlations were evident between migration and a) cortical thickness (p&gt;0.854, n=21) or b) BMD Mean (p&gt;0.246, n=21)</td>
<td>32</td>
</tr>
<tr>
<td>5. Strong positive correlation between creep and residual displacement for each and the pooled group (r=0.923, p&lt;0.001, n=21)</td>
<td>33</td>
</tr>
</tbody>
</table>
CHAPTER 1
INTRODUCTION

In 1907 Edward Angle, often regarded as the father of modern orthodontics, wrote in his book, Malocclusions of the Teeth: “According to the well-known law of physics, action and reaction are equal and opposite, hence it must follow that the resistance of anchorage must be greater than that offered by the tooth to be moved, otherwise there will be displacement of anchorage and failure in the movement of the teeth to the extent, or possibly, in the direction desired. The sources at our disposal for securing anchorage or resistance, are first, the teeth themselves, and second, sources external to the teeth. (E, 1907)”

Orthodontists often require a stable and rigid “anchor” to which teeth can be attached to by wires and appliances so that they can be moved, and this concept of anchorage made this feat challenging. Anchorage, defined as “resistance to unwanted tooth movement, (J, 2000)” usually include grouping anchor teeth together to maximize anchorage. However, due to the physiological nature of the underlying soft junction between the tooth and bone which are responsible for the movement of teeth, these anchor teeth are vulnerable to unwanted movement. Extraoral anchorage, such as headgear, which uses a stable nonmovable structure outside the mouth, is another way to provide anchorage but effectiveness of this treatment is dependent on whether the patient is using the headgear.
In 1945, Gainforth and Higley placed a fairly biocompatible vitallium alloy into canine test subjects to retract canine teeth (Gainsforth & Higley, 1945). Nearly 20 years later in 1983, Creekmore and Eklund used vitallium alloy miniscrews on the first human clinical case to correct a deep bite (Creekmore & Eklund, 1983). Despite its successful case studies in providing skeletal anchorage, use of miniscrews did not become popularized until Kanomi introduced the biocompatible titanium orthodontic miniscrew in 1997 (Kanomi, 1996) after which Costa designed a miniscrew with a bracket head (Costa, Raffaini, & Melson, 1998). Today, there are a variety of miniscrews including those with different diameters, different lengths, and different compositions.

This recent introduction of orthodontic miniscrews is often considered by many clinicians and biomechanical experts to be a “paradigm” shift in the field of orthodontics. Orthodontic miniscrews provide an opportunity to move teeth around without unwanted movement of anchor teeth and do not require patient compliance. Orthodontic miniscrews are shaped like small screws and are made of titanium metal which is not known to cause an allergic reaction. Wires and appliances use miniscrews to provide stable anchorage during the movement of teeth. Unlike the teeth, these stable anchors do not have a soft tissue attachment to the bone which is responsible for tooth movement. Therefore, forces exerted on the screw do not initiate the cascade of events that eventually leads to the movement of our teeth.

One of the disadvantages to orthodontic miniscrews include the device’s high failure rate. A review of literature found the failure rate of miniscrews to be between 13.4 – 20.1% (Schatzle, Mannchen, Zwahlen, & Lang, 2009). Moreover, recent studies have suggested that Orthodontic miniscrews do move (Eric, Liou, Betty, Pai, James, & Lin,
The damage in the bone caused by installation and force application of the insertion of the miniscrew into the bone can lead to migration of the screw (Chen, Kang, Bae, & Kyung, 2009). Other studies on the histological study of the implant site suggest that migration of miniscrews is caused by a reactive remodeling process of the bone (will find reference).

In the literature, the immediate stability and the prolonged stability of the miniscrew after placement is separated into two stages: primary stability and secondary stability. Primary stability or initial stability describes the mechanical retention and stability of the miniscrew to forces of immediate loading (Wilmes, Rademacher, Olthoff, & Drescher, Parameters affecting primary stability of orthodontic miniscrews, 2006). Researchers suggest that the main factors influencing the primary stability of miniscrews are: miniscrew design, operator placement, and both the quality and the quantity of bone in the implant site (Cha, Kil, Yoon, & Hwang, Miniscrew stability evaluated with computerized tomography scanning, 2010).

Secondary stability refers to the stability of the implants in the later stages of loading after initial and progressive biological factors such as remodeling, formation/resorption of bone, and secondary tissue formation (Wilmes, Rademacher, Olthoff, & Drescher, Parameters affecting primary stability of orthodontic miniscrews, 2006). It differs from primary stability in that its progressive or degradation of stability is influenced primarily by the biological host factors. Primary stability factors as part of total secondary stability and is an important positive or negative influence on these biological factors of secondary stability (Baumgaertel, 2010).
Several measures of primary stability have been documented. Some researchers have used insertion torque (Cha, Kil, Yoon, & Hwang, Miniscrew stability evaluated with computerized tomography scanning, 2010) (Wilmes, Rademacher, Olthoff, & Drescher, Parameters affecting primary stability of orthodontic miniscrews, 2006). Other researchers have explored pull out or tension tests (Pickard, Dechow, Rossouw, & Buschang, 2010) (Huja, Litsky, Beck, Johnson, & Larsen, 2005). Another measure of primary stability could be derived from the displacement observed in a shear test or tangential loading test (Morarend C., et al., 2009) (Iijima, et al., 2012) where a miniscrew is inserted into the bone and force application is directed perpendicularly into the head of the miniscrew to simulate forces in orthodontic applications.

