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VALIDATION OF AN ARTIFICIAL TOOTH-PERIODONTAL LIGAMENT-BONE COMPLEX FOR IN-VITRO ORTHODONTIC RESEARCH

For the degree of Master of Science in Mechanical Engineering

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VALIDATION OF AN ARTIFICIAL TOOTH-PERIODONTAL LIGAMENT-BONE COMPLEX FOR IN-VITRO ORTHODONTIC RESEARCH

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ABSTRACT

Favor, Trevor E. M.S.M.E., Purdue University, August 2015. Validation of an Artificial Tooth-Periodontal Ligament-Bone Complex for In-Vitro Orthodontic Research. Major Professor: Jie Chen.

Orthodontics research investigates the methods in which tooth displacement may be directed in the tooth-periodontal ligament-bone-complex. In the biological environment, the periodontal ligament is the soft tissue responsible for the absorption of forces on teeth and has a direct connection to tooth mobility. Current research is limited in that it must be conducted in an in-vivo capacity. A major advancement in orthodontics research would be a testing method that allows for the development and analysis of orthodontic devices without a patient present.

This study outlines the development and testing methods for the validation of an artificial periodontal ligament to be used in conjunction with an artificial-tooth-periodontal ligament-bone-complex. The study focused on finding the criteria in which consistent results were produced, the mixture that best simulated the human periodontal ligaments mechanical behavior, and the robustness of the artificial-periodontal ligament-bone-complex.

This study utilized a geometrically accurate denture mold filled with varying compositions of an artificial periodontal ligament for testing. Experiments focused on findings of viscoelasticity, curing times, and instantaneous responses of the teeth under direct orthodontic loading, as well as the changes in response from different teeth within the denture mold. Tests confirmed that a mixture composed of 50% Gasket Sealant No. 2 and 50% RTV 587 Silicone produced a substance that could adequately serve as an artificial periodontal ligament.

1. INTRODUCTION

1.1 Significance

Orthodontic load triggers tooth movement. Orthodontic treatment strategies are implemented by control of the orthodontic load system. Thus, quantification of the load system is important, but is challenging. In-vivo measurement of the orthodontic load system is not practical, hence most measurements are done in-vitro. Accuracy requires that the measurements are done under the same boundary condition as it is for in-vivo studies. Therefore, it is imperative to build a testing model that simulates the clinical conditions.

A tooth moves when an orthodontic force is applied. The amount of force sensed by the tooth depends on the initial crown displacement. This is critical especially for orthodontic treatments, such as sliding mechanics [1]. In these cases, the initial crown displacements are different, but their relative positions affect the orthodontic load system of each tooth. Therefore, the ability to simulate the crown initial displacement is critical for in-vitro measurement of orthodontic load system.

The crowns initial displacement in response to a force depends on the tooths surrounding tissues. The periodontal ligament (PDL) is an intricate tissue composed of collagen, blood vessels, nerves, and fluid [2]. It is a soft tissue that connects teeth to the bone socket and allows for the transmission of forces on the teeth to the surrounding bone [3,4]. The structure of the PDL is a complex composition of various collagen fibers, cellular debris, and fluids [2–9]. The PDL's ability to absorb forces such as those that occur during chewing is due to a combination of tension in the principal fibers and compression of the fluids. The hydrodynamics of the fluids affect the elasticity of the tissue [3]. Biomechanical analyses of viscoelastic responses such as hysteresis, creep, and stressrelaxation of the PDL are often conducted to observe the tissues' role in tooth support.

The PDL dominantly affects the crown displacement yet there is limited understanding as to what this effect may fully hold. Consensus exists that the PDL's nonlinearly elastic and viscous properties are the dominant factors on the tissues' mechanical responses. Some studies have been able to quantify the necessary orthodontic load systems on teeth that encourage tooth movement, despite its mechanical complexity [2, 10]. Orthodontic load systems on teeth heavily rely on the initial crown movement. In-vitro studies often lack a suitable PDL, limiting their findings on how these initial forces may affect crown movement [11]. Thus the ability to correctly simulate the crown displacement as well as other mechanical behaviors of the tooth-PDL-bone complex (TPBC) is critical for in vitro measurement of the orthodontic loads on teeth.

1.2 Gaps

Quantification of the orthodontic load system on the tooth is necessary for the optimization of tooth movement and to ensure the desired results during an orthodontic treatment are achieved. Because in-vivo force measurement is not possible and the effects of the periodontal ligament (PDL) need to be considered when testing, an artificial tooth- PDL-bone-complex (ATPBC) is needed for in-vitro orthodontic force measurements. Ideally, an ATPBC should be able to exhibit the major behaviors of a tooth-PDL-bone complex (TPBC), including the force-displacement relationship, stress relaxation, creep, and hysteresis so that the orthodontic load experienced by the tooth can be measured reliably. When orthodontic load is concerned, the PDL is one of the most influential factors that affect tooth displacement and the aforementioned behaviors, extenuating its need to be taken into consideration. An ATPBC that exhibits clinical crown displacement as well as similar viscoelastic behaviors of a

PDL has not been reliably established. The purpose of this study was to investigate suitable materials that can replace the PDL in in-vitro experiments.

The goal of the project was to develop a method that can simulate clinical cases when measuring orthodontic load systems. This paper details the methods that set out to validate previous findings from a study conducted by Xia [11] of an artificial periodontal ligament (APDL) composed of a silicon-sealant mixture so as to establish a proper testing method for the measuring of orthodontic loads in simulated clinical cases. The mixture was composed of two materials, gasket sealant No. 2 (GS) and RTV 587 silicone (Si). The previous study used a simplified ATPBC. The geometric effects were not assessed. This investigation was conducted primarily on a dental mold with real geometry. The focus was on analyzing the mixture's ability to simulate orthodontic displacement as it is used with the denture mold. Both incisor and canine were tested in order to evaluate the effects of tooth geometry on the orthodontic load system. The ATPBC system would be validated by the clinical results published previously [2]. The goal can be achieved by attaining the following four objectives.

1.3 Objectives

1.3.1 Objective 1: Replication of Findings of Viscoelasticity

The first objective of this study was to determine that the viscoelasticity of the materials being used is preserved as they were in the previous study by Xia [11] when use in conjunction with an anatomically correct denture mold. The properties observed that pertain to viscoelasticity are force-displacement, stress-relaxation, creep, and hysteresis. The viscoelasticity of the PDL is instrumental in the biological environment for tooth movement; therefore these viscoelastic properties should also be evident in-vitro.

1.3.2 Objective 2: Steady State

The second objective was to establish the criteria that would produce the best results, which is the factors in which consistency of results is achieved. Consistency throughout testing is of upmost importance for if data is not consistent, results during orthodontic tests will be skewed, unreliable, and irreproducible, rendering it useless for prolonged study. Consistent results can be achieved by ensuring a proper testing set up that is easily reproducible. Emphasis on consistency has been the focus of in-vitro investigations that have sought to develop new testing methods for the quantification of the behaviors of the PDL [10].

Building off this idea, a proper protocol that aids in reproducibility would also be beneficial and prevent minor changes that can hinder the reliability of data. A set of guidelines from start to finish will minimize the chances of noise that may further cause skewed and unreliable results. The protocol must be as simple as possible to reduce the chances of error between steps. It will also allow for future parties to conduct continuing experiments.

1.3.3 Objective 3: Accuracy

The third objective is determining which composition of mixture of the materials is best for testing. The mixture of materials must result in crown displacement that agrees to that measured clinically. Altering the composition of the mixture may affect the tooth response from orthodontic forces; therefore finding the correct mixture that produces the clinical displacement is paramount.

1.3.4 Objective 4: Limits of Application

The final objective was to confirm that the chosen material could be used for the study of different teeth. Of major concern was the limit in which the APDL may be applicable. Ideally, it should be expandable to be used in all teeth, thus multiple teeth must be tested to see if findings are reproducible despite the possible differences geometries. This will speak to the robustness of the materials and ensure that the composition will serve as an APDL capable of simulating the biological environment.

1.3.5 Hypothesis

An in-vitro system that can simulate orthodontic treatments will serve as an ideal method of testing orthodontic devices without the use of living subjects. Therefore, it is hypothesized that the APDL in an ATPBC results in the same crown load-displacement as that in clinical studies.

2. REVIEW OF LITERATURE

The PDL displays both elastic and viscous characteristics; hence it is regarded as a viscoelastic material. Agreement exists that the mechanical response of the PDL is time-dependent [5, 8, 12], which is attributed to the elastic collagen fibers and the fluid composition of the tissue. In addition to the time-dependent behavior, a non-linear stress-strain response, a common feature of soft tissues, has also been reported [3, 5, 7, 9, 13–16].

Previous investigations into the PDL's material properties used an isotropic material model whose elastic properties could be considered linear [2]. This view allows for assumptions and general calculations to be made that simplified studies in which numerical calculations were the basis of analysis. More recent studies have investigated the PDL from an anisotropic view due to the recognition of the important role of the collagen fibers suspended in the PDL's matrix have on force propagation [7]. Most recently studies have adopted using viscoelastic behaviors for the analysis of the PDL tissue [2, 5, 9, 11, 14, 17].

A 2014 experimental study by Papadopoulou concluded that the viscoelastic response of the PDL is a result of the combination of the elastic structure and the fluid that lies within the tissues [14]. The properties of tissues are non-linear and time dependent. Loading history can affect the mechanical behavior of PDL tissues. For instance, prolonged held displacement leads to remodeling of the alveolar bone, causing tooth migration. This biological adaptation to applied loads is referred to as mechanotransduction [18].

Papadopoulou et al. investigated the time-dependent behavior [14]. To do this, teeth from pigs were loaded to different displacements under controlled velocities. Displacements were a combination of PDL deformation, tooth displacement, bone deformation, and deformation of the encasing resins. Maximum force was found to vary along with the rate of applied loads.

It was further observed that removal of applied force allowed the tooth to return to its original position [14]. Small forces initiate the initial tooth mobility and may be attributed to tipping. The level of tipping depends center of resistance, which is the location on the tooth that will result in tooth translation when forces pass through it. The center of resistance has been previously found to be approximately one quarter of the distance of the entire tooth relative to the tooth's root [16]. Poppe et al. located the center of resistance to be 0.42 and 0.43 of the proportion of alveolar height from conical to apical for the incisor and canine respectively [19].

Past research has shown a disagreement in the PDL's mechanical properties [9] which has been attributed to inconsistent testing protocols [20]. A large variation in reported results is a direct result of a lack of standardized protocols for testing. Researchers have collected data using a myriad of methods and based on the experiments conducted on a wide berth of subjects [3, 9]. The way testing subjects were handled may also have an adverse effect on results. Ex-vivo cases in which tissue samples were frozen may be harmful to collagen [21]. Since tooth movement is greatly dependent on tooth geometry there is also some degree of variability of responses between individuals [2, 3, 9].

The properties of the PDL differ among the teeth since collagen fiber structure and orientation change depending on location [4]. Properties may also change depending on the species of the subjects tested. Responses to forces vary between individuals, teeth tested, the type of movement, and tooth geometry [6, 22]. In summary, behavior may differ between teeth unpredictably.

The variations in findings in mechanical response may be the result of unique biological factors differing between individual patients. Factors can include occlusion, systemic metabolism, age, and bone structure [9], all of which makes it difficult to create a universal model for PDL behavior. The viscoelastic nature of the PDL also leads to variable results since the properties will vary depending on the mode of loading [9]. Factors that may influence orthodontic movement are genetic predisposition, the types of treatment, plaque, gum disease, smoking, drug history, and lifestyle to name a few [23].

The mechanisms which lead to controlled tooth movement are still not fully understood. A general understanding exists that a force applied to a tooths crown can cause an uneven distribution of stress and lead to tooth tipping [20]. Extraneous movements lead to inconsistent and inaccurate data. Most studies lack acknowledgement of the multiple phases of orthodontic tooth movement which is dependent on the force and the rate at which the force is applied [20]. Large variations in the reported rates of orthodontic tooth movement occurs between individuals even when standardized and consistent testing methods were employed [9, 20, 24].

The distribution of force, stress and strain propagation through the PDL, and strains in the surrounding bone socket are critical factors that trigger mechanotransduction and influence remodeling [20]. When forces cause the tooth to interact with the alveolar bone, root resorption occurs [25] which encourages the breakdown of the tooth's root structure. This is a complication that may arise from orthodontic treatment and could lead to tooth loss if not treated [26].

Tooth mobility allows for the observation of tooth movement in the periodontium. Proper and standard locations of force application are necessary to establish tooth displacements as opposed to tipping [13]. In tooth displacement, force causes an increase in PDL space, allowing for the tooth to encroach on the compressed side. The sites where compression of tooth roots and alveolar bone occurs is the portion where mechanotransduction is activated, leading to bone resorption [9, 15].

The major concern in orthodontic research is how the tooth and its surrounding tissues respond to forces applied from orthodontic devices. The PDL is directly related to tooth movement. To understand a tooths mechanical behavior, the mechanical properties of the PDL must be understood [9].

An ideal force level could result in maximum tooth displacement without damage to the tissue or surrounding bone [16, 20]. This ideal force, the magnitude of which may be specific to the individual patient, would produce minimum discomfort while providing the most efficient crown displacement. Previously it has been believed that higher stresses can lead to faster tooth movement if constraint conditions are established to prevent rotation of the tooth [16], however higher stresses may lead to higher strains in the elastic components of the PDL [27], indicating that it is critical to fully understand the orthodontic load system experienced by the tooth for optimal treatments.

It has been reported that a minimum of four to eight hours of force application is necessary for orthodontic tooth movement [9]. The optimal movement occurs when continuous force is applied. Response of the PDL is best characterized as an instantaneous displacement followed by a lag phase. It is believed that the fluid within the tissue is a key feature in dampening from occlusal loads [9, 14].