Though miniscrew stability is determined by a variety of factors including miniscrew design, location, placement angle, and bone density, if the load of force on the miniscrews is high enough, consequential damage and creep from the load may lead to unintentional tooth movement and may eventually cause miniscrew failure. In previous literature, the bone quality and quantity was often seen assessed as important factors in the primary stability of the miniscrew. Quantity refers primarily to cortical bone thickness and quality usually suggests bone density. A popular modality in the placement in endosseous dental implants, orthodontics have utilized 3D CBCT as a noninvasive way to assess not only bone quality and quantity, but to provide guidance in placing TADs around vital structures. The use of CBCT to determine cortical bone thickness and bone mineral density for the placement of orthodontic miniscrews has become increasingly popular recently to assist in their placement (Lee, Joo, Kim, Lee, Park, & Yu, 2009) (Cha, Kil, Yoon, & Hwang, Miniscrew stability evaluated with computerized tomography...
By using voxel attenuation or the gradual loss of voxel intensity in CBCT images, researchers are able to determine the bone mineral density (Kimpe & Tuytschaever, 2007).

The objective of this study was to examine potential factors that can influence the miniscrew migration in mandible under physiologic forces. The factors tested in this study include screw diameter, cortical bone thickness, bone mineral density (BMD), and standard deviation (SD) and Coefficient of Variation (COV) of BMD. Our null hypothesis is that an increase in diameter, cortical bone thickness, BMD, and SD/COV of BMD will not affect the amount of initial stability and therefore will not provide a statistically significant amount of displacement and creep observed after the physiologic force application of 2 N.
CHAPTER 2

MATERIALS AND METHODS

Miniscrews:

Twenty-one miniscrews of three different diameters were used in the study (7 of 1.4 mm, 7 of 1.6 mm, and 7 of 2 mm diameters). The miniscrews, Dual-Top TADs from Rocky Mountain Orthodontics, were available in three diameters with other properties such as thread design, length, and head design/dimensions being the same. All miniscrews were 8 mm in length, had a reverse buttress thread design, and a standard head (screw top with collar) for loading (RMO # G00412[1.4mm]/# G00414[1.6mm]/# G00417[2.0mm])

Bone preparation:

Whole mandibles, which were derived from 7 subjects (4 males and 3 females, 69.7±13.1 years), were donated mandibles from the OSU College of Medicine. The mandibles were not chemically treated and had all soft tissue was carefully removed. Using a high speed bone saw, mandibles were sectioned anteriorly into two halves. Then, the condyles were removed at the level of the occlusal plane. Any dentition with metal
restorations were removed using a handpiece with a cutting disc. Mandibular specimens would then be sectioned into 10 mm sections using a precise parallel blade slow speed bone saw. The sections were placed in individual labeled containers and frozen. To keep the specimens from surface desiccation and to avoid leaching of minerals out of the specimens, thawed specimens were kept in an environmental solution (Table 1)

**Miniscrew placement**

Miniscrew placement is planned in areas were miniscrews can be placed away from root structures and placed on the day of loading. At the location of miniscrew placement, a pilot hole using a 1mm RMO pilot bur (RMO # G00231) in a slow speed handpiece and miniscrews are placed with a miniscrew driver perpendicular (90°) to the cortical bone. Miniscrews are tightened until only one thread is showing outside the cortical bone. The collar therefore is about 1.5 mm above the cortical bone. The side is then carefully examined for large cracks or fractures. An attempt was made to place screws of all three diameters into the same specimen.

**Biomechanical Testing**

The study utilized a Bose Bone ELF 3320 machine with an environment chamber containing a the solution above to keep the bone segment from drying out. The ELF is testing machine is designed for high resolution/accuracy displacement measurements using both a internal measuring device and a external laser displacement measurement device capable of measuring displacement to the nearest micron. The instrument also
provides for dynamic loading for creep study. The segment will be placed in the environmental chamber to thaw completely for 2 hours before the experiment. Using a loading jig directly from the machine to the miniscrew head on the bone, ELF machine will both apply the force of 2 N on the TAD to the bone and measure the amount of creep (distance in mm), maximum displacement, and residual displacement from the baseline. A force of 2N was found to be within the physiologic parameters of the normal everyday function of bone (Crismani & Bertil, 2010) and was found to provide adequate force for orthodontic movement with no significant influence on the stability of the screw. During the experiment, the specimen was kept in a chamber filled with an environmental solution to avoid surface drying which could alter our results.

After the miniscrew has completed the loading process, the miniscrew is removed and another round of scans is initiated (200, 300, and 400 voxels) on the same iCAT CBCT machine. About 10 specimens were included in each scan. Raw Dicom files were processed using iCATVision and then analysis of the scans conducted using ImageJ, a public domain image processing software developed by the National Institute of Health (http://rsbweb.nih.gov/ij/). Using ImageJ, the Dicom files were converted into a single TIFF file where each specimen is cropped into separate TIFF files.
Cone Beam CT Analysis

Bone density and cortical bone thickness will be analyzed through clinical cone beam computed tomography (CBCT) using an iCAT (PA) both before miniscrew insertion and after miniscrew removal. The CBCT scan will be used to assess cortical bone thickness and bone mineral density (BMD) at 200, 300, and 400 voxels. Preinsertion scans were used primarily to determine ideal locations for implant placement. The average thickness of buccal cortical bone was found to be 1.26 to 2.91mm (Park & Cho, 2009). An attempt was made to keep the bone thicknesses of implant site within this range. The site would also be free of vital structures such as roots and of adequate length to provide clearance for up to 8mm to accommodate the 8 mm of the intrabony portion of the miniscrew to avoid bicortical contact..

Cortical bone thickness was measured through the average of four measurements made at the top, the bottom, and the two sides of the miniscrew hole. Bone mineral density, SD, and COV of BMD were all assessed within the cortical bone. By utilizing the attenuation of gray scales on the CBCT images, the density of bone was can be examined. The 200 voxel size scan was used in all of the image analysis. A gray level attenuation value of each voxel was used to measure the BMD of the bone segment being represented. Units were reported in Hounsfield units (HU).

After removal of the mini-implant, the bone specimens were scanned using a three dimensional (3D) cone beam computed tomography (CBCT) (iCAT, PA) with a 200 micron voxel size (120 kV, 5 mA, and 26.9 seconds) (Fig. 1b). Thus, X-ray artifacts
from the metal mini-implant, which may interrupt gray levels in the CBCT image, were avoided. Using the CBCT image, the intrabony portion of the mini-implant was examined. Any specimens involving bicortical contact or root contact were omitted from the study. Cortical bone thickness (Thickness) was estimated using the 3D CBCT image by counting the number of image voxels through the quadrant sides of the mini-implant hole, which were multiplied by 200 microns voxel size, and an averaged value of thickness was obtained. Bone mineral density was assessed using the gray levels in CBCT image. As a CT attenuation value (Hounsfield Unit, HU) of an air voxel is -1000, the gray level of bone voxel was computed by adding 1000 to the CT value using imaging software (ImageJ, NIH). Then, a mineral density of each bone voxel was obtained by calibrating the gray level using a linear relationship with known densities from a commercial hydroxyapatite phantom set following the similar process of calibration used in a previous study [3]. A two-voxel size (400 micron) layer along the longitudinal axis of the mini-implant hole was isolated (Fig. 1b). The mean (Mean) bone mineral density (BMD) was computed by dividing the sum of mineral densities of the bone voxels by its total volume. A standard deviation (SD) of voxel BMD distribution was also computed and a coefficient of variation (COV) was obtained by dividing the SD by the Mean.

**Statistical Analysis**

Based on a preliminary study for viscoelastic migration of other commercial orthodontic mini-implants in bone [10], the sample size of experiment was estimated to be 4 for each group with a power of 0.95. Thus, the current study assigned 7 mini-implants for each group. Inter-rater reliability for measures of BMD parameters for 5
specimens was determined using the intraclass correlation coefficient single score method of Shrout and Fleiss [11]. Analysis of variance (ANOVA), followed by Fisher’s PLSD post hoc test, was utilized to compare the initial condition (Thickness, and BMD Mean, SD and COV) and migration parameters (d, C, d_{max}, and d_{res}) between the three different mini-implant diameter groups. Pearson’s correlation was used to examine correlations between the initial condition and migration parameters, and between migration parameters. Analysis of covariance (ANCOVA) was used to determine if the correlations were different depending on the mini-implant diameters. Significance was set at p<0.05 for all statistical tests.
CHAPTER 3
MANUSCRIPT

Effect of Mini-implant Diameters on Primary Stability and Viscoelastic Migration of Mini-implants under Orthodontic Loading

Jim Lee¹, Yonghoon Jeong², Joseph Pittman¹, William Johnston², Henry Fields¹, Do-Gyoon Kim¹,*

¹Division of Orthodontics, College of Dentistry, The Ohio State University
²Division of Restorative and Prosthetic Dentistry, College of Dentistry, The Ohio State University, Columbus, OH, USA

*For correspondence

Do-Gyoon Kim, Ph.D
Assistant Professor
Division of Orthodontics, College of Dentistry
The Ohio State University
4088 Postle Hall
305 W. 12th Ave
Columbus, OH 43210
USA
email: kim.2508@osu.edu
Tel) (614) 247-8089
Fax) (614) 688-3077

Submitted: Clinical Oral Investigations (3/29/2013)
**Introduction**

Mini-implants have been widely used as a temporary anchorage device to sustain a constant load during orthodontic treatment [1, 2]. As orthodontic loading is applied immediately after implantation, the primary stability of the mini-implant system is essential in maintaining a high success rate [3, 4]. On the other hand, long-term migration of mini-implants was observed in previous clinical studies [5, 6]. Because these studies observed the progressive migration of a mini-implant up to 1.6 mm during orthodontic treatment, it was recommended that the mini-implant should be placed at least 2.0 mm away from adjacent critical structures, such as adjacent roots and nerves. However, factors involved in the primary stability relating to long-term migration of mini-implants are not been fully understood.