Qian and researchers observed the non-linear and time-dependent characteristics of the PDL in an analytical study [4]. The collected data was compared to a finite element model to test if the responses could be calculated and predicted. Digital images of the surface of tested samples during mechanical testing were taken. Samples were blocks of PDL, alveolar bone, and gingiva harvested from pig mandibles. They were able to observe that high strain mostly occurs in the PDL and stress in the tooth and alveolar bone. Small displacements led to localized strain purely in the PDL, around the roots of teeth. Larger forces applied on dental crowns resulted in larger strain in the PDL. High displacements show some strain in the surrounding tissues, but the PDL always displays higher values. Qian and researchers highlighted that geometry and individual's variation in material properties lead to different deformations.

Qian was able to approximate the time-dependent characteristics of the PDL using a generalized Maxwell model. Models could only be called assumptions due to the shortcomings in being able to observe continuous change of the elastic modulus and the displacement fields in the entire tooth-PDL-bone-complex. The viscoelastic behavior of the PDL is responsible for tooth displacement and the deformation throughout the TPBC. Researchers also concluded that the fluid and collagen fibers within the PDL tissue provide the viscoelastic behaviors.

Sanctuary and researchers have also conducted an experimental study to monitor the time-dependent behavior of the PDL [5]. A microtensile machine monitored crown response from specimens produced from bovine teeth applied ramp tests and a sinusoidal strain. Researchers concluded that all specimens needed to be preconditioned. Preconditioning prior to data collection allows for results to be more reproducible by relaxing soft tissues from their naturally stressed state [28]. One may think of preconditioning as being similar to a tooth's increased mobility as a result of orthodontic tooth movement. A study by Drolshagen found a noticeable difference in maximum force applyable as a result of preconditioning [2].

A main goal of the study by Sanctuary was to determine the PDL's response as caused by the tissues' fluids. A major assumption was made that collagen fibers uncoil under initial deformation and that a majority of the stress response is from the fluid. Stress and strain was calculated by collecting load and displacement data. The subsequent results were plotted to create stress-strain diagrams. These plots proved the PDL's response was non-linear. A stiffening of the tissue was evident at higher loading velocities. This is an expected property of viscoelastic tissues and has been reported by other researchers [27, 29].

Sanctuary and researchers also concluded that age contributes to variabilities in results. The samples from older subjects have historically displayed a lower stiffness and a reduced relaxation rate. This corresponds to findings by Tanne et al. [6]. Komatsu has also conducted studies into how the effects of age contribute to the properties of the tissue and concluded that an increase in tissue stiffness is evident in tissues from older subjects [17]. These researchers also investigated the viscoelastic properties of the tissue.

Komatsu detected a distinct change in tissue stiffness between rats of different ages [17]. Shear stress in the rats decreased by 36% and stiffness decreased by 54% with increasing age. Contrarily shear strain, the extensibility of the tissue, increased by 59%. The observed changes between subjects ages were hypothesized to arise from changes in geometry. The geometry of the alveolar bone and the biological characteristics of the tissue can result in deviations of the initial force values [6, 14]. The fiber components may experience changes in mechanical properties. Komatsu reported a decrease in stress that may be related to decreases in fluid flow and macromolecules in the collagen fibers as a result of increasing age.

Komatsu used failure energy as a measure of the tissues' ability to withstand forces. It was calculated as the area under the stress-strain curve and was found to be invariable between 2 and 24 month old subjects. This was attributed to the increase in maximum shear strain.

Changes related to age may signify changes in PDL width [17]. As age increases, the density of the alveolar bone and the cementum thickness change. An increase in stiffness is evident in older individuals [6, 7]. PDL tissues in older humans show a loss of collagen fibers and a decrease in cellular tissues. Areas containing blood vessels increase in size of the surrounding areas. Additionally, different states of dental development due to age would result in varying responses [14, 25].

In-vivo studies have applied forces on the middle of the distal surface of teeth [24]. Small forces in orthodontic research are more reminiscent of forces that occur periodically in the native environment [20]. Small forces also are less painful to the patient. Higher forces compress the PDL. Prolonged force exposure leads to tooth displacement and remodeling in the alveolar bone. To observe the response of the PDL, quick displacements and low forces allow for the instantaneous response readings [6, 14, 22]. This may lead to an overestimate in material stiffness from the buildup of fluid pressure [29].

A number of recent studies have set out to better determine PDL response in vivo. Jones and researchers set out to validate a 3D finite element model using data collected from human tooth movement [24]. Their desire was to have an accurate model that displays the behavior of teeth and accompanying tissues in response to orthodontic loads. Researchers confirmed the initial elastic response and the viscoelastic behavior of the tissues. The length of the various phases differed between individuals. In order to validate their model, researchers first found the initial responses of teeth of human subjects exposed to direct load. This simultaneously established the material properties of the PDL.

Ages of subjects in the Jones study ranged from 24 to 36 years of age. All volunteers displayed healthy oral tissues and need to be free of “tight contact” between teeth. Variations in displacements were evident between subjects as well as between readings of individual’s displacement data. The level of variation that greatly differed between subjects can be attributed to a host of factors such as, but not limited to, age and prior periodontal disease.

It was concluded that in-vivo testing showed variation between subjects and the degree of such variation varied. The PDL displays an elastic response initially, but a viscoelastic response when exposed to continuous load. The developed FEM model proved that strain was localized within the PDL, signifying its importance in tooth response to load and tooth movement.

The purpose of the Goellner study was to evaluate tooth displacement under horizontal loading [10]. Their goal was also to prove the reproducibility of experiments conducted on subjects. The study utilized photogrammetric techniques to monitor tooth movement, which was highly advantageous due to the non-destructive and less intrusive capabilities of these techniques to provide optical measurements. Tests were conducted on the left and right central incisor, the lateral incisor, and canine tooth of the maxillary jaw.

The experimental set up was composed of two CCD cameras, a custom made loading device with load cell, a signal amplifier, and an A/D converter. 23 subjects, who were determined periodontally healthy, ranged in gender and age. 9 men ranging in age from 22-29 and 14 women ranging in age from 22-28 participated in the study. Photographs were taken using high-resolution digital cameras oriented at 30 deg and the loads were applied to the subjects in 3 N intervals from 0 N to 18 N, allowing for the production of 3D images of the teeth during loading. Unloaded teeth were used as a reference.

The focus of the Goellner study was proving that their experimental setup could perform accurately and provide statistically significant and reproducible results. Central and lateral incisors were shown to have higher displacement values than on canines. Outliers of the data were believed to be caused by higher individual tooth mobility in the subjects. Intraclass correlation proved that results were highly significant and speaks to the stability of the experimental setup. The relevance of this study to tooth mobility is derived from the range of displacements found for each tooth and the reproducibility of the obtained results. This article does not delve into the viscoelastic properties of the human PDL, but still serves as a prime and accurate measurement of force and the resulting displacements.

Drolshagen et al. conducted one of the more recent investigations into the force-displacement relationship of the human PDL in the in vivo case [2]. Researchers determined the thickness of the PDL tissue to be 0.2 mm. They set this thickness as the maximum displacement that was applied. Their device was able to record displacement and resulting forces while using loading velocities from 0.05 mm/s up to 6mm/s.

Force-displacement curves were established by obtaining the resultant forces from applied displacements of 0.05 and 0.1 mm. Experiments were conducted on the maxillary incisors of volunteers. Displacements were applied within 0.1s and held for 1 s. For a displacement of 0.05 mm, force ranged from 5 N to 5.6 N, specifically in descending order. The force decreased slightly after reaching its maximum force reading. Researchers attributed this to the hydrodynamics of the PDL tissue.

The previous study conducted by Xia and Chen [11] analyzed the responses of an ATPBC composed of the same silicon-sealant mixture. The responses investigated include load-displacement, stress-relaxation, creep, and hysteresis. These properties are the major behaviors that all viscoelastic tissues express. While the results were compared to other in vivo studies, some of the studies analyzed were not from investigations of human PDL, but rather that of other animals, which may limit the applicability of these findings to orthodontic implications. The ATPBC tested was simplified, which did not represent the tooth geometry.

The APDL used in these experiments were composed of gasket sealant No. 2 and RTV 587 silicone. These two were used to create a flexible film similar to the soft PDL tissue. Forces applied laid within the common orthodontic force range.

The study by Xia [11] was conducted on a simplified TPBC and found that the ideal composition for an APDL in their tested ATPBC was a 50% gasket sealant and 50% RTV silicon. The results of the study were comparable to a human PDL using observed crown displacement.

The study reported that at a force of approximately 4 N, the displacement matched results from other human tooth displacement studies. This is however outside the normal range of orthodontic loading which normally maxes out at 3 N. The remaining viscoelastic properties were also determined to be comparable to the biological tissue. An issue with these comparisons is that they relate to other animals such as dogs and rabbits.

It has been determined from other studies that tooth geometry plays a role in mechanical behavior [2–4, 6, 9, 30], signifying a need of further investigations into the findings by Xia. Previous studies rely on biological tissues for testing. In-vitro tests too often employ the use of frozen tissues from various animals [3, 8, 9, 14]. In-vivo studies require willing participants or animals that do not share the geometry that humans display [2, 10]. Orthodontics research would benefit from having an in-vitro testing method that can simulate clinical crown displacement. This signifies the need for the previously developed APDL materials responses to be tested in an environment in which similar biological geometry is used. Using the reported displacement data, the mobility of artificial teeth using viscous and elastic materials to simulate a periodontal ligament could be examined and then be compared to that of human displacement data for their relative displacements at various levels of force. Utilizing the findings of various biological studies of the PDL, a better testing method must be established that can recreate similar testing conditions as those in clinical studies.

3. METHODS

The purpose of this study was to establish that an anatomically correct ATPBC is able to simulate human crown displacement, thus aiding in the in vitro quantification of orthodontic load systems as a result of different treatment strategies, including sliding mechanics. Through the use of an experimental testing set up, the properties of the ATPBC could be observed. A number of steps were taken to ensure an accurate set up was created that was capable of producing reliable results. The properties and data collected may then be compared to those found in previously reported clinical studies.

The Drolshagen study [2] was the primary study selected for validation due to the use of a device that utilize a similar method of force application as was created for this study, the major difference being that their device was automated and able to instantaneously impose a force.

3.1 Experimental Setup

3.1.1 Denture Mold

The subject of these in-vitro tests was a premade mandible denture mold. The mold has the dental arch with the teeth. The bony structure and teeth were segmented from human cone beam computed tomography images. Their solid models were created and saved as .stl file. The molds were made through rapid prototyping. The PDL was not included, which constitutes a space between the root and alveolar bone. The space is about 0.3 mm wide, which is slightly wider than the PDL thickness reported previously [2]. The space would be filled with the APDL. The geometry of the mold was anatomically correct. The incisor, lateral incisors, canines, and premolars were prototyped separately and could be easily inserted into their corresponding sockets.

3.1.2 Testing Base

Prior to testing, an appropriate testing apparatus needed to be constructed. The apparatus needed to apply a force to the crown of a tooth and measure its displacement reliably. A custom made device consisting of a load cell and a micrometer was used to observe the force-displacement behaviors of the crown, which was governed by the APDL and denture mold. The device needed to be able to hold both the micrometer and load cell as well as hold the denture mold. The micrometer was used to apply displacement while the load cell was used to record the resultant forces. The apparatus needed to be sturdy and remain still throughout testing. Unneeded movement would easily influence the force readings and provide inaccurate results. The most important feature was making sure that each tooth in the denture mold had the ability to be tested individually.

3.1.3 Testing Apparatus

The design of the testing base consisted of a testing plate in which the denture mold could be set upon as shown in Figure 3.1. The mold was secured to the testing plate to prevent its moving during testing. A micrometer was used to control the displacement and a load cell (ATI Nano16) was used to measure the load. The load cell was mounted in line with the micrometers arm. A probe extruded from the load cell would contact the crown surface and apply the load.

Of major concern was insuring that the pieces that constituted the base were not in fact loose and remained flush. Lose parts would create unwanted moments and extraneous movements when forces were applied. The micrometer-load cell arm was able to maneuver around the denture mold, ensuring that each tooth could be tested.

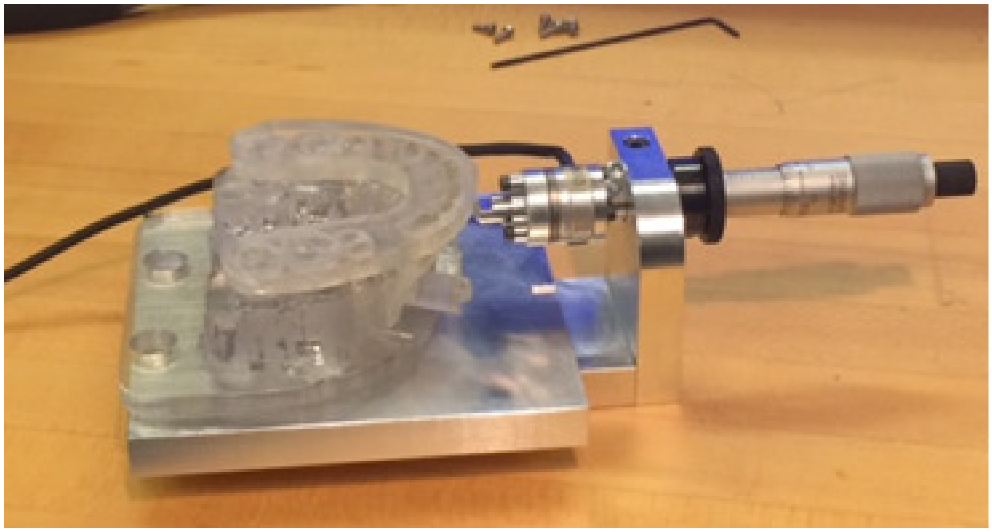

Figure 3.1. Testing Base

The probe from the load cell was designed to touch the midpoint of the labial surface of the teeth as shown in Figure 3.2. Applying a point load allowed the force to be accurately applied, creating pure labio-lingual load on the tooth crown, similar to the methods in which Drolshagen and Jones generated force. Contact needed to be maintained between the tip of the load cell and the tooth being tested.

Figure 3.2. Location of the load cell-tooth interface

Data was recorded by means of a load cell as shown in Figure 3.3. The load cell was a Nano 17 F/T (ATI Industrial Automation, Apex, NC). It was used to measure the resultant forces acting on the tested teeth. The accompanying ATI DAQ software allowed for monitoring of sensed forces as well as the option to record those forces. The load cell has a sensing range -17 to +17 N and a resolution of 7 x 10^{-4} N. The load cell was serially aligned to a non-rotating spindle of a micrometer and fastened by means of a screw. A custom made tip that could allow the application of directed force was attached to the load cell.

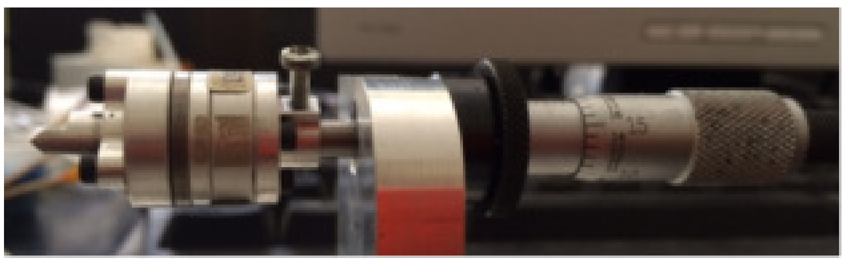

Figure 3.3. Micrometer with load cell and tip

The micrometer spindle (No. 261L, L S Starrett Company, Athol, Mass.) has an accuracy of +/- 0.00254 mm and an extendable length of 12.7 mm, shown in Figure 3.3. The micrometer allowed for the application of displacement upon the tested tooth while the load cell simultaneously measured the resulting force experienced by the crown. The micrometer was able to apply a controlled displacement to the load-cells tip, applying a measurable force to the crown of the ATPBC. Turning the micrometer head while recording the readings from the transducer established the force-displacement relationships.

Force was applied at the center of the tested teeth as measured from top of the tooths socket. This is roughly the same location used in previous in-vivo studies [2,24].

Prior to each trial, the tested tooth was loaded to 5 N for preconditioning. In order to determine the nominal displacement, the Zero position of each tested tooth was found. The micrometer was adjusted so the load cell barely touched the tooth, approximately reading a force of 0.1 N. The load cell was then biased to zero out the force readings. This initial position of micrometer was recorded and served as the zero position in which the recorded values were substituted from. The same method of achieving the initial position was followed before each experiment.

3.1.4 Materials

The APDL was a mixture of a viscous material, gasket sealant No. 2 (GS), and an elastic material, RTV 587 silicone (Si). These materials had previously been investigated for their ability to produce a viscoelastic mixture [11]. The combination of the two materials produced a viscoelastic material that was believed to simulate a human periodontal ligament. Adjusting the ratio of these materials allows for the viscosity to be controlled [11]. The percent mixtures were 30% gasket sealant and 70% silicone (30/70), 40% gasket sealant and 60% silicone (40/60), 50% gasket sealant and 50% silicone (50/50), 60% gasket sealant and 40% silicone (60/40), and 70% gasket sealant and 30% silicone (70/30).

3.1.5 Mixing

These mixtures were used between the artificial teeth and the walls of the sockets of the denture mold. Due to the viscosity of the materials, exact measurement of the volume mixed was difficult to discern. Instead, abundant mixtures were created that focused on the percent composition per volume of the two materials mixed. The compositions were mixed in a small and clear liquid measuring cup shown in Figure 3.4. This was chosen due in part to its small size and wide mouth, allowing for easy insertion and mixing of material. It also allowed for a more accurate estimate of the amount of materials being mixed. Later, in order to reduce materials used, small caps were favored. These caps were filled with the respective materials and then mixed in a larger clear cup.

After mixing, materials were wrapped around the root of the tooth to be tested as depicted in Figure 3.5. When the root was significantly covered, it was inserted into

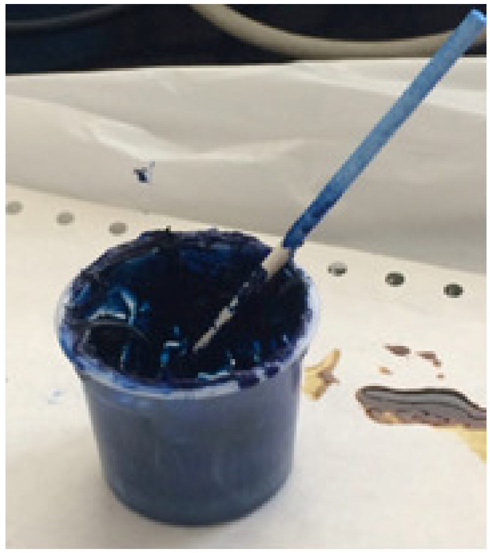

Figure 3.4. Mixing cup for gasket sealant and silicon

the socket of the denture mold. Excess material that extruded out from the socket was removed prior to testing.

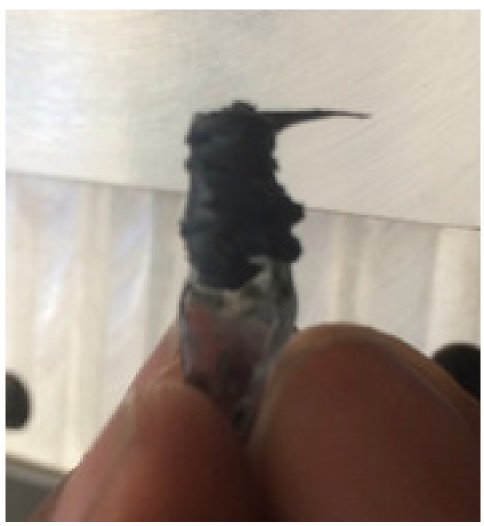

Figure 3.5. Tooth prepared for insertion into the denture mold

3.2 Goals and Methods - Viscoelasticity

The first experiments were to validate the viscoelasticity of the materials when used with the denture mold. Four viscoelastic properties were investigated. 1) Force-Displacement at the center of the labial surface of the crown; 2) Stress-Relaxation, holding the displacement constant and recording the dropping of the force; 3) Creep, holding the force and record the changing of the crown displacement; and 4) Hysteresis, loading and unloading while recording the crown displacements.

3.2.1 Experimental Design

To establish the force-displacement relationship, the initial position of micrometer was first recorded. Displacement was applied gradually and recorded at each 1 N reading up to 5 N. This was repeated for three trials for each mixture. 3 N is normally the highest force that may be experience in orthodontics; however a larger final force was selected to ensure data trends continued. Larger forces also aligned better to in-vivo studies [2, 10].

Stress-relaxation curves were found by first finding the initial position of micrometer recorded. This relationship investigated the tendency of the material to relax resulting in a lower reactionary force over time. A displacement was applied until load cell registered 5 N. This displacement was held and the force reading was recorded every 30 seconds for the first 5 minutes, and then again every five minutes until an additional 30 minutes elapsed. The exception to this was the 70/30 mixture, the first mixture tested, where it was established that readings every 5 minutes for thirty minutes resulted in extraneous data points. Three trials of the stress-relaxation were performed and averaged.

To find creep, like the previous properties, first the initial position of micrometer recorded. Additional displacement was then applied until a force of 5 N was reached. The 5 N force was held constant for 30 minutes by continuously adjusting displacement. Displacement readings were recorded every 30 seconds for the first 5 minutes, and then every five minutes until a total time of 30 min elapsed.

Hysteresis, the difference between loading and unloading curves, was the final property to be observed. The initial position of micrometer recorded as previously described. Displacement was then advanced to 0.25 mm past the starting point and then returned back to starting point at a constant rate of approximately 0.025 mm/s, allowing for recovery behavior to be monitored. Force readings were recorded at every 0.01 mm interval.

A change in stiffness of the extraneous materials was observed as the various tests were conducted. This provided concern that the mixtures' properties changed over time. Further testing was needed to observe the effects of time on the materials.

3.3 Goals and Methods - Time Study

The force-displacement properties found from the ATPBC should be validated by the values obtained in the clinical studies. As the mixtures were being created during the testing for viscoelastic properties, noticeable changes in the stiffness of the materials were observed. This observation was expanded to the realization that the materials may take time to settle and dry and thus require a curing time. The variability of each mixture over time became of concern. Building upon this, the time in which readings are taken may correspond to a specific time in which the more desired responses would occur. Of the five compositions, the properties found that best match the clinical studies and yielded the most consistent results would be selected as the ideal composition. This ideal composition could then be compared to the study by Xia [11], in which the APDL was first developed.

Tooth geometry is believed to play a factor in the response to orthodontic loading. This implies that results may vary from tooth to tooth. Two teeth were selected as the focus for testing, the central incisors and the canines. An assumption was made that the mechanical behavior of the canine and incisor are not side-specific since the left and right canines and the left and right incisors have similar geometry.

To observe the variability in the materials' properties, Force-Displacement readings were taken over time. Displacement was observed for the various mixtures over the course of 48 hours. The goal was to determine the times in which the mixtures yielded the most consistent results. These findings could then be applied to a testing protocol for future research.

A force-displacement curve was established by rotating the micrometer spindle to impose displacement until specific force readings were reached. This was achieved by gradual displacement, being careful that the desired forces were reached and able to be maintained for a minimum of a few seconds. Displacement was recorded at each 1 N reading, from 0 N (the starting position) up to 5 N. Data was collected every 30 minutes for 4 hours, which was the approximate time it took to complete the initial tests [11], and again 24 and 48 hours after the initial test. Four readings were taken at each time interval for each mixture. The four trials were averaged to find the mean displacements at each corresponding force level.

The standard deviation of the four trials at each time point was taken. This serves as a measurement of how consistent the data was throughout the trials. As is common with viscoelastic materials, the starting positions were expected to vary to some extent, and were observed to hold true. Starting positions before each trial were recorded establishing an absolute value for the change in displacement. Since the goal of this study is to find the material and condition in which the variability was lowest, the data that produced the lowest mean standard deviation would prove to be part of the ideal conditions.

When testing began, the APDL's viscoelastic response was of concern. As viscoelastic materials slowly return to their initial position, the teeth being tested displaced varying distances before the necessary forces could be achieved and displayed differing starting positions. The decision in question was whether or not it was appropriate to manually reset the tooth to its starting position. The materials at the beginning of testing tended to be wet and still lose. Resetting the tooth appeared to be necessary to ensure that it went back to a starting position. With this in mind, two studies were conducted, a Reset study where the tooth was manually reset to its starting position, and a Non-Reset where testing was conducted at whichever point the tooth displaced to at the end of each trial.

3.4 Goals and Method - Instantaneous Testing

The previously collected data for the assessment of viscoelastic properties was the direct result of gradual loading until the desired forces were achieved. The analysis of the change in final displacement and the observed shifting in starting positions indicated that this loading could in fact be too slow compared to the loading speeds used in in-vivo studies [2, 10]. The material was able to adjust and accommodate the incoming forces, which in turn expanded the displacement the tooth could undergo. The initial response under more rapid loading is needed to determine the instantaneous response of the APDL. This also better fits the clinical experiments conducted by Drolshagen in which displacement was applied quickly and the instantaneous force was recorded [2].

The goal of these tests was to determine the instantaneous force readings. The key idea being that force needs to be registered before the material has time to relax so that the complex can be validated using the clinical data.

Much like the previous experimental set up, APDL mixtures were prepared under the same concentrations. These concentrations were tested at two time points: within ten minutes of being mixed and again 48 hours after. Instead of applying displacement until a specific force was reached, set displacement values were used. These were 0.05mm and 0.1mm, the same displacement values from the previous study by Drolshagen [2]. Displacement velocity of about 0.1 mm/second was used with some error since this was done by hand. Loading and unloading of displacement was performed and resultant loads were found. The ATI DAQ software was set to constantly record as these displacements was applied. Between 600 and 700 data points were taken. Using MATLAB, the first 300 data points from the recorded files was plotted. Since time between readings differed due to the micrometer needing to be manually adjusted to make contact with the tested tooth, showing fewer data points allowed for the distinct impulse curves to be observed clearly. The maximum of the peak loads were found and compared with those from the aforementioned clinical study in order to asses which composition has the ability to produce the best match.

3.5 Goals and Method - Reproducibility

After establishing the proper testing conditions while testing the incisor, results should be able to be confirmed with different teeth. The canine also has a single root and is composed of different geometry from the incisor. It is expected that the force-displacement behavior will be similar and will be evaluated in this study.

The previous experiments were reproduced using the canine as the testing subject. Only the selected APDL mixture was used and the force-displacement behavior was tested. The viscoelastic properties were excluded since it was already established that the material compositions display them. Results from these sets of experiments were expected to yield similar findings to their incisor counterparts. These would confirm the previous conclusions while proving that the use of the material may be expanded to incorporate other teeth within the denture mold.

While the change in consistency over time was of concern, the overall change in displacement from the initial readings from the beginning of the viscoelastic property testing to the final readings at the 48-hour mark, were also of interest. Observation of the progression of the material properties over time revealed that 48 hours is prime for the materials to set. Inspection of the change in mean final displacement is necessary for the understanding of how tooth displacement is effected by the settling of materials.

The data from the time studies of both the incisor and canine were sufficient enough to be re-appropriated for further analysis. Using the displacement values at the initial time period and at the final time period from the previous tests, the percent change in displacement could be found. The percent difference between the initial and final mean displacements would then serve as a measurement of how much the materials response to varying levels of force changed over time shown in Equation 3.1. In other words, the goal was to determine how much the displacements of the four trials for each force level, on average, change from when the material was first created as compared to the end of the 48-hour period. The equation used was:

$$Error = \frac{FinalDisplacement - InitialDisplacement}{InitialDisplacement} \times 100\% \tag{3.1}$$

The goal of this study is to fully explore how the changes in mechanical properties of the materials changed nominally from start to finish. This can be achieved using the obtained force-displacement data for each material. As it was determined, which will be described later, that the Non-Reset method was the best choice for data acquisition, the data obtained from those experiments were used. The mean standard deviation of the initial readings, the 0 minute mark, and the final readings, after 48 hours, could then be compared in order to assess which composition displays the most consistent values across the four trials.