Many studies examined contributing factors to the primary stability including bone thickness and mineral density as host factors, and dimensions of the mini-implant design as implant factors [2, 7, 8]. These studies measured axial pull out strength and insertion torque of mini-implant in boney specimens. While these loading regimes may help understand the elastic, plastic and contact mechanical behavior of mini-implant system, they did not reflect the small tangential constant load associated with functional orthodontic loading. Few studies have been performed to examine the mechanical response of mini-implant system to the functional orthodontic tangential loading.
We hypothesized that the migration of mini-implants will be explained by the factors that are related to its stability. Thus, the objectives of the current study were to examine the effects of mini-implant diameters on 1) the primary stability and 2) the time-dependent migration of mini-implant systems under a tangential loading.

Materials and methods

Based on a preliminary study for viscoelastic migration of other commercial orthodontic mini-implants in bone [9], the sample size of experiment was estimated to be 4 for each group with a power of 0.95. Thus, the current study assigned 7 mini-implants for each group.

Similarly, seven human mandibles were obtained from cadavers (4 males and 3 females, 69.7±13.1 years) which were donated to a university college of medicine department of anatomy. These cadaveric bones were not chemically treated. After careful removal of all soft tissue, the mandibles were stored in a -22 ºC freezer until used. The frozen mandibles were thawed at a room temperature and dissected into 10 mm thick sections in the bucco-lingual direction using a low speed saw with double parallel diamond blades under irrigation (Fig. 1a). Edentulous mandibular regions were not included.

Twenty-one mini-implants of 3 different diameters (7 of 1.4 mm, 7 of 1.6 mm, and 7 of 2 mm diameters) were used in the study. The mini-implants were purchased (Dual-Top TADs, Rocky Mountain Orthodontics, Denver, Colorado)). All other dimensions other than the diameter of mini-implant including thread design, length (8
mm), and head design were identical. The locations of implantation were carefully
determined to avoid contact with critical structures such as roots. [5, 6]. A pilot hole was
drilled using a 1mm RMO pilot bur (RMO # G00231 Rocky Mountain Orthodontics,
Denver, Colorado) with a slow speed handpiece. Then, the mini-implants were inserted
into the pilot hole with a mini-implant driver perpendicular to the surface of bone
specimen. Caution was used to minimize angulations of the mini-implants. The mini-
implants were torqued until the collar was about 1.5 mm above the surface of bone
specimen. Each diameter mini-implant was placed in each specimen. Specimens that
developed large cracks or fractures during placement of the mini-implants were
eliminated. The bone specimen was hydrated during the insertion process.

A calcium buffered solution, containing a Ca++ concentration (57.5 mg/l) equal to
that in bone specimens [10], and an antifungal agent (0.2 g/l gentamicin) were used to
soak the bone specimen with mini-implant for at least 2 hours prior to mechanical testing.
An environmental chamber filled with the calcium buffered solution was mounted on the
loading machine (Bose ELF 3320, Minnetonka, MN) (Fig. 2). A 450 N load cell and a
high resolution (15 nm) displacement transducer were used to measure force and
displacement during mechanical testing. A special jig was fabricated to hold the
specimen in the environmental chamber on the load cell. The mechanical testing was
performed at room temperature.

Another jig was used to apply elastic and creep tangential loads perpendicularly to
a point between the collar and pin head of each mini-implant (Fig. 2). A compressive
preloading of up to 0.5 N and a precyclic loading of up to -2 N were achieved for a solid
contact between the loading jig and the mini-implant head surface. After the
preconditioning, the mini-implant head was compressively loaded using -2 N at a loading rate of 0.1 N/sec following an orthodontic functional tangential loading of up to -2 N (approximately 200 g), which has been recommended in previous studies [1, 2]. The constant load of -2 N was maintained for 2 hours and fully unloaded at the same loading rate (Fig. 3). Static elastic displacement ($d_i$) and maximum displacement ($d_{max}$) were measured at the end of static loading and 2 hours creep loading, respectively. Creep ($C_l$) was obtained by subtracting the initial elastic displacement ($d_i$) from the maximum displacement ($d_{max}$) (Fig. 3). Residual displacement ($d_{res}$) was measured as an unrecovered permanent deformation at the end of the unloading process.

After removal of the mini-implant, the bone specimens were scanned using a three dimensional (3D) cone beam computed tomography (CBCT) (iCAT, PA) with a 200 micron voxel size (120 kV, 5 mA, and 26.9 seconds) (Fig. 1b). Thus, X-ray artifacts from the metal mini-implant, which may interrupt gray levels in the CBCT image, were avoided. Using the CBCT image, the intrabony portion which had contact with the mini-implant was examined. Any specimens involving bicortical contact or root contact were omitted from the study. Cortical bone thickness was estimated using the 3D CBCT image by counting the number of image voxels on four sides of the mini-implant hole, which were multiplied by 200 microns voxel size, and an averaged value of thickness was obtained. Bone mineral density was assessed using the gray levels in CBCT image. Because a CT attenuation value (Hounsfield Unit, HU) of an air voxel is -1000, the gray level of bone voxel was computed by adding 1000 to the CT value using imaging software (ImageJ, NIH, Bethesda, MD). Then, a mineral density of each bone voxel was obtained by calibrating the gray level using a linear relationship with known densities.
from a commercial hydroxyapatite phantom set following the similar process of calibration used in a previous study [3]. A two-voxel size (400 micron) layer along the longitudinal axis of the mini-implant hole was isolated (Fig. 1b). The mean (Mean) bone mineral density (BMD) was computed by dividing the sum of mineral densities of the bone voxels by its total volume. A standard deviation (SD) of voxel BMD distribution was also computed and a coefficient of variation (COV) was obtained by dividing the SD by the mean.

Inter-rater reliability for measures of BMD parameters for 5 specimens was determined using the intraclass correlation coefficient single score method of Shrout and Fleiss [11]. Analysis of variance (ANOVA), followed by Fisher’s PLSD post hoc test, was utilized to compare the initial condition (Thickness, and BMD Mean, SD and COV) and migration parameters (d, C, d_{max}, and d_{res}) between the three different mini-implant diameter groups. Pearson’s correlation was used to examine correlations between the initial condition and migration parameters, and among migration parameters. Analysis of covariance (ANCOVA) was used to determine if the correlations were different depending on the mini-implant diameters. Significance was set at p<0.05 for all statistical tests.

**Results**

All mini-implants penetrated completely through the cortical bone (Fig. 1). Viscoelastic creep was assessed for all of the mini-implants in the human cadaveric mandible at the functional load level of orthodontic treatment. Inter-rater reliability
between raters JL and JP was 0.94, 0.99, and 0.97 for mean, SD and COV of BMD, respectively.

All parameters examined in the current study were not significantly different between the different mini-implant diameter groups (p>0.275) (Table 1). An amount of the migration ($d_{max}$) of the mini-implant, which combined elastic displacement ($d_l$) and viscoelastic creep ($C_l$), was detected in the loading direction for all specimens (Fig. 3). The displacement was not fully recovered after the mini-implant was unloaded leaving the residual displacement also for all specimens ($d_{res}$).

The bone mineral density (BMD) variability (SD and COV) did not have significant positive correlations with creep and residual displacement (Table 2). Neither the cortical thickness nor mean BMD had significant correlations with migration (Fig. 4). There was no correlation between the static elastic displacement and creep (p>0.310). However, the static elastic displacement had positive correlations with the maximum displacement and residual displacement (p<0.015). Creep had strong positive correlations with the maximum displacement and residual displacement (p<0.001) (Fig. 5). The maximum displacement had a strong positive correlation with residual displacement (p<0.001). The significant correlations were independent of the mini-implant diameters (ANCOVA, p>0.11).

**Discussion**

The experimental conditions used in the current study allowed investigation of the exclusive effects of mini-implant diameters on the early mechanical response of mini-implants to orthodontic loading. Primary migrations combining the static elastic
displacement and the time-dependent viscoelastic creep were detected for all the mini-implants under the functional orthodontic loading level. These migrations occurred independent of mini-implant diameters. The traditional parameters of bone thickness and mean BMD, which were thought to have an effect on the mechanical initial stability of mini-implant system, did not explain the primary migration. However, the variability (SD and COV) of BMD had significant correlations with creep suggesting that bone quality, rather than bone quantity, plays a role in determining the viscoelastic mechanical behavior of mini-implant systems regardless of the mini-implant diameters.

Most of studies to examine the mechanical stability of mini-implant systems measured static pull-out strength of mini-implants in bone specimen [8, 12, 13]. Forces of greater than 200 N were applied to assess the pull-out strength. However, this amount and direction of pull-out loading could not represent the normal orthodontic loading in the tangential direction at the small functional level of 2N. In addition, the bone-implant interface was destroyed after static pull-out testing making it impossible to investigate the time-dependent changes of mini-implant stability under constant loading following insertion. The constant tangential loading conditions used in the currently study could provide more practical information for the mechanical behavior of mini-implant systems at the immediate stage in implantation in a clinical setting.

Placement of the mini-implant at an oral bone location with high bone mass (bone thickness and BMD) has been strongly recommended to obtain better initial mechanical stability and long-term success of the mini-implant system [3, 8, 12, 14]. It was frequently indicated that an increase in diameter and length of mini-implant helped enhance the stability of mini-implant system [7, 15]. Longer mini-implants with larger
diameter had greater bone-implant contact area enhancing stability against forces of applied loading. In the current study, with the same level of tangential loading (2N) applied to all mini-implant systems, the static displacement accounted for their primary stability. As a result, we found no significant effects of mini-implant diameters on the stability of mini-implant system. This lack of significant effects may have resulted from that the substantial thickness (average 2.15 mm) of dense cortical bone specimens in the current study, which was measured at the range of 0.86 to 2.49 mm for human mandibular bone thickness [16], likely provided strong stability of the mini-implant systems reducing the effect from different diameters.

Micromotion at a bone-implant interface has been considered as one of the most important factors in determining the primary success of the implant system [4, 17]. It was indicated that excessive micromotion increases the risk of fibrous tissue development at the implant interface, which can lead to failure of the implant system [18-20]. In particular, the micromotion resulting from immediate loading prior to bone healing can interrupt bone regeneration at the bone-implant interface. The current finding demonstrates that mini-implant migration could develop by the immediate tangential loading at the level used in orthodontic clinical practice. This migration likely produces a gap between bone and implant surface at the opposite side of loading direction. The discontinuity from this interfacial gap may delay bone apposition to the surface of mini-implant and cause more complications including fibrous development and infection.