4. RESULTS

4.1 Mechanical Properties - Viscoelasticity

The viscoelastic properties were investigated using the central incisor.

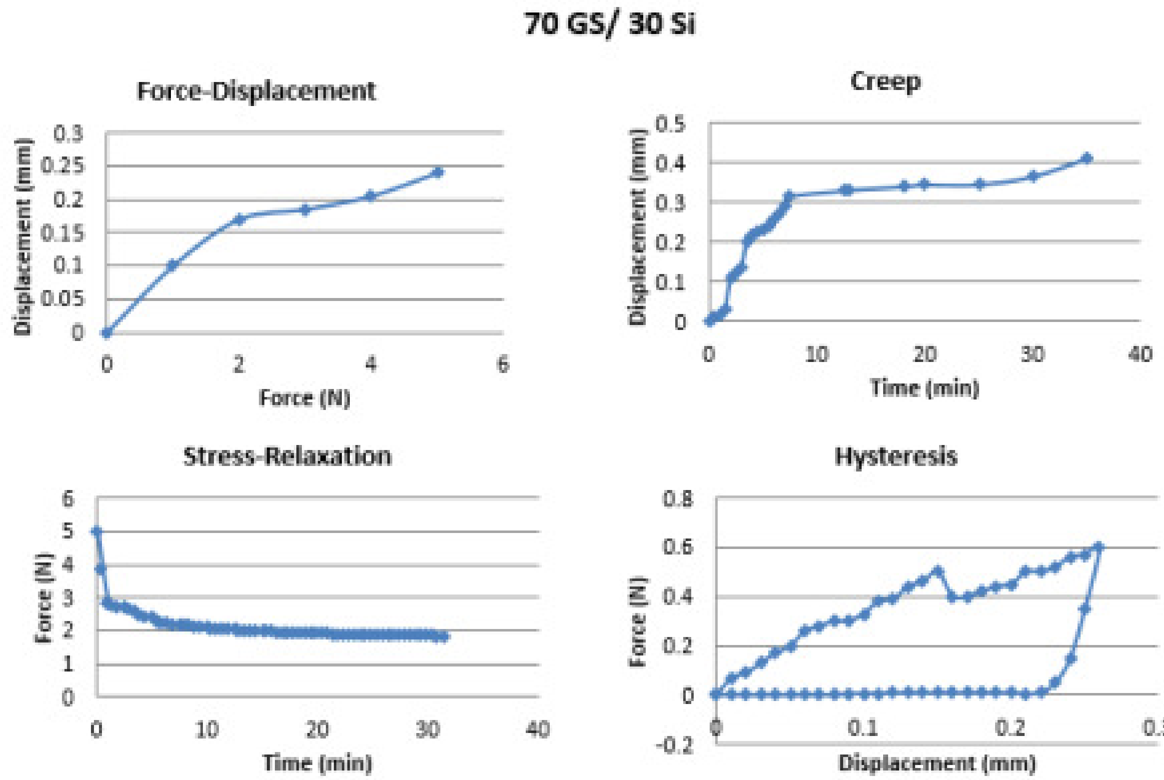

Figure 4.1. Plots of the viscoelastic properties of the 70/30 mixture

The 70/30 mixture displayed viscoelastic material properties as shown in Figure 4.1. A non-linear force displacement curve was found. Creep displayed a steep viscoelastic phase followed by a plateau. Stress-relaxation showed that within the first two or three minutes, the force droped to a minimal value, which continueed to decrease slightly as time progressed. Hysteresis, defined as the differences in loading and unloading, showed two distinct curves, reaching a maximum force of 0.9 N.

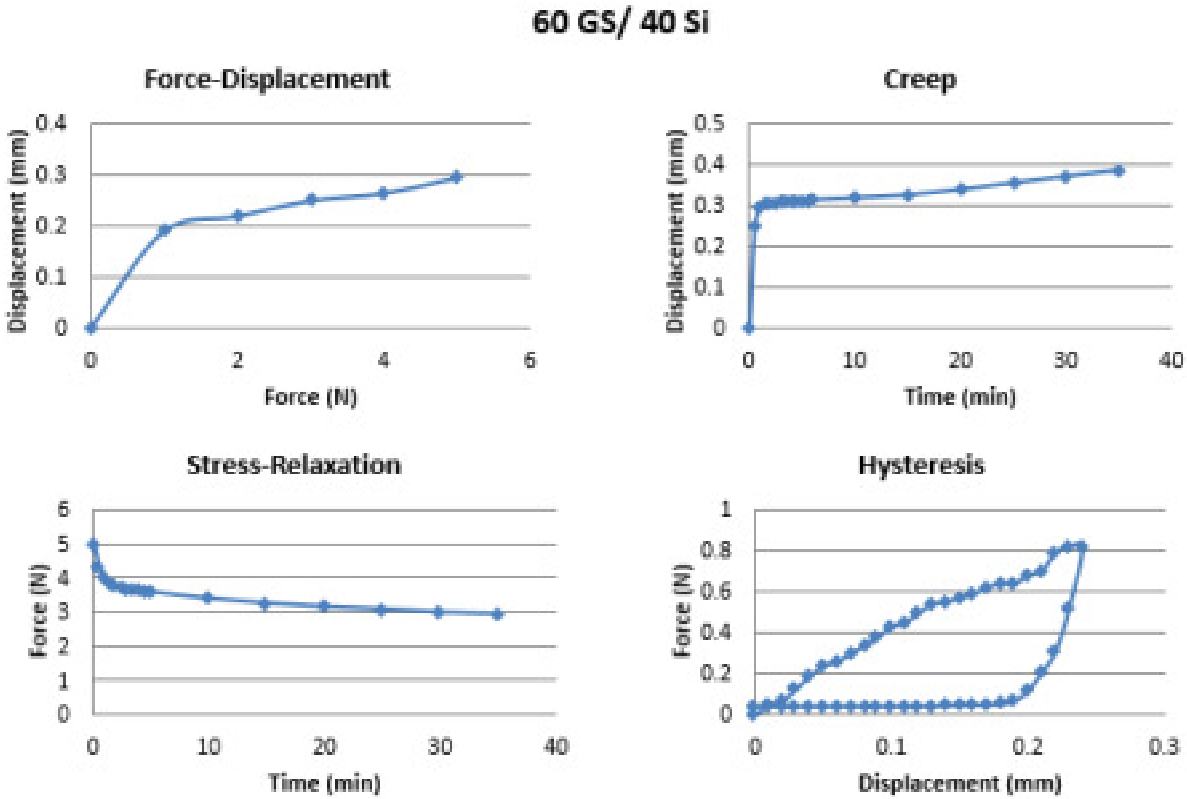

Figure 4.2. Plots of the viscoelastic properties of the 60/40 mixture

The 60/40 mixture also displayed viscoelastic material properties shown in Figure 4.2. The force displacement curve was non-linear. Creep displayed a very steep viscoelastic phase followed by a plateau. Stress-relaxation showed that within the first two minutes, the force droped to a minimal value, which continued into a more gradual decrease as time progressed. Hysteresis showed two distinct curves, reaching a maximum force just above 0.8 N.

The 50/50 mixture maintained viscoelastic material properties as displayed in Figure 4.3. The force displacement curve was non-linear. Creep displayed a very steep viscoelastic phase within the first minute followed by a plateau with a slight

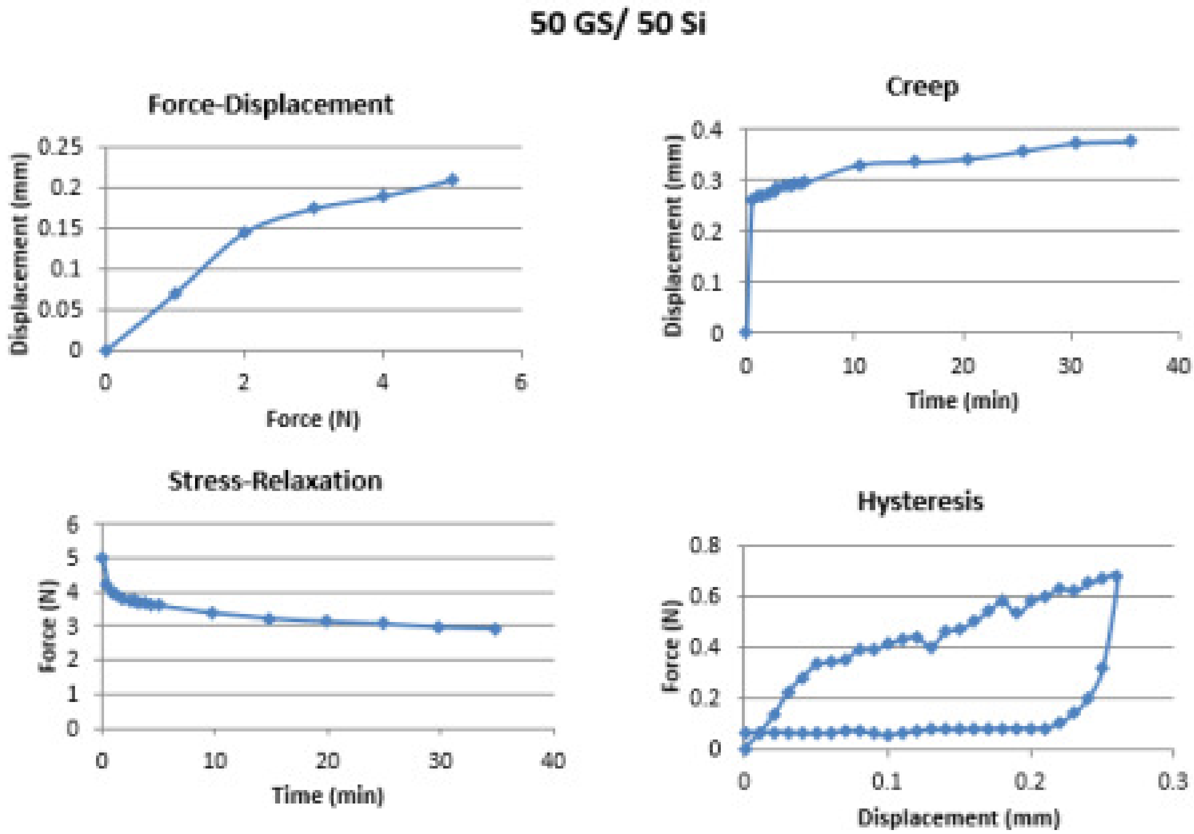

Figure 4.3. Plots of the viscoelastic properties of the 50/50 mixture

upward slope. Stress-relaxation showed that within the first minute, the force drops approximately 1 N and continues into a more gradual decrease as time progresses, ending at 3 N. Hysteresis displayed two distinct curves; the loading was a gradual increase reaching a maximum force of about 0.7 N while the unloading was a quick drop off and reached a minimum value before 0.1 mm was removed.

The 40/60 mixture showed viscoelastic properties as plotted in Figure 4.4. The force displacement curve displayed two phases, both of which were non-linear. Creep displayed a very steep viscoelastic phase within the first minute followed by a plateau with a slight jump to another plateau. Stress-relaxation showed that within the first minute, the force drops about 0.5 N and continues into a gradual decrease as time

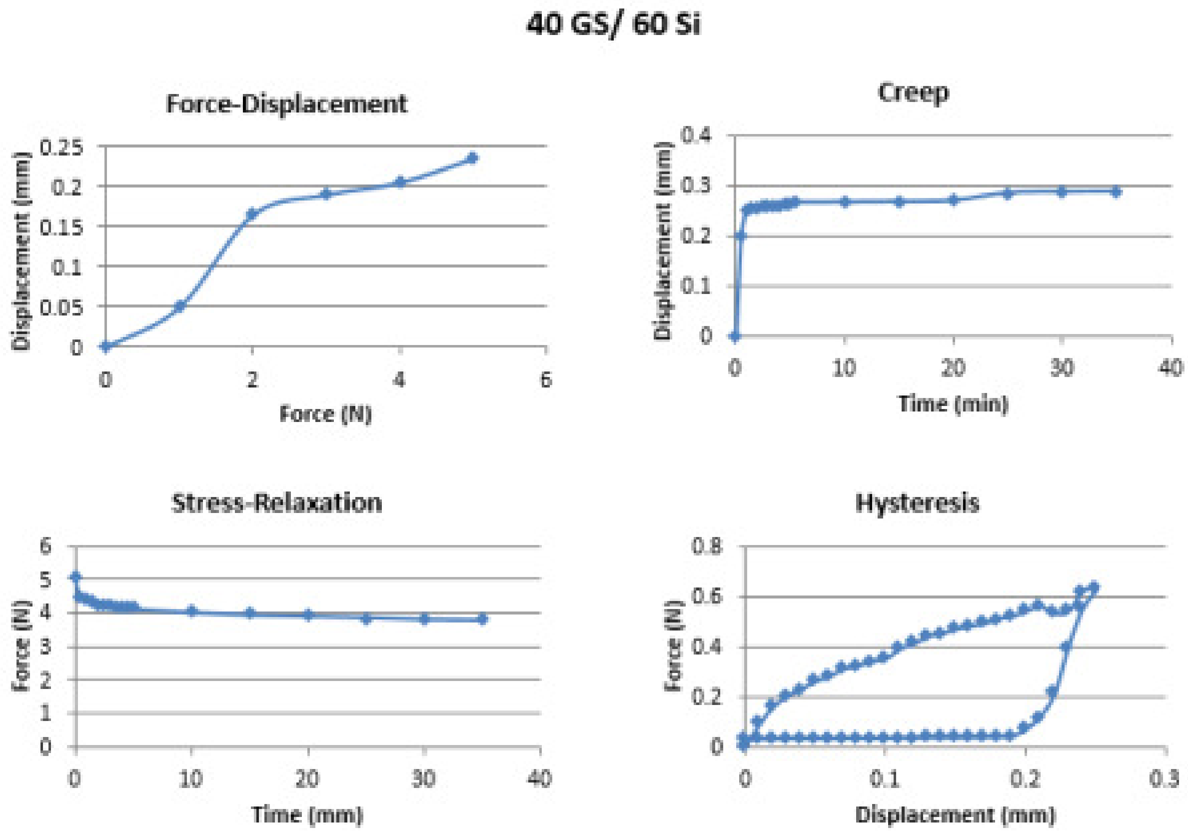

Figure 4.4. Plots of the viscoelastic properties of the 40/60 mixture

progressed, ending just under 4 N. Hysteresis displayed show two distinct curves, the loading was a gradual increase reaching a maximum force of about 0.6 N and displayed a slight shift in force values. The unloading was a rapid decrease of force and reached a minimum value with the removal of 0.05 N.