The current study observed that the migration continued under the prolonged constant post-implantation loading. This time-dependent viscoelastic creep was responsible for the permanent deformation of the interfacial bone with strong positive
correlations between the creep and residual displacement independent of mini-implant diameters. Many studies have observed the creep behavior of bone tissue under a long-term constant physiological loading [21-23]. It also has been observed that the magnitude of creep was controlled by variability of tissue mineral density [21, 22]. The current finding was consistent with those results since there were positive correlations between the variability (SD and COV) of BMD with the creep and residual deformation, but the magnitude of the relationship was low and minimally explained the variability. Nevertheless, the variability of bone mineralization increased by active bone remodeling may produce more new (less mineralized) bone tissue following resorption of pre-existing (more mineralized) bone tissue [21, 24, 25]. It is well known that the active remodeling is involved in healing at the bone-implant interface [4, 26, 27]. Taken together, these observations suggested that the migration of mini-implant may continue under immediate loading during the process of healing after placement. If the migration continuously progresses, it is possible that mini-implant may violate critical structures such as nerves and roots of adjacent teeth. This concern is aligned with the recommendation regarding 2 mm of clearance between critical structures and the location of mini-implant placement [5, 6].

A limitation of the current study was that the mini-implant was placed after removal of the soft tissue on the human cadaveric mandibular bone. Some previous clinical studies indicated that the soft tissue type is associated with stability and success rates of mini-implant systems [28, 29], while other case showed no correlation between survival rates and the surface characteristics including soft tissue [30]. More studies need to clarify the soft tissue effects on stability of mini-implant. Another limitation was that
we performed in vitro experiments with no biological activities during healing at the bone-implant interface. Thus, the current findings were limited to the primary stability and migration of mini-implant systems under the immediate loading after implantation.

In conclusion, the mini-implant diameters did not have an effect on determining the primary stability and migration of mini-implant in the thick and dense human mandibular cortical bone. Time-dependent viscoelastic creep migration continued following the static displacement at the small level of orthodontic constant loading suggesting that more complications may be produced due to the micromotion at the immediate loading after implantation. The viscoelastic creep of mini-implant under the orthodontic functional constant loading demonstrated limited associated with variability of BMD surrounding the implant site, which is likely due to active remodeling during the post-implantation healing period. Thus, further systematic studies need to examine the effect of biological activities on long-term migration of mini-implant system.

**Acknowledgement**

This study was, in part, supported by Delta Dental Foundation. We would also like to thank Dr. Amanda M. Agnew at Division of Anatomy, College of Medicine of the Ohio State University for providing the human cadaver mandibles.
References


Tables and Figures

**Table 1.** Initial conditions and migration parameters measured for migration of the mini-implants with different diameters.

**Table 2.** Significant or marginally significant correlations between initial conditions and migration parameters for the pooled group (n=21) and among migration parameters.

**Fig. 1** a) Site of a mini-implant installation in a human mandibular specimen, and b) a 3D CBCT image (200 micron voxel size) of a human mandibular specimen after removal of the mini-implant (white arrow head).

**Fig. 2** Constant functional loading (2N) applied via a loading machine to a mini-implant installed in a wet human mandible. The red arrowhead and white arrow indicate the mini-implant head and loading direction, respectively.

**Fig. 3** A typical force-displacement curve.

**Fig. 4** No significant correlations were evident between migration and a) cortical thickness (p>0.854, n=21) or b) BMD Mean (p>0.246, n=21).

**Fig. 5** Strong positive correlation between creep and residual displacement for each and the pooled group (r=0.923, p<0.001, n=21).
Table 1. Initial conditions and migration parameters measured for migration of the mini-implants with different diameters.

<table>
<thead>
<tr>
<th>Initial conditions</th>
<th>Diameters</th>
<th>p values</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>1.4 mm</td>
<td>1.6 mm</td>
</tr>
<tr>
<td>Thickness (mm)</td>
<td>2.13±0.408</td>
<td>2.018±0.344</td>
</tr>
<tr>
<td>BMD (mg/cm³)</td>
<td>Mean</td>
<td>1925.4±67.7</td>
</tr>
<tr>
<td></td>
<td>SD</td>
<td>243.6±50.9</td>
</tr>
<tr>
<td></td>
<td>COV</td>
<td>0.126±0.022</td>
</tr>
<tr>
<td>Migration parameters</td>
<td></td>
<td></td>
</tr>
<tr>
<td>dᵣ(mm)</td>
<td>0.017±0.003</td>
<td>0.018±0.005</td>
</tr>
<tr>
<td>Cᵣ(mm)</td>
<td>0.007±0.004</td>
<td>0.004±0.002</td>
</tr>
<tr>
<td>d_max=dᵣ+Cᵣ(mm)</td>
<td>0.023±0.006</td>
<td>0.023±0.006</td>
</tr>
<tr>
<td>d_res(mm)</td>
<td>0.008±0.004</td>
<td>0.006±0.004</td>
</tr>
</tbody>
</table>

Table 2. Significant or marginally significant correlations between initial conditions and migration parameters for the pooled group (n=21) and among migration parameters.