The 30/70 mixture did not display full viscoelastic properties as it can be seen in Figure 4.5. The force displacement curve displayed non-linear properties, the initial increasing curve, and linear properties, a slope after the initial curve. Creep displayed a very steep viscoelastic phase within the first minute followed by a short plateau, which then lead to an increasing curve. Stress-relaxation showed that within the first minute, the force droped about 0.5 N and continued into a gradual decrease

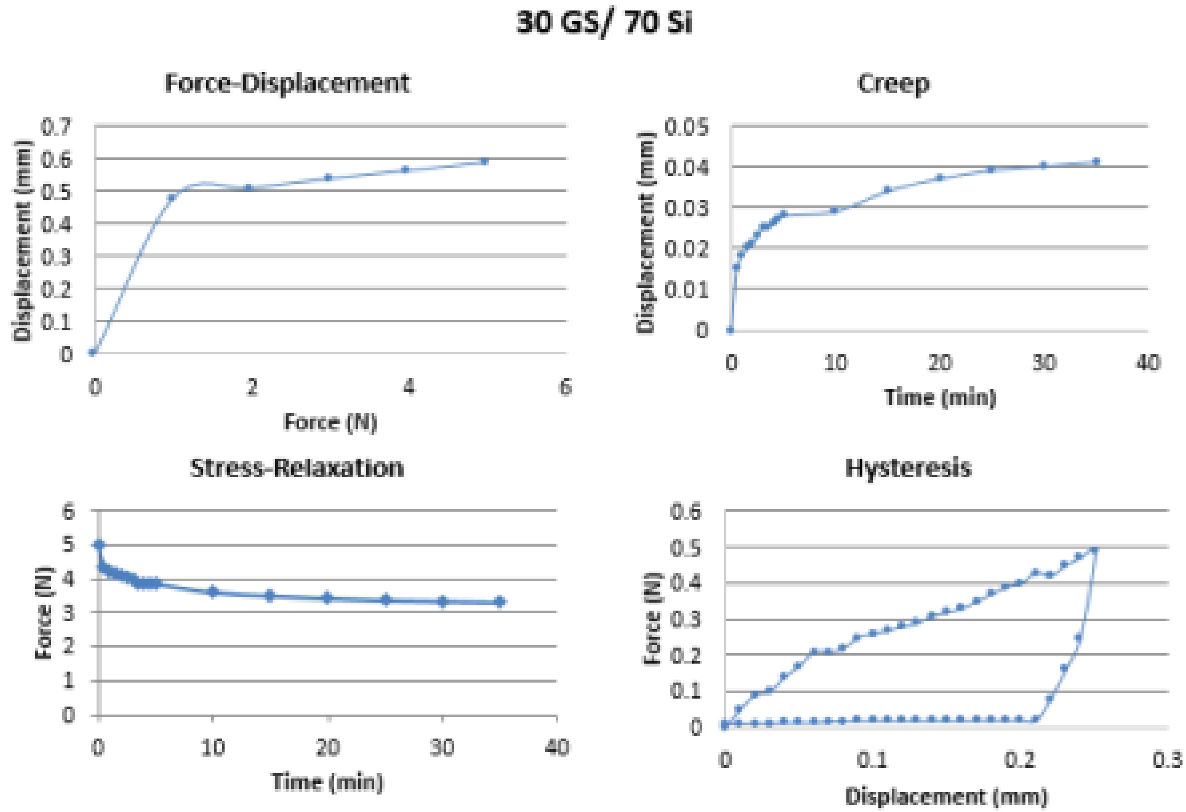

Figure 4.5. Plots of the viscoelastic properties of the 30/70 mixture

as time progressed, ending just above 3 N. Hysteresis displayed two distinct curves; the loading was a gradual increase reaching a maximum force of about 0.5 N and displayed a slight shift in force values at about 0.2 mm. The unloading was a rapid decrease of force that was nearly instantaneous and leveled out to nearly 0 N.

4.2 Testing Conditions - Time Study

The central incisor was the first tooth tested. The averaged Reset and Non-Reset displacement values were graphed.

The Reset plots showed distinct jumps in displacement values where at one time period displacement is low and at a following it may be high again which is shown in

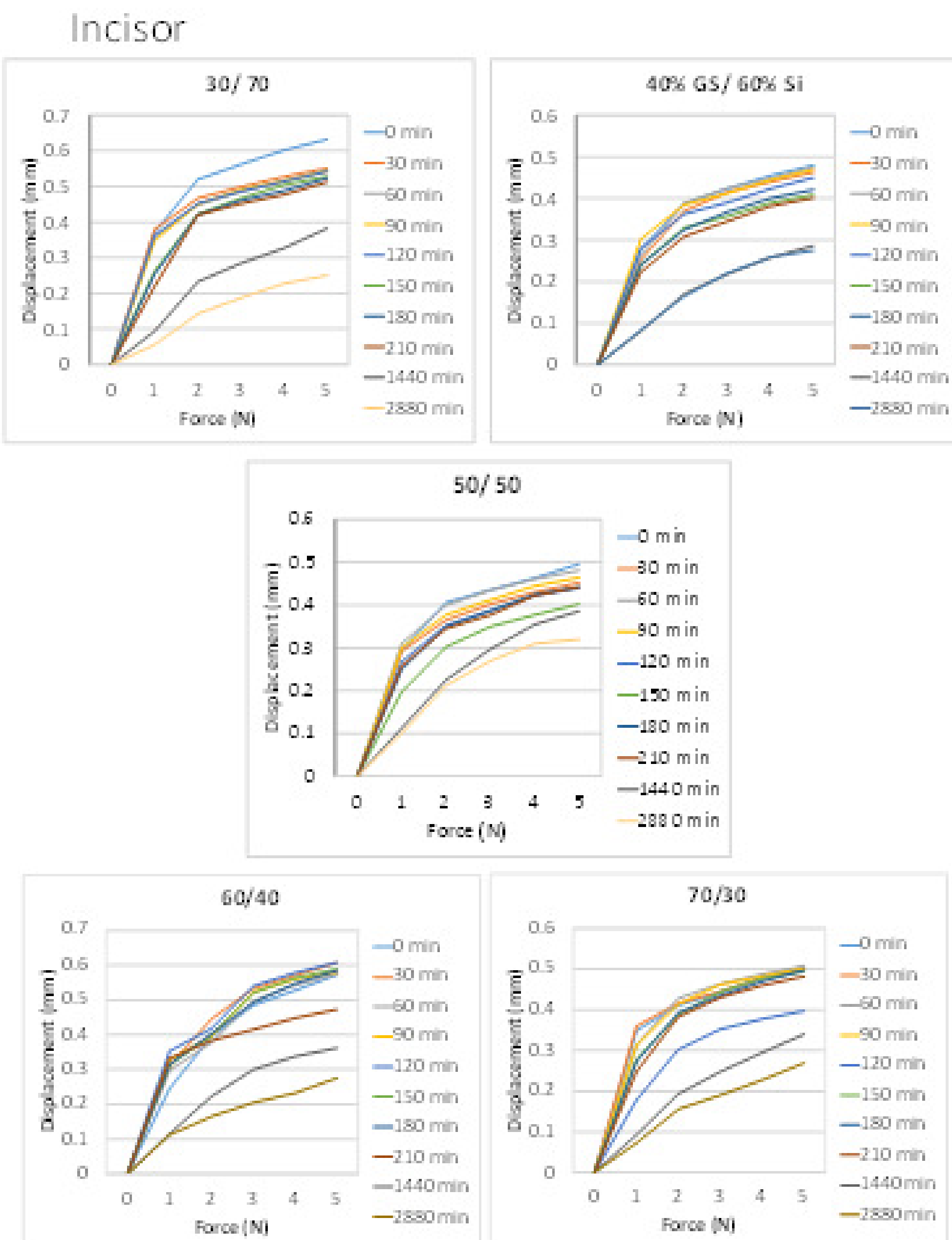

Figure 4.6. Displacement values of the Reset Experiments performed on the incisor

Figure 4.6. The smallest total displacements did not occur until after 24 hours and were smaller yet at the 48-hour period, meaning increasing stiffness. The 70/30 and the 30/70 mixtures experienced the lowest displacements.

The Non-Reset values were lower than the Reset counter parts. The final displacements of each mixture were also nearly half that of the Reset values as it can be seen in Figure 4.7. The differences between each at 48 hours were around 0.01 or 0.02 mm. Larger displacements were consistently found in the 60/40 mixture. The 40/60 mixture experienced the lowest displacements until the 24-hour mark. The 50/50 mixture was the next lowest earlier in the experiments and maintained a lower displacement throughout the entire experiment. Looking specifically at the 48-hour mark, the 50/50 mixture and the 30/70 experienced similar displacements with the 50/50 obtaining lower values until 5 N is reached.

4.3 Mechanical Response - Instantaneous Force

The Incisor's instantaneous force-displacement response was tested. Figure 4.8 shows the data used to display force responses.

The initial instantaneous tests were inconclusive as shown in Figure 4.8. The 60/40, 40/60, and 30/70 force readings did not come close to the accepted true value of 5 N [2]. These values of all the mixtures were at best half of the desired values.

Table 4.1. Maximum force obtained and the error between experimental and clinical values for a 0.05 mm displacement

Incisor Initial 0.05mm Displacement

Mixture(GS.Si)	Max Value(N)	Error
30.70	2.51	-0.50
40.60	2.56	-0.49
50.50	1.71	-0.66
60.40	2.12	-0.58
70.30	1.13	-0.77

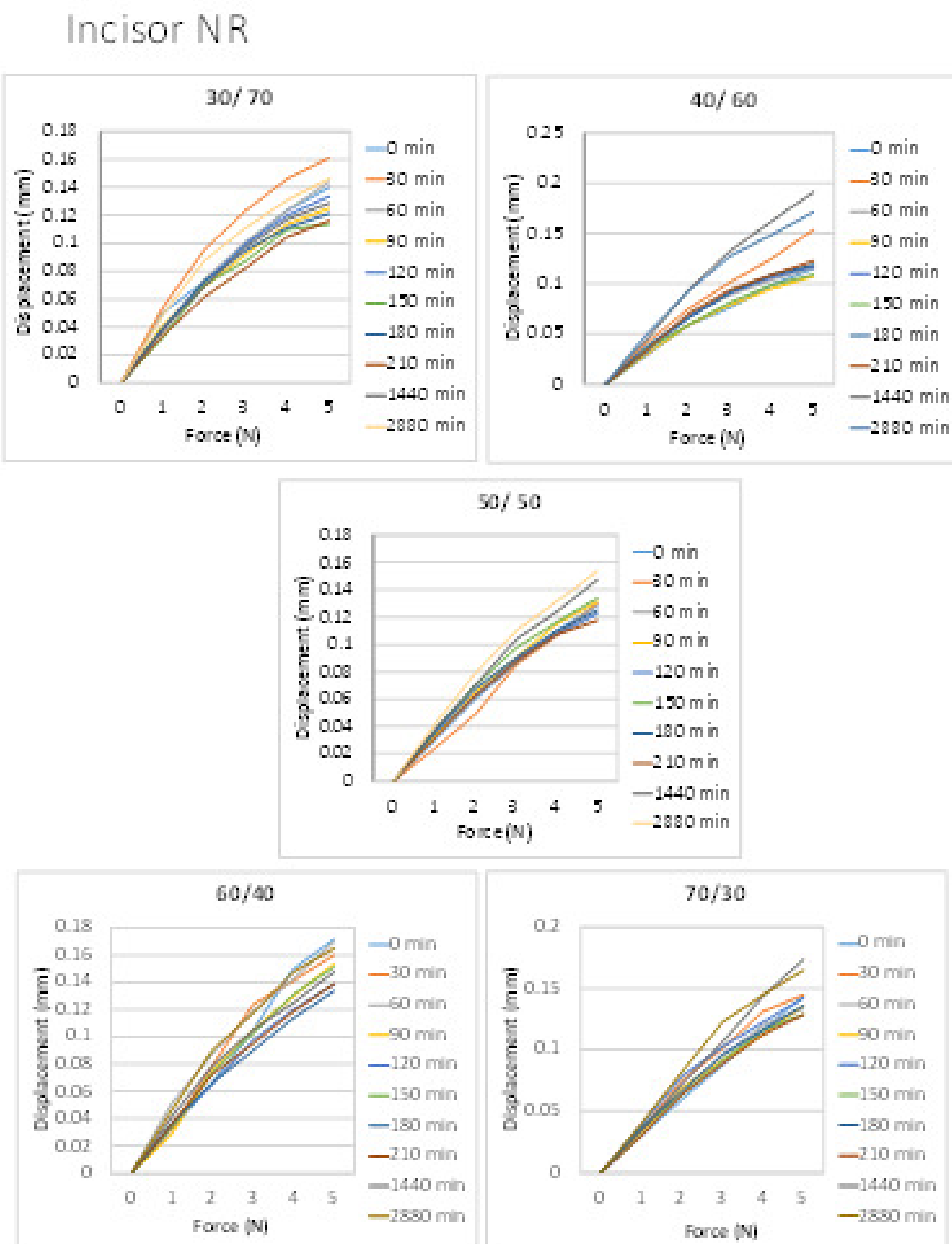

Figure 4.7. Displacement values of the Non-Reset Experiments performed on the incisor