<table>
<thead>
<tr>
<th>x</th>
<th>y</th>
<th>Correlation</th>
<th>r</th>
<th>p-values</th>
</tr>
</thead>
<tbody>
<tr>
<td>BMD SD (mg/cm³)</td>
<td>C_l (mm)</td>
<td>y = 0.0000259x - 0.00075</td>
<td>0.420</td>
<td>0.057</td>
</tr>
<tr>
<td></td>
<td>d_res (mm)</td>
<td>y = 0.0000316x - 0.00081</td>
<td>0.417</td>
<td>0.060</td>
</tr>
<tr>
<td>BMD COV</td>
<td>C_l (mm)</td>
<td>y = 0.0575x - 0.0017</td>
<td>0.411</td>
<td>0.064</td>
</tr>
<tr>
<td></td>
<td>d_res (mm)</td>
<td>y = 0.0685x - 0.0018</td>
<td>0.400</td>
<td>0.073</td>
</tr>
<tr>
<td>d_l (mm)</td>
<td>d_max (mm)</td>
<td>y = 1.1753x + 0.0022</td>
<td>0.851</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td></td>
<td>d_res (mm)</td>
<td>y = 0.4787x - 0.0017</td>
<td>0.523</td>
<td>0.015</td>
</tr>
<tr>
<td>C_l (mm)</td>
<td>d_max (mm)</td>
<td>y = 1.3145x + 0.0153</td>
<td>0.711</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td></td>
<td>d_res (mm)</td>
<td>y = 1.132x + 0.0005</td>
<td>0.923</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>d_max (mm)</td>
<td>d_res (mm)</td>
<td>y = 0.5816x - 0.0064</td>
<td>0.878</td>
<td>&lt;0.001</td>
</tr>
</tbody>
</table>
Fig. 1 a) Site of a mini-implant installation in a human mandibular specimen, and b) a 3D CBCT image (200 micron voxel size) of a human mandibular specimen after removal of the mini-implant (white arrow head).
**Fig. 2** Constant functional loading (2N) applied via a loading machine to a mini-implant installed in a wet human mandible. The red arrowhead and white arrow indicate the mini-implant head and loading direction, respectively.
Fig. 3 A typical force-displacement curve.
Fig. 4 No significant correlations were evident between migration and a) cortical thickness (p>0.854, n=21) or b) BMD Mean (p>0.246, n=21).
Fig. 5 Strong positive correlation between creep and residual displacement for each and the pooled group ($r=0.923$, $p<0.001$, $n=21$).
CHAPTER 4
RESULTS AND COMPREHENSIVE DISCUSSION

RESULTS

All mini-implants penetrated completely through the cortical bone (Fig. 1). Viscoelastic creep was assessed for all of the mini-implants in the human cadaveric mandible at the functional load level of orthodontic treatment. Inter-rater reliability between raters JL and JP was 0.94, 0.99, and 0.97 for mean, SD and COV of BMD, respectively.

All parameters examined in the current study were not significantly different between the different mini-implant diameter groups (p>0.275) (Table 1). An amount of the migration (d_{max}) of the mini-implant, which combined elastic displacement (d_{l}) and viscoelastic creep (C_{l}), was detected in the loading direction for all specimens (Fig. 3). The displacement was not fully recovered after the mini-implant was unloaded leaving the residual displacement also for all specimens (d_{res}).

The bone mineral density (BMD) variability (SD and COV) did not have significant positive correlations with creep and residual displacement (Table 2). Neither the cortical thickness nor mean BMD significant correlations with migration (Fig. 4). There was no correlation between the static elastic displacement and creep (p>0.310). However, the static elastic displacement had positive correlations with the maximum displacement and residual displacement (p<0.015). Creep had strong positive correlations
with the maximum displacement and residual displacement (p<0.001) (Fig. 5). The maximum displacement had a strong positive correlation with residual displacement (p<0.001). The significant correlations were independent of the mini-implant diameters (ANCOVA, p>0.11).
DISCUSSION

These experimental conditions allowed investigation of the exclusive effects of mini-implant diameters on the early mechanical response of mini-implants to orthodontic loading. Primary migrations combining the static elastic displacement and the time-dependent viscoelastic creep were detected for all the mini-implants under the functional orthodontic loading level. These migrations occurred independent of mini-implant diameters. The traditional parameters of bone thickness and mean BMD, which were thought to have an effect on the mechanical initial stability of mini-implant system did not explain the primary migration. However, the variability (SD and COV) of BMD had significant correlations with creep suggesting that bone quality rather than bone quantity plays a role in determining the viscoelastic mechanical behavior of mini-implant systems regardless of the mini-implant diameters.

Most of studies to examine the mechanical stability of mini-implant systems measured static pull-out strength of mini-implants in bone specimen (Shah, Behrents, Kim, Kyung, & Buschang, 2011) (Lemieux, et al., 2011) (Wang, Zhao, Xue, Song, Deng, & Yang, 2010). Forces of greater than 200 N were applied to assess the pull-out strength. However, this amount and direction of pull-out loading could not represent the normal orthodontic loading in the tangential direction at the small functional level of 2N. In addition, the bone-implant interface was destroyed after static pull-out testing making it impossible to
investigate the time-dependent changes of mini-implant stability under constant loading following insertion. The constant tangential loading conditions used in the currently study could provide more practical information for the mechanical behavior of mini-implant systems at the immediate stage in implantation in a clinical setting.

Placement of the mini-implant at an oral bone location with high bone mass (bone thickness and BMD) has been strongly recommended to obtain better initial mechanical stability and long-term success of the mini-implant system (Cha, Kil, Yoon, & Hwang, Miniscrew stability evaluated with computerized tomography scanning, 2010) (Shah, Behrents, Kim, Kyung, & Buschang, 2011) (Lemieux, et al., 2011) (Brettin, et al., 2008). It was frequently indicated that an increase in diameter and length of mini-implant helped enhance the stability of mini-implant system (Duaibis, Kusnoto, Natarajan, Zhao, & Evans, 2012) (Liu, Chang, Wong, & Liu, 2012). Longer mini-implants with larger diameter had greater bone-implant contact area enhancing stability against forces of applied loading. In the current study, as the same level of tangential loading (2N) applied to all mini-implant systems, the static displacement accounted for their primary stability. As a result, we found no significant effects of mini-implant diameters on the stability of mini-implant system. This lack of significant effects may have resulted from the substantial thickness (average 2.15 mm) of dense cortical bone (average 1905 mg/cm³) in the specimens used in the current study, which likely provided strong stability of the mini-implant systems reducing the effect from different diameters. Studies using less bone mass may find more diameter effects on mini-implant stability.

Micromotion at a bone-implant interface has been considered as one of the most important factors in determining the primary success of the implant system (Melsen &
It was indicated that excessive micromotion increases the risk of fibrous tissue development at the implant interface, which can lead to failure of the implant system (Wazen, Currey, Guo, Brunski, Helms, & Nanci, 2013) (Szmukler-Moncler, Salama, Reingewirtz, & Dubruille, 1998) (Lioubavina-Hack, Lang, & Karring, 2006). In particular, the micromotion resulting from immediate loading prior to bone healing can interrupt bone regeneration at the bone-implant interface. The current finding demonstrates that mini-implant migration could develop by the immediate tangential loading at the level used in orthodontic clinical practice. This migration likely produces a gap between bone and implant surface at the opposite side of loading direction. The discontinuity from this interfacial gap may delay bone apposition to the surface of mini-implant and cause more complications including fibrous development and infection.

The current study observed that the migration continued under the prolonged constant post-implantation loading. This time-dependent viscoelastic creep was responsible for the permanent deformation of the interfacial bone with strong positive correlations between the creep and residual displacement independent of mini-implant diameters. Many studies have observed the creep behavior of bone tissue under a long-term constant physiological loading (Kim, Navalgund, Tee, Noble, Hart, & Lee, 2012) (Kim, Shertok, Ching Tee, & Yeni, 2011) (George & Vashishth, 2005). It also has been observed that the magnitude of creep was controlled by variability of tissue mineral density (Kim, Navalgund, Tee, Noble, Hart, & Lee, 2012) (Kim, Shertok, Ching Tee, & Yeni, 2011). The current finding was consistent with those results since there was a positive correlations between the variability (SD and COV) of BMD with the creep and residual deformation, but the magnitude of the relationship was low and minimally
explained the variability. Nevertheless, the variability of bone mineralization increased by active bone remodeling may produce more new (less mineralized) bone tissue following resorption of pre-existing (more mineralized) bone tissue (Kim, Navalgund, Tee, Noble, Hart, & Lee, 2012) (Yao, et al., 2007). It is well known that the active remodeling is involved in healing at the bone-implant interface (Melsen & Costa, 2000) (Garetto, Chen, Parr, & Roberts, 1995). Taken together, these observations suggested that the migration of mini-implant may continue under immediate loading during the process of healing after implantation. If the migration continuously progresses, it is possible that mini-implant may violate critical structures such as nerves and roots of adjacent teeth. This concern is aligned with the recommendation regarding 2 mm of clearance between critical structures and the location of mini-implant placement (Wang & Liou, 2008) (Liou, Pai, & Lin, 2004).

A limitation of the current study was that the mini-implant was placed after removal of the soft tissue on the human cadaveric mandibular bone. Some previous clinical studies indicated that the soft tissue type is associated with stability and success rates of mini-implant systems (Topouzeliz & Tsaousoglou, 2012) (Herman, Currier, & Miyake, 2006), while other case showed no correlation between survival rates and the surface characteristics including soft tissue (Chaddad, Ferreira, Geurs, & Reddy, 2008). More studies need to clarify the soft tissue effects on stability of mini-implant. Another limitation was that we performed in vitro experiments with no biological activities during healing at the bone-implant interface. Thus, the current findings were limited to the primary stability and migration of mini-implant systems under the immediate loading after implantation.
In conclusion, the mini-implant diameters did not have an effect on determining the primary stability and migration of mini-implant in the thick and dense human mandibular cortical bone. Time-dependent viscoelastic creep migration continued following the static displacement at the small level of orthodontic constant loading suggesting that more complications may be produced due to the micromotion at the immediate loading after implantation. The viscoelastic creep of mini-implant under the orthodontic functional constant loading demonstrated limited associated with variability of BMD surrounding the implant site, which is likely due to active remodeling during the post-implantation healing period. Thus, further systematic studies need to examine the effect of biological activities on long-term migration of mini-implant system.
Works Cited