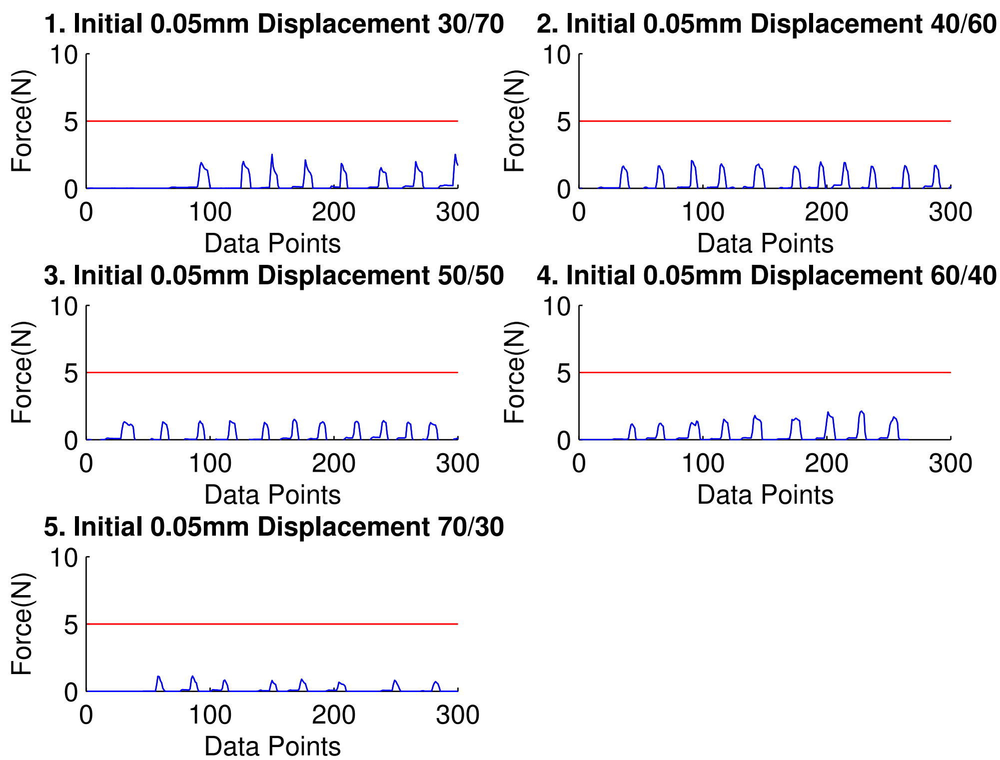

Figure 4.8. Instantaneous Responses of Five Mixtures at the Beginning of Testing for a 0.05 mm Displacement

The maximum value obtained was 2.56 N for the 40/60 mixture as shown in Table 4.1. Additionally, the lowest error was of 49% for the 40/60 mixture.

Much like the 0.05 mm displacements, none of mixtures achieved the true value of 10 N as shown in Figure 4.9. The values obtained were half of the expected value. These readings, therefore, were also determined to be inconclusive.

The 50/50 mixture produced a maximum value of 8.05 N and had the smallest error with of 33% as shown in Table 4.2.

At the 48-hour period, all the materials showed force readings at least half the expected value of 5 N. The 50/50 mixture was closest to this value and was able to reach it on the last reading recordeded as shown in Figure 4.10.

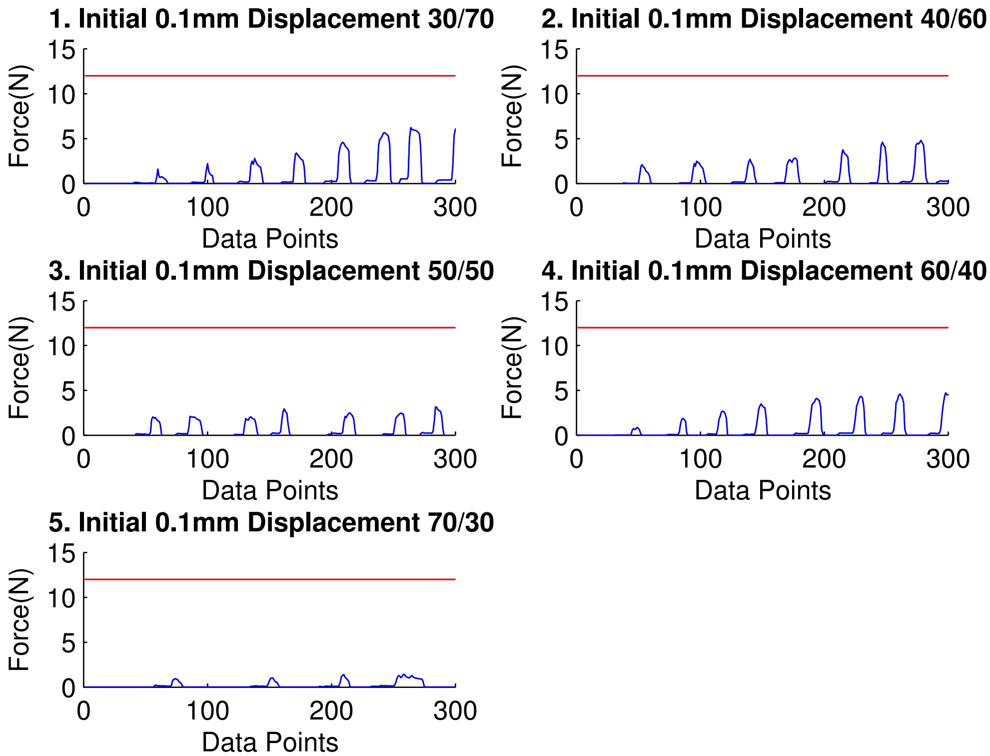

Figure 4.9. Instantaneous responses of five mixtures at the beginning of testing for a 0.1 mm displacement

Table 4.2. Maximum force obtained and the error between experimental and clinical values for a 0.1mm displacement

Incisor Initial 0.1mm Displacement

Mixture(GS.Si)	Max Value(N)	Error
30.70	6.23	-0.48
40.60	5.45	-0.55
50.50	8.05	-0.33
60.40	4.73	-0.61
70.30	3.08	-0.74

The 50/50 mixture obtained the highest maximum value of 4.8 N as shown in Table 4.3. This correlates to an error of 4%.

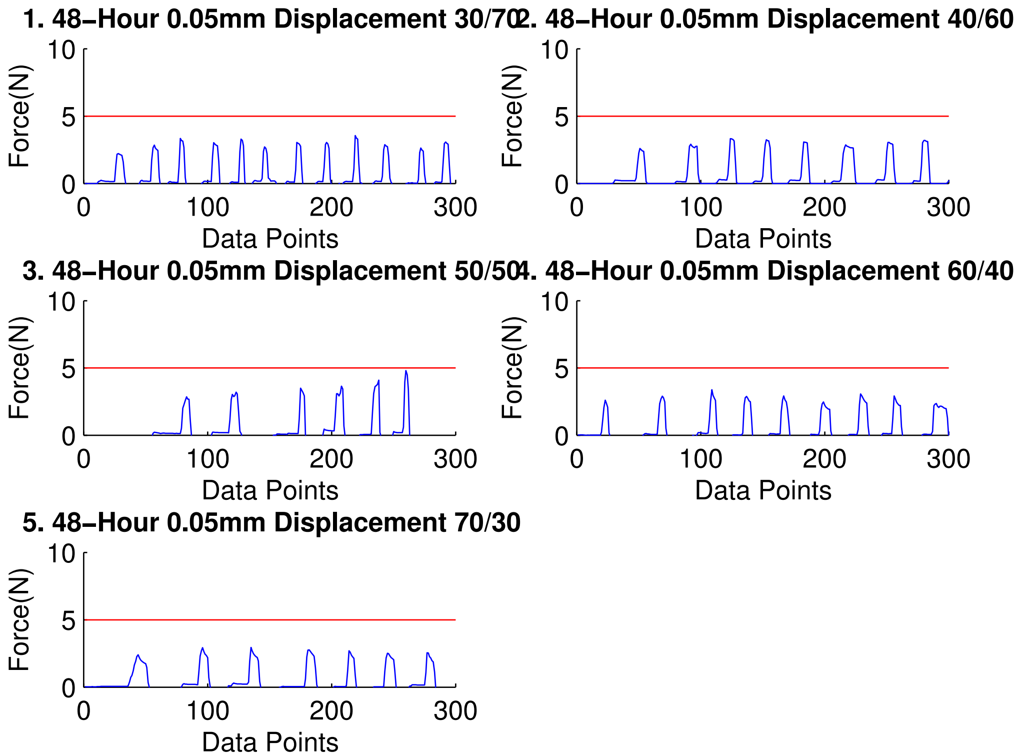

Figure 4.10. Instantaneous responses of five mixtures at the end of testing for a 0.05 mm displacement

Table 4.3. Maximum force obtained and the error between experimental and clinical values for a 0.05 mm displacement at the end of testing

Incisor 48-Hour 0.05mm Displacement

Mixture(GS.Si)	Max Value(N)	Error
30.70	3.65	-0.27
40.60	3.39	-0.32
50.50	4.80	-0.04
60.40	3.38	-0.32
70.30	3.47	-0.31

The materials undergoing a 0.1 mm displacement were not able to achieve the expected value of 10 N as shown in Figure 4.11. The closest was the 50/50 mixture reaching about 9 N. All other values reached a maximum of 5 N.

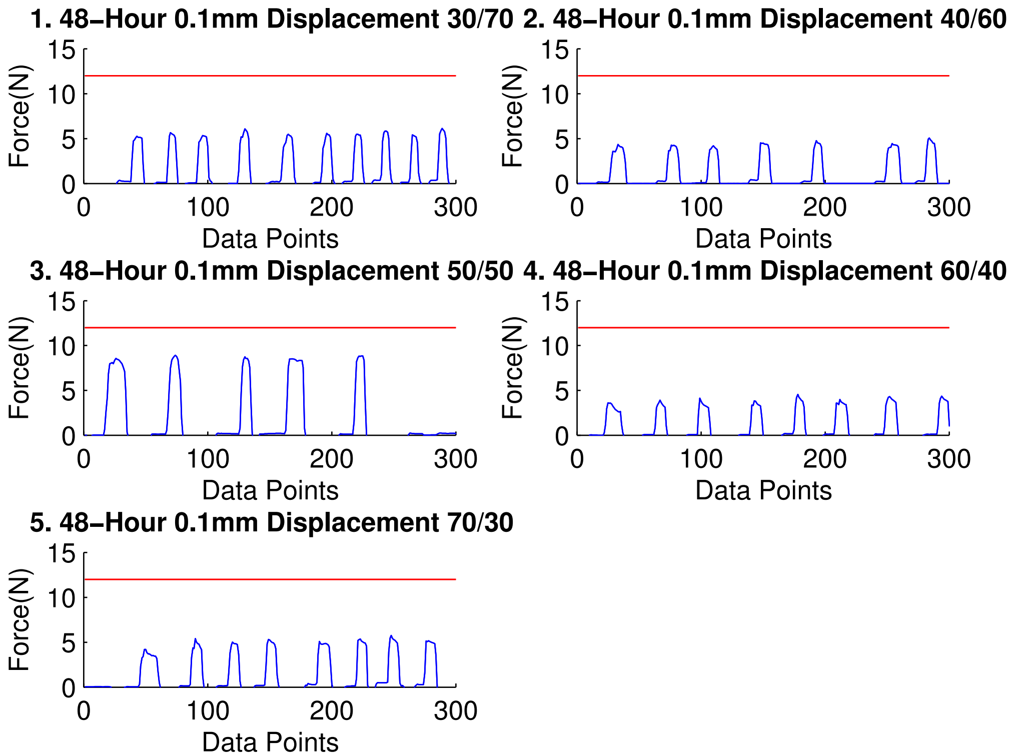

Figure 4.11. Instantaneous responses of five mixtures at the end of testing for a 0.1 mm displacement

Table 4.4. Maximum force obtained and the error between experimental and clinical values for a 0.1 mm displacement at the end of testing

Incisor 48-Hour 0.1mm Displacement

Mixture(GS.Si)	Max Value(N)	Error
30.70	6.20	-0.48
40.60	5.50	-0.54
50.50	9.10	-0.24
60.40	4.55	-0.62
70.30	6.35	-0.47

The 50/50 mixture obtained the highest force value and lowest percent error with 9.1 N and 24% respectively as shown in Table 4.4.

4.4 Reproducibility - Canine

The canine was tested last. Just like for the incisor, the averaged Reset and Non-Reset displacement values and the standard deviations of the four trials were tabularized.

At the beginning of the experiments, the 70/30, 40/60, and the 30/70 experienced the lowest displacements as shown in Figure 4.12. After 48 hours, the 40/60 and 30/70 undergo the smallest displacements. Throughout the entire experiments, the 40/60 undergoes the lowest displacement with the 30/70 consistently experiencing the second lowest.

The Non-Reset displacements for the canine, much like the incisor, were at least half or lower that of the Reset values as shown in Figure 4.13. At the beginning of the experiment the 60/40 mixture and the 50/50 mixture underwent the smallest displacements. At the end of the experiment the 70/30 and the 30/70 mixture underwent the smallest displacements. After 48 hours, all mixtures attained the 1 N reading at a closer distance than before, but some, namely the 60/40 and the 40/60, had to displace greater to achieve 5 N.

Utilizing displacement data from the Incisor and Canine time experiments, the average percent change in displacements can be assessed as shown in Table 4.5. Ideally, a trend in how a mixture composed of more of either the Si or the GS would have been evident, signifying how the materials change in stiffness occurs, however this trend was not observable. For instance, the 40/60 mixture had the lowest percent change of all the mixtures for the Canine Non-Reset experiment, but the highest in the Incisor Non-Reset by a large margin. The change in displacement appears to be random and significant among the tooth tested and the mixture used. What this in-

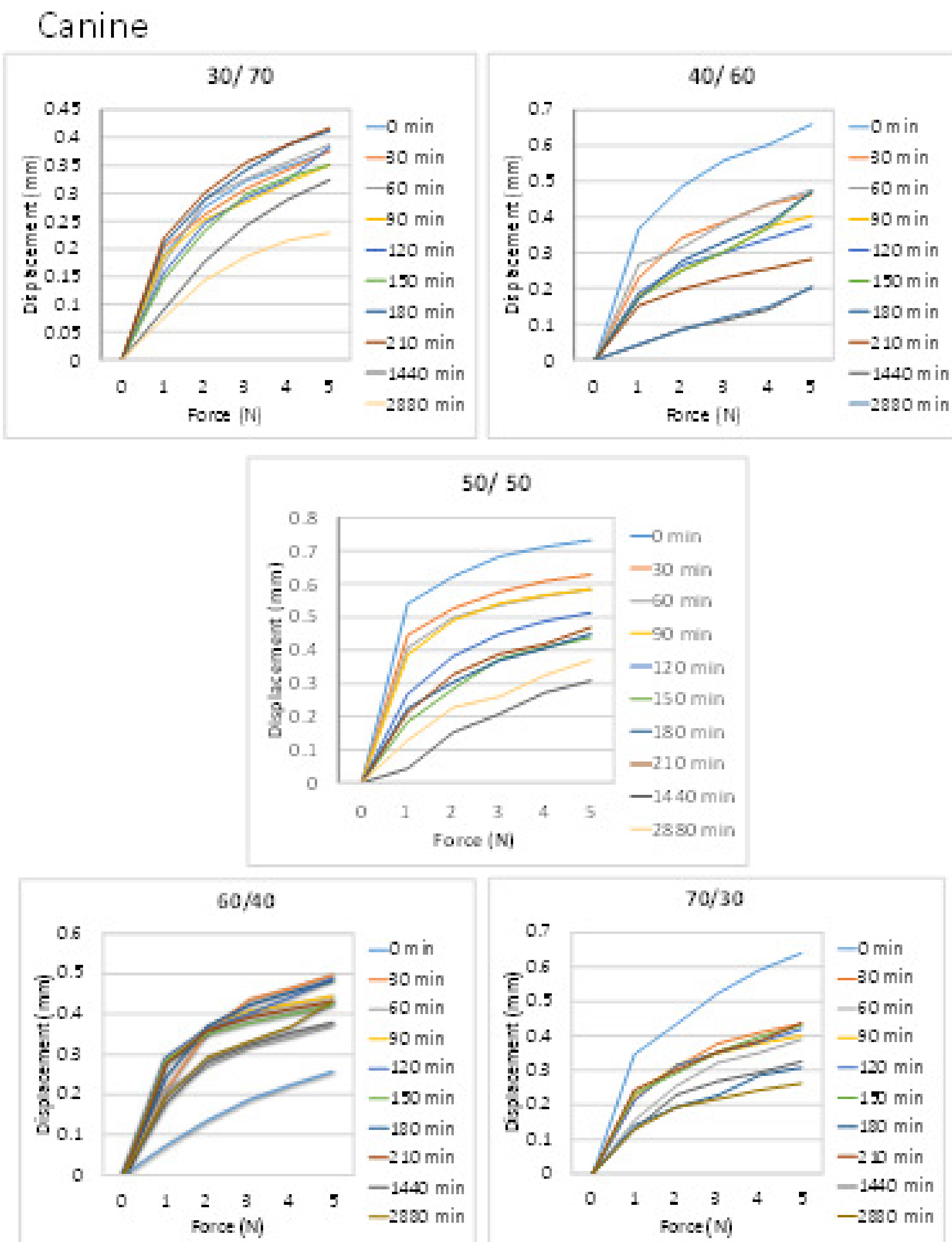

Figure 4.12. Displacement values from the reset experiments performed on the canine

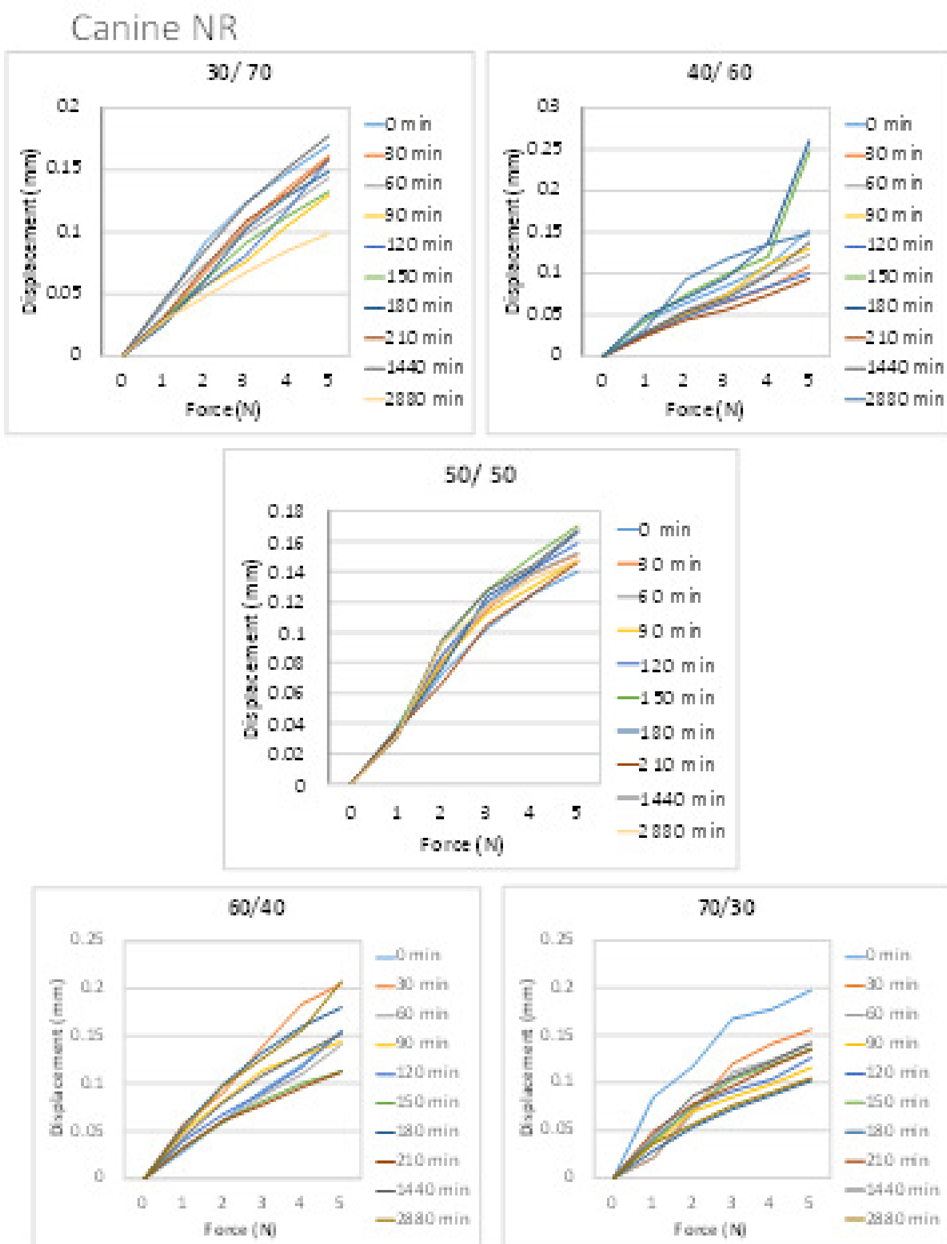

Figure 4.13. Displacement values from the Non-Reset Experiments performed on the canine

Table 4.5. Average percent change in displacements

Average Percent Change Through Experiment				
Mixture	Incisor	Incisor NR	Canine	Canine NR
70/30	46.91%	14.78%	144.29%	46.84%
60/40	51.97%	3.65%	40.75%	33.87%
50/50	35.52%	19.42%	49.32%	5.36%
40/60	42.71%	48.91%	69.32%	3.28%
30/70	60.28%	4.46%	39.20%	41.91%

formation displays is that there is not a discernable trend in how the various mixtures will behave during the curing process.

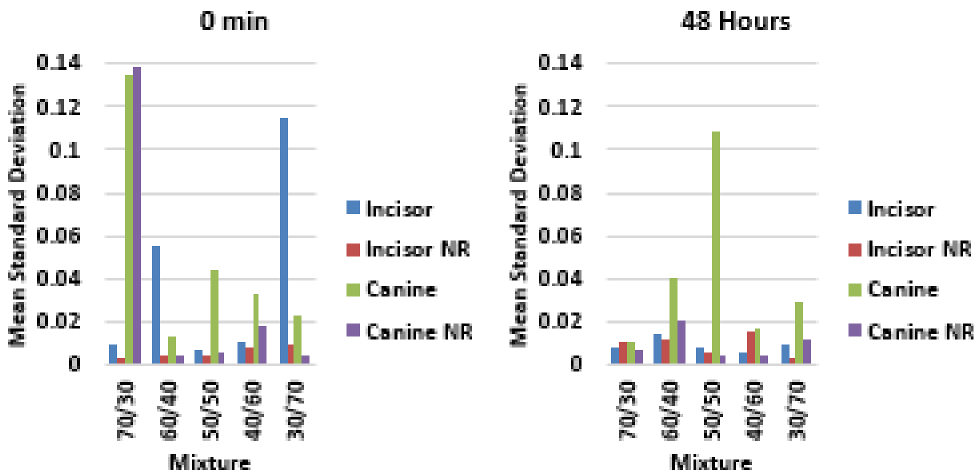

Figure 4.14. Plots for the standard deviations of each of the mixtures at the beginning (left) and the end (right) of the experiments.

Clear distinctions between the Reset, written with just the name of the tooth tested, and Non-Reset, denoted with "NR", values are evident in Figure 4.14. Assessing results of the canine and the incisor, the 60/40, 50/50, and 30/70 had the lowest standard deviation of their four trials at the start of the experiment. The Non-Reset values consistently produces lower standard deviation values in both the incisor and canine.

The mean standard deviation of the incisor reveals that the 30/70 and the 50/50 experience the smallest deviation after 48 hours as shown in Figure 4.14. At the initial reading for the Non-Reset trials for the incisor, the 70/30 and the 50/50 mixture displayed the lowest mean standard deviation while the 30/70 showed the highest.

Initially the 30/70 mixture is arguably the most consistent along its four trials for the canine with the lowest mean standard deviation of the entire experiment occurring at the beginning of the experiment as shown in Figure 4.14). Inspection of the standard deviation values after 48 hours reveals that the 50/50 and the 40/60 mixtures contain the lowest standard deviations throughout their trials. The mean standard deviation of the canine revealed the 50/50 and the 40/60 mixtures to have the lowest value after 48 hours as apparent in Figure 4.14. The 50/50 and 40/60 have very similar results for the Non-Reset experiments, however the 50/50 mixture remains one of the mixtures that shows the lowest deviations in the Non-Reset incisor and Non-Reset canine.

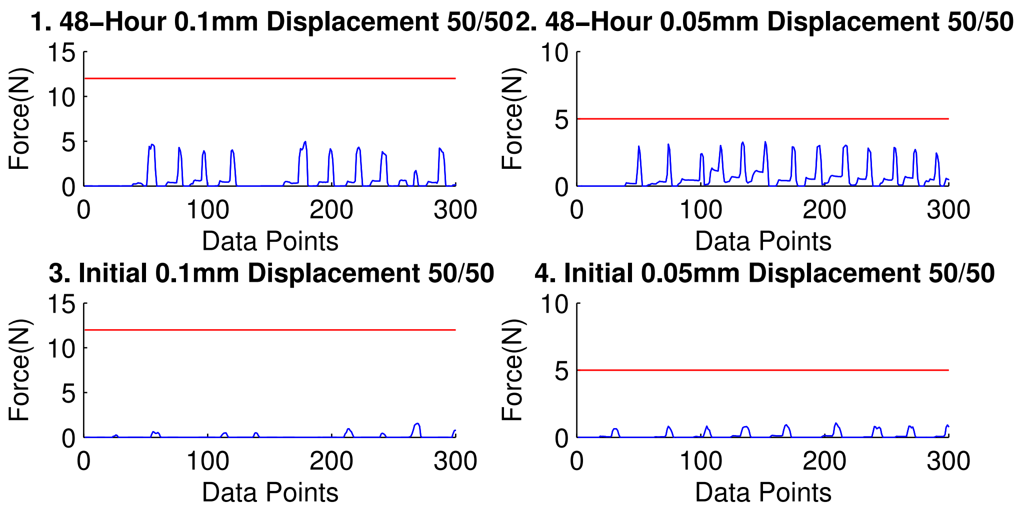

Figure 4.15. Instantaneous responses of the 50/50 mixture at the beginning and end of testing for 0.05 mm and 0.1 mm displacements

Instantaneous testing revealed similar results to that of the incisor as displayed in Figure 4.15. The 0.1 mm displacement after 48 hours was only able to reach 5 N, half the expected value. The 0.05 mm displacement after 48 hours was very close to reaching the 5 N benchmark. Initial tests were inconclusive and barely reached 1 N.

Table 4.6. Maximum force obtained and the error between experimental and clinical values at given times for 0.05 mm and 0.1 mm displacement

50/50 Canine Displacement

Time(hour)	Distance(mm)	Max Value(N)	Error
48.00	0.10	5.88	-0.51
48.00	0.05	4.03	-0.19
0.00	0.10	3.55	-0.70
0.00	0.05	1.56	-0.69

The 50/50 mixture of the canine produced a relatively low error of 19% after 48 hours as it can be seen in Table 4.6. A decrease in error is evident after the 48-hour period as compared to the initial 0 hour mark.

5. DISCUSSION

5.1 Discussion - Viscoelasticity

To begin testing, the viscoelasticity of the materials needed to be confirmed when used with the denture mold. The properties investigated were force-displacement, stress-relaxation, creep, and hysteresis. The average of three trials of each viscoelastic property was found.

Each mixture displayed viscoelastic properties. Force vs. Displacement data of the 5 mixtures each produced similar curves. The 50/50 mixture however had the smallest displacement of about 0.2mm. This value was reported by the previous study as the maximum distance a tooth should displace, for it is the thickness of a PDL [2].

The Force vs. Time curve established the stress-relaxation response of the tested tooth. The 70/30 displayed a drastic and sudden drop in force while each of the other mixtures displayed a more gradual decay. The curves each relaxed to a different asymptote, showing distinctions in material stiffness.

Creep was found by plotting displacement with respect to time. The curves obtained by the 60/40, 50/50, and the 40/60 each produced plots comparable to the viscoelastic phase and following plateau that Jones and researchers reported when subjects' teeth were exposed to continuous load. The 70/30 and the 30/70 produced questionable viscoelastic phases that showed jumps in displacement. Additionally, the 30/70 mixture shows a linear portion to its force-displacement curve.

Jones et al. reported a recovery phase that resulted in a lingering displacement of 23% to 30% of the maximum value [24]. The hysteresis curves created found a similar phenomenon in which the tooth did not return to its initial position. The remaining displacements were about 2% of the maximum value, much smaller than the percentage reported by Jones, however a much larger force was applied in this study. The hysteresis for the mixtures, excluding the 60/40 mixture, was much larger than previously reported by Xia [11]. With the exception of the 60/40 mixture, the force drops nearly immediately to the lowest value. It is unclear if this was the tooth's rapid return to its position or the tooth not returning at all, which would mean that the lingering forces are noise from the transducers.

The results in viscoelasticity do prove that the materials created display viscoelastic properties, however they do not align with the findings in the previous ATPBC study. Of these the force-displacement curves are the most important since sliding mechanics focuses on the tooth retraction in response to varying force levels. The sockets of the denture mold are also of a different geometry than the ATPBC used in the study by Xia [11], which contributes to the differences in findings.

5.2 Discussion - Time Study

The goal of this study was to determine the mechanical response of the various compositions of the APDL over time. Observing the values as time progressed proved to be a fitting endeavor. For all the materials, the maximum displacement over time decreased, meaning stiffness increased. In addition, the results in each material composition became more consistent across the four trials as time progressed, which is what was desired.

Consistency was easily observed by monitoring the changes in standard deviation of the averages of the four trials at each time point for every level of force. After 210 minutes, the standard deviation of materials for the incisor and canine decreased significantly. After 48 hours, the standard deviation in almost every material was at its lowest. Of these the 50/50 mixture consistently produced one of the lowest standard deviations at each level of force. While there are times were other mixtures have a lower standard deviation at certain forces, the 50/50 is amongst the lowest more frequently.

These experiments set out to determine which material produces the most consistent results and when this occurs. This section of the study revealed that time is an important factor for testing with the APDL. A 48-hour period is necessary for the materials to set and reach their nominal mechanical responses. It is therefore part of the testing protocol that the material is allowed to set for 48-hours before testing may begin.

An addition to the protocol for future study stems from the Non-Reset versus the Reset results. It is the recommendation that a Non-Reset method is used during testing. For longitudinal studies, this will not be an issue. Orthodontic devices will be mounted on the mold, applying small and constant loads on the teeth. There should not be a reason to move the tooth in any further capacity aside from the forces being applied, which in turn further rules out a Reset method. Also, when considering the biological environment, a tooth gradually resets back to a neutral position. There is no need to forcefully reset the tooth unless the roots and surrounding tissues have been damaged. Therefore, if any force-displacement readings are taken prior to a longitudinal study, it is imperative that the tooth be left in the position it stops at.

5.3 Discussion - Instantaneous Response

The inability for any of the materials to reach an adequate force level in the initial instantaneous readings further showed that the materials must settle prior to their use. Mixtures tested immediately produced only fractions of the expected force values. After 48-hours each material was able to achieve a force reading, however the 50/50 mixture once again was shown to be the ideal choice.

The instantaneous force plots show that the 50/50 is the closest to achieving the accepted true values of 5N for 0.05mm displacements and 10N for 0.1mm displacements as was found in the clinical studies.

The method of testing employed in these experiments is most like that of literature sources in which clinical studies were conducted [2]. While the previous experiments further defined the characteristics and limitations of the mixtures [11], this experiment truly revealed which mixture is closest to that of the biological tissue. This in fact elevates the 50/50 mixture from not just being the most consistent, but to being the most like a human periodontal ligament.

The goal of these experiments was to determine the instantaneous response of the materials. Readings were taken at the initial concoction of the mixtures and again after 48 hours. The ATI softwares built in record function allowed for the focus of maintaining as close to possible displacement velocity and as close to possible displacement limits. Fixing the displacement allows for the direct analysis of the force readings and a fixed velocity eliminated sources of extraneous noise factors.

The plotting of the collected values allowed for easy assessment of the force levels achieved by the various mixture. Not all of the maximum values obtained were viewable in the window size chosen, however these values were still tabularized and the percent errors accounted for. The initial readings produced results that were too far deviated from that of the literature. The 48-hour readings produced results closer to the expected value. Only in the 50/50 mixture were the desired values obtained, highlighting it as the ideal mixture as far as the instantaneous response is concerned. The 50/50 mixture also able to produce the lowest percent error in three out of the four phases, the closest of which was after 48 hours for a .05mm displacement. This established that the 50/50 mixture has the highest potential of simulating human tooth displacement.

5.4 Discussion - Reproducibility

Testing on the incisor established that the 50/50 mixture is the best choice for use with the denture mold. The final task was to see if these findings could be expanded to include other teeth. The canine was the focus for this testing. It provides geometry that differs enough from the incisor that it was expected to produce unique results.

The time study of the canine produced results comparable to that of the incisor. After 210 minutes, the standard deviation of materials for the canine decreased significantly, just like in the incisor. After 48-hours the materials produced more consistent results, of these, the 50/50 mixture was the lowest in standard deviation.

The results of the Non-Reset versus Reset experiments in both the canine and incisor varied as well. A side-by-side comparison shows Reset values of the maximum displacement after 48 hours were at least double and at times tenfold that of the Non-Reset values for each of the mixtures. The thought is that resetting the artificial tooth after each trial caused additional breakage in the APDL materials. This breakage would have been detrimental during the settling time and may have caused voids in the material. The jumps in the displacement results suggest that the materials fractured due to the forced return to the starting positions. This would have contributed to the tooth's ability to displace further. The Non-Reset proved to be the best choice when approaching in-vitro testing.

Throughout testing the 50/50 mixture was the most consistent with the lowest standard deviation through the four trials at each time interval for both teeth during the Non-Reset time experiments. The consistency of results gives credence to this mixture being the ideal mixture for orthodontics testing in the ATPBC. The 50/50 mixture was also determined to be the best composition in the previous ATPBC study [11]. The aim was to observe which is the most consistent; therefore the 50/50 mixture again proves to be the best choice.

Looking purely at the displacement results, both the incisor and canine seem to displace further overtime. This is seemingly counter-intuitive since the materials should be hardening and expressing a greater stiffness. The answer to this conundrum was linked to the observation of starting positions. The starting positions increase in number, meaning that the micrometer traveled shorter distances to make contact with the tooth as time progressed. The teeth were "sinking" in the socket as time progressed. One may conclude that the initial force responses, since the displacement and starting positions were shorter, may have been more of a result of the tooth making contact with the back walls of the dentures socket. This highlighted a need to observe the change in total displacement. A higher change in total displacement could be indicative of the tooth making contact with the back walls of the socket.

The average percent change in displacement was the measurement of how the total displacement changed over the 48-hour period. The hope was to observe any possible trends in how the displacement changes as materials cure. The results yielded no observable trends, signifying that there is not a set way the materials will cure over time. Materials must set for 48-hours.

With the 50/50 mixture proven to be the best choice in composition, instantaneous testing with this mixture in the canine would prove that the material could be used with more than one tooth. Instantaneous testing of the canine confirmed the 50/50 mixture produces results comparable to the human PDL. The percent error was larger than that of the incisor 50/50 mixture, but still lower than all of the other mixtures. The larger error can be attributed to the results being compared to tooth mobility of the incisor. The change in geometry has been proven in this study and in a previous clinical study by Goellner [10] to effect displacement and may vary between teeth. Additionally, the force readings 0.05mm displacement aligns with the findings of the study by Goellner for tooth displacement of the canine exposed to 6 N. This finding, however, was performed using a different method of force application (force was applied on the lingual surface as opposed to the labial), making a solid comparison difficult to make. It is nonetheless established that the 50/50 mixture is the best choice for an APDL.

6. CONCLUSION

6.1 Summary

Current orthodontics research that capitalizes on the use of sliding mechanics is limited in that there is not a functional periodontal ligament in the dental model. The ligament is paramount in the native environment for absorbing occlusal loads. This viscoelastic tissue must be accounted for when assessing the forces necessary to influence tooth movement.

This study set out to validate claims of an appropriate combination of gasket sealant No. 2 (GS) and RTV 587 silicone (RTV) materials that could suffice as an artificial periodontal ligament (APDL). First, the viscoelastic properties of the materials were determined. Next, the proper conditions in which testing should occur needed to be established. This was judged on the consistency of responses to orthodontic loading. Third was the selection of the best composition of materials for the APDL. Selections were based on the statistical analysis of the force-displacement results throughout the experiments and comparison to clinical studies.

Materials went through a sequence of tests of force-displacements readings. The focus of which was not just testing various combinations of the materials, but how these materials behaved over time. The end of the experiment was defined as the point in which the materials' standard deviation was at their lowest. The data from these findings were then analyzed to find the standard deviations of multiple trials at various time points, the change in mean standard deviation from the start of the experiment to the end, and the percent change in final deformation as compared from the start of the experiments until the end of the experiments. Finding an appropriate mixture as well as the point in which the materials mechanical responses became consistent would serve as proof as to the validity of the materials usefulness.

The materials' responses varied greatly until after a 48-hour period, establishing a need for a curing period to allow for materials to set over this time period. The most consistent results, defined as the lowest standard deviations at varying force levels, was the 50% gasket sealant and 50% silicone mixture. After 48 hours, the standard deviation at each level of force was lower than 0.005 for both the incisor and canine.

The final analysis that further contributed to the consistency in results was the percent change in final displacement. Comparing with increases and a few decreases in change of final displacement of both teeth, the 50/50 mixture was the most consistent. The 50/50 mixture was found to have the smallest increase of the materials for the canine, about 5% more, but a slightly larger increase for the incisor, about 19% more, after 48 hours, indicating an increase in elasticity. This elasticity is responsible for the materials ability to return back to its starting position and displace the same amount with each level of force.

The change in percent displacement was in fact extremely revealing in what the tooth is actually doing as the materials settle. There was an observed decrease in starting position, which is the micrometer had to travel less of a distance to make contact with the tooth as time progressed. This indicates that the tooth is shifting or "sinking" forward. The weight of the tooth is actually forcing the material around it to displace. While this will not affect the application in force in a singular direction and does not change the results of tooth displacement after 48 hours, this hints to the fact that the APDL may not be maintaining equal thickness around the tooth root. This also means that part of the force readings of the initial trials may be the tooth interacting with the denture mold as opposed to being the result of the APDL resisting the force.

The 50/50 mixture of the APDL was proven to produce the closest results to that of the human PDL. After 48-hours, displacements of .05mm aligned with negligible error within the incisor. The 50/50 mixture in the canine produced a slightly larger error, but when compared to the findings of the mixtures in the incisor, it was still produced the lowest percent error, signifying the ability for the findings to be extrapolated to other teeth.

6.2 Conclusions

A number of conclusions may be drawn from this study.

- Materials must cure for 48 hours prior to testing for increased reproducibility
- The 50/50 mixture is the best choice to simulate human tooth response
- The 50/50 mixture is applicable for use within all artificial single root teeth
- The denture mold with the 50/50 mixture is suitable for use as an ATPBC

A Protocol for testing is as follows:

- Gasket Sealant and RTV Silicon should be well mixed in a 1:1 ratio
- The APDL mixture will then be wrapped around the root of the teeth of the mold
- APDL prepared teeth should then be inserted into the appropriate sockets as completely as possible with excess material removed
- The teeth must settle in the appropriate tooth sockets of the mold for a minimum of 48 hours prior to testing
- ATPBC is prepared for testing

If displacement tests are to be conducted on the teeth:

- Each tooth must be preconditioned by loading to 5N prior to testing

- The load cell must be aligned properly to make contact with the surface of the tested tooth
- Adjust displacement until load cell barely touches the desired tooth (0.01N) and bias the readings

A relatively larger displacement was evident in the denture mold as compared to the native environment. This may be attributed to the wider socket size of the denture mold. With this in mind, the displacements observed cannot be made of concern, but more contribute to understanding the robustness of the materials. Instead, the consistency in which the material responds to force must be the main focus. In addition, the forces that were applied on the denture mold are approximately two fold larger than what is normally applied for orthodontic testing. Larger forces in the native environment are translated from the PDL to the surround alveolar bone, contributing to a smaller possible displacement. The larger force was applied purely to determine that the consistency in results held true. In this regard, the gasket sealant and silicon mixture is a prime choice for an APDL. The best combination of which is 50% gasket sealant and 50% silicon.

6.3 Future Research

Future research will benefit from the use of this APDL. This mixture can work to provide researchers with a better estimate of how tooth movement will progress under orthodontic loads. The expansion of this is deeply rooted in the testing models. A method of preventing the tooth from sinking as the materials set needs to be investigated. A method of applying displacement in which a controlled velocity can be applied would also eliminate potential noise factors. As said before, the surrounding alveolar bone majorly effects part of the displacement under high loading conditions. If the goal is to create a highly accurate model that simulates the entire tooth remodeling under mechanotransduction, a denture mold that is accurate in geometry, including teeth with multiple roots, and accurate in tissue density could be of better use. This however may still prove to be futile, as it has already been proven that part of tooth remodeling is contingent in biological factors. Simply put, it is highly difficult to account for every factor that will influence tooth movement in-vitro. The development and analysis of the APDL was the first step in developing a more accurate testing model for orthodontics research. Implementation is the next.

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